

1 X-Rays

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Introduction

X-rays routinely are used to non-invasively examine internal anatomy. The technique is called **Radiography**. The image created using x-rays is called a **radiograph**. Many people confuse the terms “x-ray” and “radiograph.” X-rays are a type of energy, whereas radiographs are images.

Radiography is a part of the medical specialty called *Radiology*. Radiology encompasses all of diagnostic imaging, including ultrasound, computed tomography (CT), magnetic resonance imaging (MRI), and scintigraphy. A major advantage of radiography over the other imaging modalities is that a large volume of the patient can be viewed with one image. A major disadvantage is that many structures are superimposed, making interpretation of the images sometimes confusing.

Our goal in radiography is to make images that display the greatest amount of usable information. This requires an understanding of x-rays; what they are, how they are made, and how they can be used safely to “look inside” our patients.

X-rays

X-rays were discovered in 1895 by Wilhelm Conrad Röntgen. Röntgen was a German scientist who won the first Nobel Prize in physics for his discovery (Figure 1.1). He used the symbol “X” to name the new rays because they were a type of radiation that was unknown at that time. Radiation is energy that spreads out in all directions as it moves away from its source.

X-rays are a type of electromagnetic radiation (EMR). There are many types of EMR, including radiowaves, microwaves, and visible light (Figure 1.2). EMR behaves both like particles and like waves, sometimes described as a stream of particles that travels in a wave-like pattern. EMR particles are called **photons**. Photons are packets of energy with no mass, no charge, and move at the speed of light. The energy of a photon is directly related to **wavelength**; the shorter the wavelength, the greater the energy. Wavelength is directly related to **frequency**; the higher the frequency, the shorter the wavelength (and the higher the frequency, the greater the energy).

In the EMR spectrum, x-rays are relatively short wavelength, high frequency, and high energy. They are able to penetrate many materials that are opaque to visible light (Box 1.1). X-rays also are **ionizing**, which means they have sufficient energy to remove electrons from atoms and form ions. Ionization can destroy molecules and damage or kill living cells. Adherence to radiation safety guidelines is crucial when working with x-rays (see the Radiation Safety section later in this chapter).

The reason we are able to use x-rays to view internal anatomy is because x-rays pass through some tissues more easily than others. Without differences in tissue transmission, radiography would not be possible. The ability of a tissue to

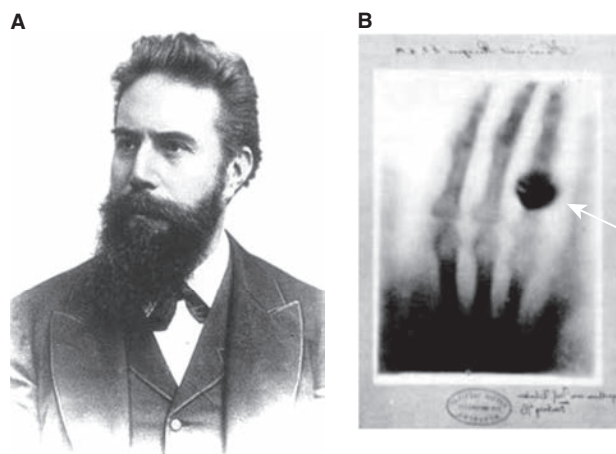


Figure 1.1 Discovery of x-rays. (A) Photo of Wilhelm Röntgen, the German physicist who discovered x-rays. Source: JdH/Wikimedia Commons/Public domain. (B) Photo of the first radiograph made by Röntgen. It is an image of his wife's hand. The ring on her finger is the first radiographic artifact (arrow). Source: NASA.

block or *attenuate* x-rays is called **radiopacity** or simply **opacity** (“radio-” is assumed since we are discussing radiography). The greater the opacity, the fewer x-rays will pass through.

Another key point in radiography is that a tissue or object will be visible in a radiograph only if it is adjacent to a material with a different opacity. There must be an **opacity interface**. An opacity interface is the boundary where two materials with different opacities meet. Without opacity interfaces there is no visible image in a radiograph. The greater the difference in opacities between materials, the stronger the opacity interface and the easier it will be to identify structures.

Radiography is possible only because x-rays penetrate some materials more easily than others. An opacity interface must be present to identify a structure in a radiograph.

Box 1.1 Properties of x-rays

- X-rays have no mass or physical form.
- X-rays cannot be detected by human senses; they cannot be seen, heard, or felt.
- X-rays travel in straight lines at the speed of light.
- The path of an x-ray cannot be affected by gravity, electrical fields, or magnetic fields.
- X-rays can penetrate many materials; the degree of penetration depends on the x-ray energy and the type of material.
- X-rays can expose photographic emulsion.
- X-rays are ionizing and can damage or destroy living cells.

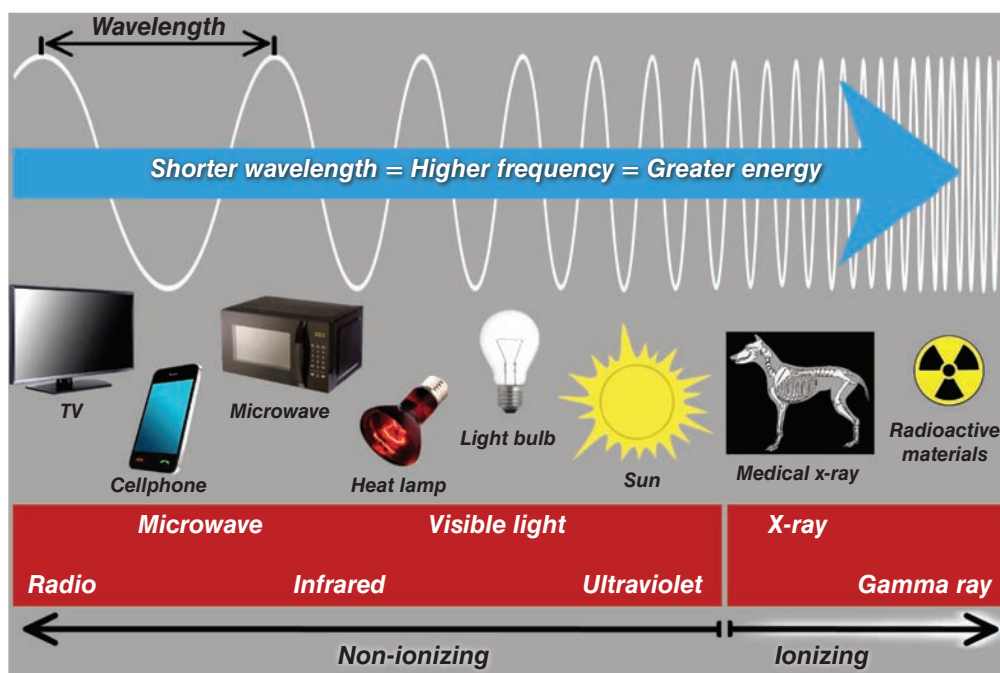


Figure 1.2 Spectrum of electromagnetic radiation (EMR). Different types of EMR are displayed in this image. The type is determined by the energy of the photons. The shorter the wavelength, the higher the frequency, and the greater the energy. High energy EMR can be ionizing. Source: trekandphoto/Adobe Stock.

As mentioned earlier, our goal is to produce radiographs that provide maximum usable information. Making quality radiographs requires adequate *density*, *contrast*, *detail* and minimal *distortion*. Each of these factors is discussed thoroughly later in this chapter. However, a brief introduction may be helpful here.

Density refers to the amount of blackness in the radiograph. The greater the radiographic density, the darker the image. Density is determined by the number of x-rays that are recorded by the image receptor. The more x-rays recorded, the darker that part of the radiograph. In most radiographs there are many levels of density. Each level is displayed as a shade of gray. Each shade of gray corresponds to an opacity. If there is too much density, the radiograph will be too dark. If there is too little density, the radiograph will be too light. In either case we won't be able to differentiate structures in the image.

Contrast refers to the amount of difference between the levels of density in the radiograph. It is the degree of change between the dark areas and the light areas. High contrast means the radiograph is mostly black and white with few shades of gray in between. Some structures will not be visible because there are no shades of gray to display their opacities. Low contrast means there are many shades of gray between white and black. If there is too little contrast, structures with similar opacities will not be distinguished because their individual shades of gray will be too similar and they will visibly blend together.

Detail refers to the sharpness of change from one density to the next in the radiograph. The greater the detail, the sharper the edges of the structures in the image. Detail – or definition – determines how well we can see the borders of

structures that are close together. If there isn't enough detail, we won't be able to distinguish the structures as separate.

Distortion occurs when the actual size or shape of an object is misrepresented in the radiograph. It results from poor positioning of the object in relation to the x-ray beam and image receptor.

X-ray production

For medical radiography, x-rays are created inside a glass vacuum tube. This is accomplished by bombarding a metal target with high-speed electrons. Inside the tube are a negatively charged **cathode** and a positively charged **anode** (Figure 1.3). The cathode contains a wire filament which can be heated with an electric current to “boil off” electrons (Figure 1.4). This process is called *thermionic emission* and it is similar to heating the filament in an incandescent light bulb.

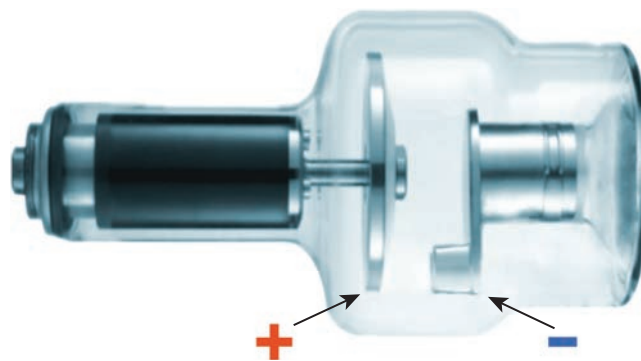


Figure 1.3 X-ray tube. This is a picture of a generic x-ray tube showing the positively charged anode (+) and the negatively charged cathode (–) enclosed in a glass vacuum tube.

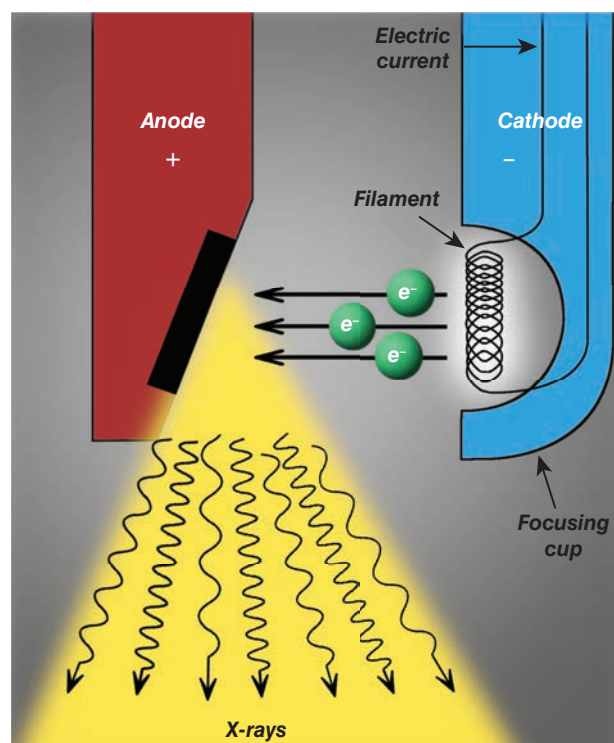


Figure 1.4 X-ray production. This illustration depicts the filament in the cathode being heated to emit a cloud of electrons. The electrons (e^-) are aimed at the anode by the focusing cup and accelerated by a high voltage. The electrons strike a target on the anode (black area) and their kinetic energy is converted into heat and x-rays. The x-rays produced vary in wavelength and energy.

The number of electrons produced is controlled by the strength and duration of the current.

The electrons collect in a “cloud” around the filament and are held in place by the *focusing cup*. The focusing cup concentrates the electrons into a well-defined beam and aims them at the anode. When high voltage is applied across the x-ray tube, a large negative-to-positive gradient is created that rapidly accelerates the electrons from the cathode to the anode. The electrons are traveling at over half the speed of light when they strike the anode.

The anode stops the electrons, converting their kinetic energy into heat and x-rays (about 99% heat and 1% x-rays). Heat is an unwanted by-product of x-ray production and must be dissipated. Methods of heat dissipation will be discussed shortly.

Electron bombardment of the anode produces x-rays with energies that vary from near zero to the maximum energy of the electrons. The variations in energy are due to the way the electrons interact with the anode atoms. There are two types of interactions: *characteristic* and *bremsstrahlung*.

Characteristic x-rays are formed when a high-speed electron from the cathode collides with an atom in the anode target (Figure 1.5). An inner level electron is

ejected from the atom, which produces an ion. Ions are unstable, so an outer level electron drops down to fill the void in the inner level. Because the binding energy of an outer electron is greater than that of an inner electron, the outer electron releases some energy in the form of an x-ray. The energy of the x-ray is equal to the difference between the outer level binding energy and the inner level binding energy and therefore is *characteristic* of the type of atom that was ionized. About 20% of medical x-ray production is due to characteristic interactions (Figure 1.7).

Bremsstrahlung x-rays are produced when a high-speed electron from the cathode passes near the nucleus in an anode atom. The positive charge from the nucleus pulls on the electron, slowing it down and altering its direction (Figure 1.6). “Bremsstrahlung” is a German word that means “braking radiation.” As it slows down, the electron loses energy which is emitted as an x-ray. The energy of the x-ray depends on how much the electron was slowed down. X-ray energies vary from near zero (the electron barely slowed down) to the total energy of the electron (the electron was completely stopped). About 80% of medical x-ray production is due to Bremsstrahlung interactions (Figure 1.7).

X-ray machine

Basic components of an x-ray machine include the x-ray tube assembly, the collimator, the electrical circuits, the operating console, the table, and the image receptor (Figure 1.8).

X-ray tube assembly

The x-ray tube assembly comprises the x-ray tube and its metal housing (Figure 1.9). The metal housing protects and supports the tube and prevents unwanted x-rays from escaping. It also encases the oil that surrounds the x-ray tube to help dissipate heat and provide electrical insulation. Most x-ray assemblies are mounted so they can be moved horizontally, vertically, and rotated to facilitate making radiographs.

The purpose of an x-ray tube is to convert electricity into x-rays. As described earlier, the **cathode** is the negative terminal in the x-ray tube. It contains the filament and focusing cup. The filament is a coil of thin wire, usually made of tungsten. Tungsten is a very malleable metal with a high melting point (over 3400 °C) and a low rate of evaporation, making it ideal for thermionic emission. The focusing cup generally is made of molybdenum. The point on the anode that is struck by the beam of electrons is called the **focal spot**. Most x-ray machines include two focal spots, a large one and a small one, with a separate filament for each.

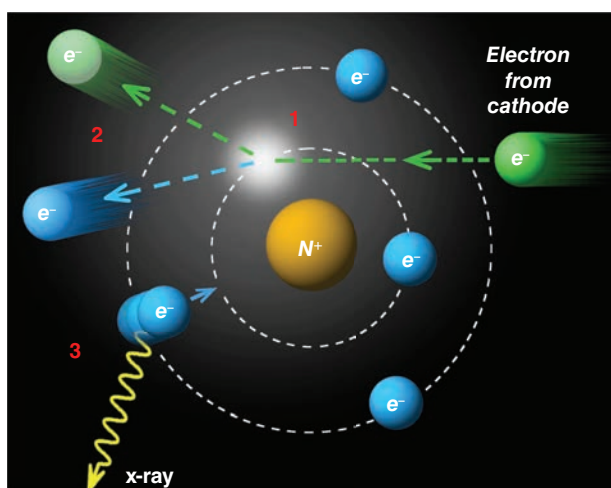


Figure 1.5 Characteristic x-ray production. 1. A high-speed electron from the cathode (green e^-) collides with an inner shell electron in an anode atom. 2. The anode electron is ejected, and the cathode electron continues in a new direction with less energy. 3. An outer shell electron gives up energy as an x-ray and moves to fill the vacancy in the inner shell (N^+ = nucleus, e^- =electron).

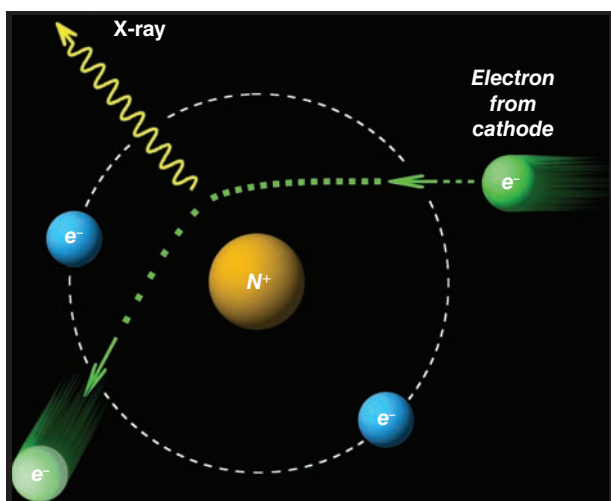


Figure 1.6 Bremsstrahlung x-ray production. A high-speed electron from the cathode (green e^-) nears the positively charged nucleus (N^+) in an anode atom. The electron slows down and changes direction. As it slows, the electron loses energy which is emitted as an x-ray. The energy of the x-ray is determined by how much the electron was slowed down.

The **anode** is the positive terminal in the x-ray tube. This is where x-rays are produced. A typical anode consists of a round, flat, metal disc made of molybdenum with a thin rim of tungsten. The high atomic number of tungsten (74) increases the likelihood of electron interactions. The anode rim is the target for the electron beam. It is beveled to create a slope, which is valuable for x-ray beam geometry, as we will discuss later.

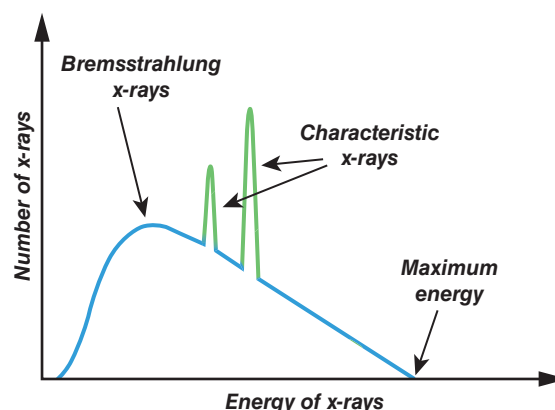


Figure 1.7 Spectrum of x-ray energies. Most of the x-rays used for diagnostic imaging will range in energy from near zero to the maximum energy of the electrons used to produce them. This continuum of energies is due to Bremsstrahlung interactions and is shown in blue in the graph above. A small amount of medical x-rays are characteristic of the material used to produce them, depicted by the green spikes on the graph.

Heat is a major limiting factor in the production of x-rays. To produce an adequate number of electrons, the tungsten filament must be heated to over 2200 °C. To accelerate the electrons, kilovolts of power must be applied across the x-ray tube. To generate x-rays, the target is heated to over 2500 °C during a single exposure. Excessive heating is a primary cause of x-ray tube failure, and the x-ray tube is one of the most expensive parts of an x-ray machine. The most vulnerable part of the tube to heat overload is the focal spot. The larger the focal spot, the greater the heat tolerance. A large focal spot is desirable because more x-rays can be produced; however, the larger the focal spot, the less definition in the radiographs.

Heat is conducted away from the focal spot by the metal in the anode disc, the oil surrounding the x-ray tube, and the protective metal housing. To further dissipate heat, the anode is mounted on a high-speed rotor which spins up to 10,000 rpm during x-ray production. Spinning the anode effectively increases the area of the target that is bombarded by the electron beam and lessens the amount of heating in any one spot.

To avoid overheating, **tube rating charts** are provided by manufacturers. These charts define the maximum *heat units* a particular x-ray tube can tolerate over time before it fails. Heat units (HU) are calculated based on the amount of power that is sent to the x-ray tube ($HU = kV \times mA$). You should never exceed 80% of the maximum heat units with any exposure. **Anode cooling charts** display the minimum time required for the anode to cool down before making another exposure.

X-ray machines that have been idle for more than 6 hours are considered “cold.” A cold x-ray tube should be

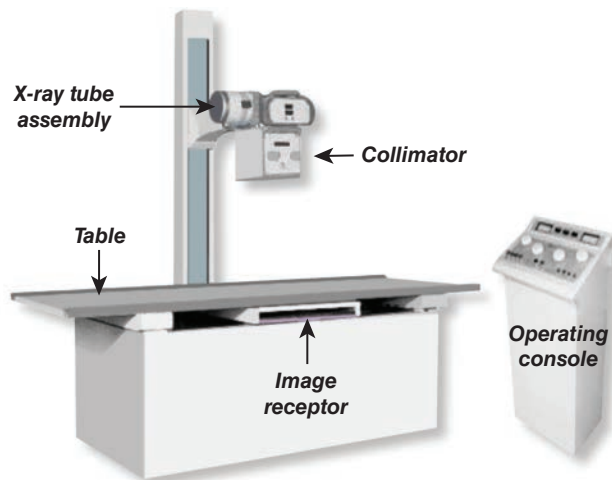


Figure 1.8 *Typical x-ray machine.* The x-ray tube assembly is mounted above the table, preferably on a slide for horizontal movement and on an adjustable column for vertical movement. The collimator is mounted to the x-ray tube. The image receptor may be positioned atop or underneath the table top. In this picture, the image receptor is secured in a tray under the table, which is aligned with the collimator. The operating console is pictured on the right.

warmed up prior to making a high exposure to avoid damaging the anode. To warm up a cold tube, close the collimator and set a low exposure technique (e.g., 60 kVp, 100 mA, 0.05 seconds). Make two exposures about 30–60 seconds apart. The tube is now warmed up and ready for use (Box 1.2).

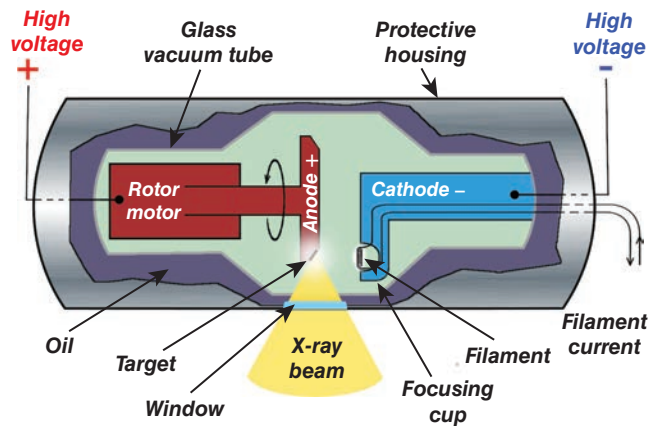


Figure 1.9 *X-ray tube assembly.* This illustration depicts a cutaway view of an x-ray tube assembly. The glass vacuum x-ray tube is surrounded by oil and encased in a protective metal housing. The anode (+) is mounted on a shaft that is attached to an induction motor. The motor spins the anode to help dissipate heat during x-ray production. The cathode (–) is supplied by a low voltage, high amperage current to heat the filament and produce electrons. The filament is mounted in the focusing cup. The x-ray tube is supplied by a high voltage, low amperage current to accelerate the electrons toward the anode target. X-rays emitted from the target travel in all directions, but the protective housing prevents them from exiting the assembly other than through a small window.

Box 1.2 Prolong the life of your x-ray tube

- Adhere to the tube rating and cooling charts.
- Do not exceed 80% of maximum heat units for any exposure.
- Warm up a cold x-ray tube before making a high exposure.
- Use the lowest mA that is practical.
- Avoid prolonged rotation of the anode.

Collimator

A collimator is an arrangement of x-ray absorbers used to limit the size and shape of the x-ray beam to a specific **field-of-view (FOV)**. The FOV is the area to be irradiated. It is the part of the patient that is exposed to the primary x-ray beam.

Collimators are mounted to the x-ray tube assembly in the path of the x-rays (Figure 1.10). Modern collimators contain adjustable shutters made of lead which can be manipulated to create the desired FOV. The smaller the FOV, the better the definition in the radiograph.

Inside many collimators are a light bulb and an array of mirrors which are used to project a visible light FOV that corresponds to the x-ray beam FOV. This allows the radiographer to accurately configure the area to be irradiated.

Filtration

The purpose of filtration is to absorb low-energy x-rays and remove them from the x-ray beam. Low energy x-rays are

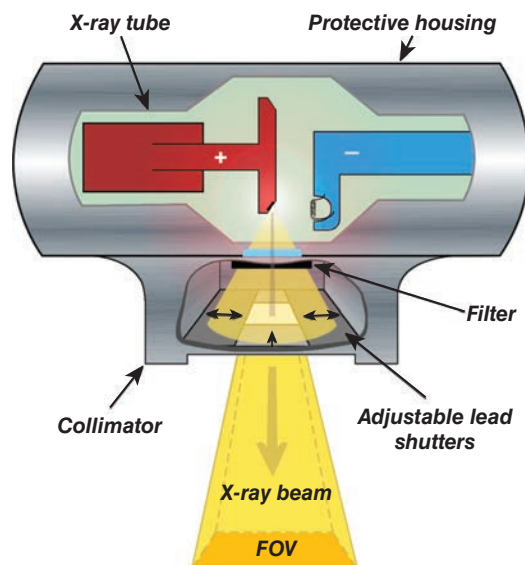


Figure 1.10 *Collimator.* In this illustration, the collimator is mounted to the x-ray tube assembly in the path of the x-rays. The side of the collimator is cut away to show the adjustable lead shutters inside. The shutters can be moved to change the size and shape of the FOV. The side of the protective housing is cut away to show the x-ray tube. An added filter is visible between the collimator and x-ray tube assembly, in the path of the x-rays.

unable to pass through the patient and therefore provide no useful diagnostic information. They do, however, increase the radiation dose to the patient. All x-ray beams contain low-energy x-rays due to Bremsstrahlung interactions. The National Council on Radiation Protection and Measurements (NCRPM) requires that all x-ray tubes operating above 70 kVp must include filtration that is equivalent to at least **2.5 mm aluminum**. This applies to most veterinary practices. There is some inherent filtration in most x-ray machines, but additional filters usually are needed.

Inherent filtration comes from the x-ray tube glass envelop, the surrounding oil, the glass window in the protective housing, and the collimator. Inherent filtration typically contributes about 1.5 mm aluminum equivalent. **Added filtration** generally consists of aluminum plates that are positioned near the x-ray tube in the path of the x-rays (Figure 1.10). The plates provide the additional 1–2 mm of aluminum required.

In addition to lowering the radiation dose to the patient, filtration also increases the average or **effective energy** of the x-ray beam. Removing the low energy x-rays from the beam is called **beam hardening**. Beam hardening increases the effective energy of the x-rays from about 1/3 of kVp to about 1/2 of kVp. A higher effective energy means the x-ray beam is more uniform in intensity, which can improve radiographic quality.

Selective filtration sometimes is used to make the x-ray beam more intense on one side than the other. This can be helpful when the thickness of the body part varies significantly. Rather than making two different exposures, one for the thick side and another for the thin side, a *compensation filter* can be used (Figure 1.11). A typical compensation filter is a wedge-shaped piece of aluminum that unilaterally absorb x-rays. It makes the x-ray beam less intense on the thin side to prevent overexposing that side of the body part while allowing full exposure on the thicker side. The resulting radiograph is more uniform in density across both thicknesses. The **heel effect** provides a similar benefit and is discussed on page 16.

Compensation filters are attached to the front of the collimator, usually with magnets or Velcro. Some collimators are equipped with an adjustable holder designed to fit a compensation filter. The filter is installed after setting the FOV because the light from the collimator will be blocked by the filter. Compensation filters are less important with digital radiography because computer processing can be used to manipulate the image and adjust for variations in density.

Electrical supply to x-ray machine

The x-ray machine must receive a constant level of power to produce x-rays in a consistent and reliable manner. Fluctuations in power can cause significant variations in x-ray production. The line of incoming electricity should be

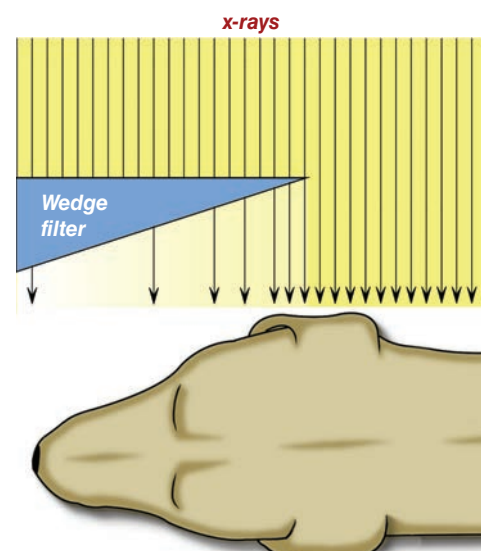


Figure 1.11 *Compensation filter.* In this illustration, the dog's head varies in thickness from the nose to the ears. A wedge filter is used to reduce the amount of x-rays to the nose while allowing the full intensity of the x-ray beam to reach the thicker skull. This allows the head to be imaged without overexposing the nose or underexposing the skull. The gradual slope of the wedge blends the different x-ray beam intensities together, so there is no sharp demarcation between the lighter and darker areas in the radiograph.

dedicated to the x-ray machine with no other equipment on the line. Appliances such as a clothes dryer or an air conditioner will pull power away from the x-ray machine. A **line voltage compensator** is a useful device that adjusts for variations in incoming voltage. A voltage compensator is recommended because some power fluctuations are beyond the control of the hospital or clinic (e.g., originate at the power company).

The source of electricity for most x-ray machines is **alternating current** (AC). Alternating current regularly reverses direction, switching the flow of electricity from positive to negative many times per second (Figure 1.12). Depending on the country of origin, AC electricity is either 50 or 60 Hz, which means it cycles from one direction to the other 50 or 60 times per second. **Direct current** (DC) moves in only one direction and does not reverse. X-ray machines use both AC and DC power. However, the high voltage generator in the x-ray machine converts AC to DC, so a separate DC supply is not needed. There are two different electrical circuits in an x-ray machine, the *filament circuit* to produce electrons and the *high voltage generator* to accelerate the electrons.

The **filament circuit** increases the amperage of the incoming AC power to provide enough current to heat the tungsten wire and produce a sufficient number of electrons. The amperage is increased using a *step-down transformer*. A transformer is a device that increases or decreases voltage.

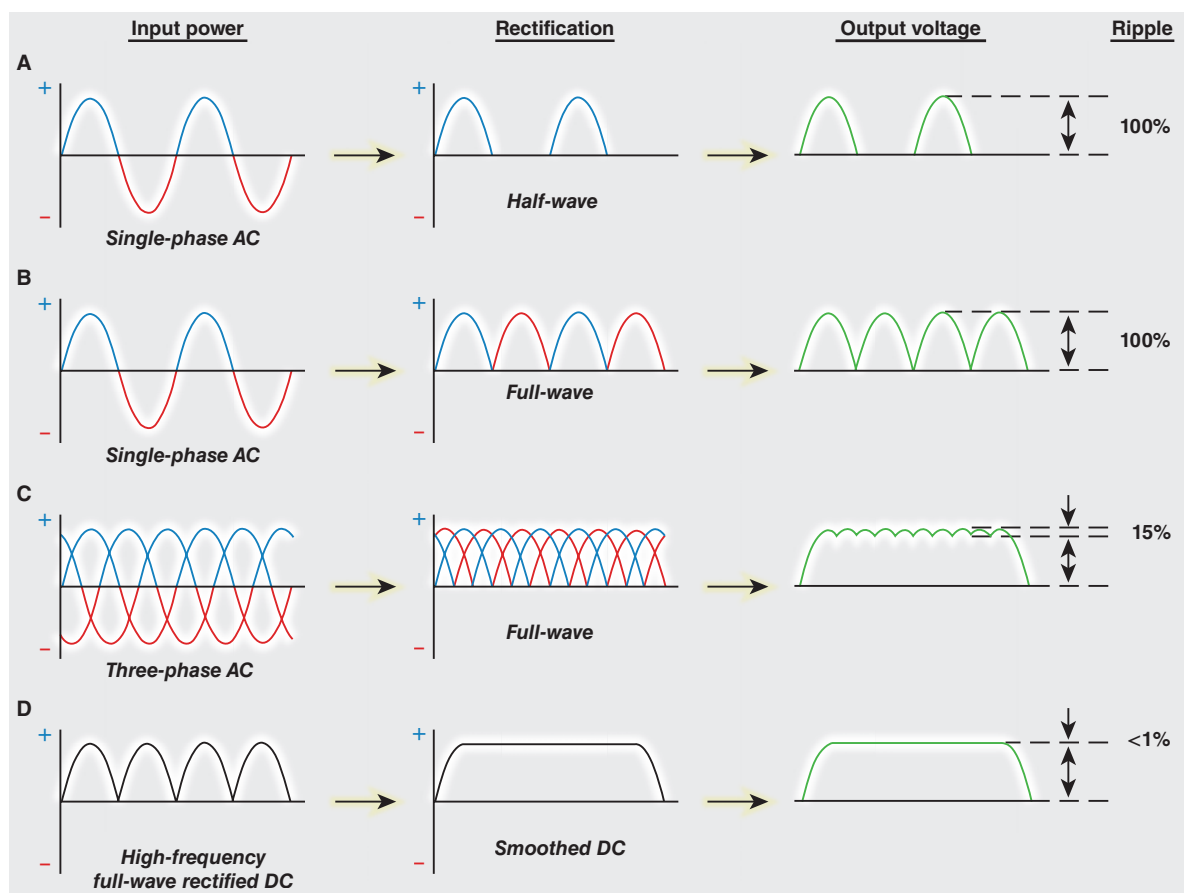


Figure 1.12 High voltage generators. This illustration depicts half-wave and full-wave rectification with single phase, three phase, and high frequency generators. The input power for single-phase generators (**A** and **B**) is a single line of alternating current (AC). AC power oscillates the flow of electricity from a positive direction (+) to a negative direction (–) many times per second. X-ray tubes require a current that moves in one direction only, which means the AC must be rectified to direct current (DC). Half-wave rectifiers (**A**) remove the negative direction of the AC to produce pulses of DC electricity. During each pulse the voltage fluctuates from zero to the maximum and back to zero again, resulting in 100% ripple. Full-wave rectifiers change the negative direction of AC to the positive direction to produce twice the number of DC pulses as half-wave rectifiers. However, the voltage still fluctuates from zero to maximum with each pulse and the ripple is still 100%. Three-phase generators (**C**) utilize three separate lines of input AC power. Each line is slightly out of phase with the other two. All three lines are full-wave rectified to produce overlapping pulses of DC and a more continuous output of voltage than single-phase units. As the voltage begins to drop in one pulse, the next pulse brings it back up to the maximum, resulting in only 15% ripple. High-frequency generators (**D**) utilize either single-phase or three-phase AC input that is full-wave rectified to DC. However, high-frequency generators then “chop” the DC pulses into smaller and smaller pulses and “smooth” them to produce near-constant output voltage with less than 1% ripple.

Voltage and amperage together constitute power, and the power leaving a transformer must be the same as the power coming into it. The equation is $W = V \times A$, where W is power measured in watts, V is voltage, and A is amperage. Any change in voltage must include an opposite change in amperage so the power can remain constant. The step-down transformer, therefore, increases amperage to the filament by decreasing the incoming voltage.

The **high voltage generator** increases the voltage of the incoming AC power to provide the *kilovolts* needed to accelerate the electrons with enough speed to generate x-rays. This is accomplished with a *step-up* transformer. The step-up transformer increases the incoming voltage 500–1000 times,

which drops the amperage to only a few milliamperes. This means the tube current (current across the x-ray tube) is very low amperage and very high voltage.

The high voltage generator also converts AC to DC. X-ray tubes are designed to use current that flows in one direction only: from the cathode to the anode. Current that flows in the other direction will not produce x-rays and may damage or destroy the x-ray tube. The process of converting AC to DC is called **rectification**. Rectification may be either *half-wave* or *full-wave*, both of which produce a DC output that consists of “pulses” of electricity (Figure 1.12).

Half-wave rectifiers suppress the negative direction of AC, so only the electricity that flows in the positive direction

is available. For example, if the input electricity is 60 Hz AC, the half-wave rectifier would produce a DC output to the x-ray tube of 60 pulses/second of electricity (Figure 1.12). As you can see, half-wave rectifiers only use half of the input AC, the other half is wasted. **Full-wave rectifiers**, on the other hand, change the negative direction of AC to the positive direction. Using the same example, a 60 Hz AC input would be converted to a DC output with 120 pulses/s, which is twice the electrical output of a half-wave rectifier.

A sidenote that may be interesting: because x-ray tubes only operate when current flows in one direction, a simple x-ray tube that receives only AC power actually is *self-rectified*. The problem with a self-rectified tube is that the anode is likely to get hot enough to emit its own free electrons and then the current would flow in the reverse direction and destroy the cathode.

Because rectification produces DC output in pulses, the amount of electricity going to the x-ray tube can fluctuate greatly. This fluctuation in power is called **ripple** (Figure 1.12). Specifically, ripple is the percent variation in voltage, from minimum to maximum, that is received by the x-ray tube. The smaller the ripple, the more constant the voltage and the more uniform and intense the x-ray beam. The amount of ripple depends on the type of high voltage generator. Currently, there are three basic types of generators: *single-phase*, *three-phase*, and *high-frequency*.

Single-phase generators receive a single source of AC power. The AC is rectified to produce a DC output with 100% ripple (Figure 1.12). 100% ripple means the x-rays are produced in bursts instead of continuously. The intermittent production of x-rays leads to longer exposure times, higher radiation doses, and a lower x-ray beam intensity (only about 70% of peak output). The shortest exposure time with a single-phase generator is 1/120 second (~8 ms).

Three-phase generators receive three separate lines of AC power. Each line is superimposed on the other and 120° out of phase with the other two (Figure 1.12). All lines are full-wave rectified so each line of 60 Hz AC produces 120 pulses of DC. The three lines are combined so together they produce 360 pulses/s of DC power. Because the lines are out of phase with each other, as the voltage in one pulse begins to drop another pulse brings it back up to the maximum so the voltage never falls to zero. The x-ray tube receives more constant power with a ripple that is only about 15%. Some three-phase generators include extra rectification to produce DC with 720 pulses/s and only 4% ripple. These are called three-phase, 12-pulse generators. Exposure times with three-phase generators can be as short as 1/1000 second (1 ms), and the x-ray beam intensity is about 95% of peak output.

High-frequency generators supply near-constant voltage to the x-ray tube. The input power may be single-phase or three-phase AC which is then full-wave rectified (Figure 1.12). High-frequency generators then use high-speed switches called *inverters* to effectively “chop” the

DC pulses into smaller and smaller pulses. Inverters can produce DC with up to 25,000 pulses/s. All of these pulses are then “smoothed” using additional rectifiers and capacitors so the x-ray tube receives voltage with less than 1% ripple. High-frequency generators enable shorter exposure times, lower radiation doses, and help extend the life of the x-ray tube. The x-ray beam intensity may be over 99% of peak output. In addition, many high-frequency generators are more compact, about 1/10 the size of a three-phase generator.

Operating console

The x-ray machine operating console includes controls to turn the equipment on and off and to adjust the mA, kVp, and the length of time the x-rays are being produced.

The **mA** (*milliamperes*) control regulates the quantity of x-rays produced. It determines the amount of current that is sent to the filament and therefore the number of electrons that are emitted. The higher the mA, the more electrons and the more x-rays. The mA control has no effect on the energy or penetrating power of the x-rays. Most x-ray machines provide fixed settings for mA (i.e., 100 mA, 200 mA, 300 mA).

The **kVp** (*kilovolt peak*) control primarily regulates the energy or quality of the x-rays. It regulates the amount of voltage that is sent to the x-ray tube which determines the speed of the electrons. The higher the kVp, the greater the kinetic energy of the electrons and the greater the energy of the x-rays.

The kVp control also contributes to the number of x-rays produced. Higher kVp means electrons are more quickly removed from the cloud, which allows more electrons to be emitted from the filament. More electrons leads to more x-rays.

The kVp control on most operating consoles displays continuously variable units of *kilovoltage* (kV). Voltage can be thought of as the “force” that moves the electrons from the cathode to the anode. Thousands of volts are required to accelerate the electrons with enough energy to create useful x-rays. The maximum speed of the electrons is determined by the maximum or **peak kilovoltage**, which is called **kVp**. It is important to realize that kVp determines the peak energy of the x-rays and not the total energy or the individual energy of the x-rays. This is because not all of the electrons will be moving at maximum speed (due to the sine wave distribution of the voltage, as shown in Figure 1.12). The slower electrons produce less energetic the x-rays. In addition, remember that many of the x-rays produced at the anode will be less than maximum energy due to Bremsstrahlung. Therefore, the average or effective energy of most x-ray beams is only about 1/3 of kVp. The effective energy can be increased to about 1/2 of kVp with high-efficiency generators and beam hardening (filtration).

The **exposure switch** is a manual control that is used to start the production of x-rays. The **timer** is an automatic control that stops x-ray production. The exposure switch

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typically operates in two stages: a *prep stage* and an *exposure stage*. The **prep stage** prepares the x-ray tube by rotating the anode and initiating the heating of the filament. It is important to make sure the anode is rotating at full speed before making an exposure, but it is equally important to avoid prolonged rotation as this will shorten the life of the x-ray tube. During the **exposure stage**, x-rays are being produced. The exposure stage is terminated automatically by the timer. There are a variety of timer mechanisms and most include a meter or digital display with units in fractions of a second (S) or milliseconds. In each case, the radiographer begins the exposure and the timer automatically stops it.

X-ray table

The x-ray machine table supports the patient during radiography. X-ray tables are made of strong materials with low x-ray attenuation, such as carbon fibers. Tables should be uniform in thickness and opacity and able to support at least 300lb (136 kg). A slightly elevated rim or edge around the table top helps prevent liquids from spilling onto the floor or into the equipment under the table. Many x-ray tables include a built-in tray designed to hold a grid and various types and sizes of cassettes. The tray should be aligned with the x-ray tube so the two can be moved simultaneously.

Floating tables help facilitate patient positioning. Most floating table tops can easily be locked, unlocked, and moved in multiple horizontal directions. The tray under a floating table generally remains aligned with the x-ray tube while the tabletop is being moved. Tables that are height adjustable also facilitate patient positioning, especially given the variety of patient sizes commonly encountered in a veterinary practice.

Image receptors

Image receptors record the x-rays that pass through an object or patient. The x-rays received are used to form a **latent** image. The latent image is invisible and must be converted to a visible image for interpretation. Methods of conversion will be discussed shortly.

An image receptor must record all of the different x-ray intensities that emerge from a patient and accurately translate them into acceptable levels of radiographic density (visible shades of gray). The minimum number of x-rays a receptor can receive and still adequately darken the radiograph (produce a visible shade of gray) is known as its **sensitivity**. The more sensitive the system, the fewer x-rays are needed. An image receptor must also display each level of density (shade of gray) with enough contrast so one can be distinguished from the next. This is called the **dynamic range** or **latitude** of the receptor. Wide latitude receptors are able to form many shades of gray over a large range of x-ray intensities, which means they can display many different levels of opacity. Narrow latitude receptors form fewer

shades of gray over a smaller range of x-ray intensities. Narrow latitude means fewer levels of opacity can be displayed. The amount of radiographic **detail** an image receptor can provide determines how well we can discriminate the edges of structures that are close together.

Currently, there are two basic types of image receptors: *film:screen* and *digital*. Both types are used with conventional x-ray machines.

Film:screen image receptors

The image receptors in traditional film radiography utilize special photographic film coupled with phosphorescent sheets called **screens**. The screens convert x-rays into visible light. X-ray film is much more sensitive to light than to x-rays so the screens act to *intensify* the effect of the x-rays on the film. This means fewer x-rays are needed to adequately expose the film when using an intensifying screen. In a typical film radiograph, over 95% of the film exposure is due to light from an intensifying screen and less than 5% is from the x-rays themselves. Intensifying screens enable lower x-ray exposures, less radiation dose, and longer life of the x-ray tube.

An **intensifying screen** consists of a thin polyester base coated on one side with phosphors. The phosphor layer is covered with a thin protective coating. The amount of light emitted by the phosphors is directly proportional to the amount of x-rays received by the screen. Most phosphors are either a *rare earth* material that emits a green light or *calcium tungstate* which emits blue light. Modern x-ray films are orthochromatic, which means the film is more sensitive to a specific color of light. It is important to match the phosphor emission color with the film absorption color.

X-ray film consists of a thin, flexible, polyester sheet coated on one or both sides with an emulsion that contains silver halide crystals. The emulsion is covered with a thin protective layer. The plastic base is transparent and often tinted blue to reduce eye strain.

Silver halide crystals become structurally altered when struck by light from an intensifying screen or from direct interaction with an x-ray. The altered crystals form the latent image. The latent image is then *processed* to become visible. Processing converts the altered silver halide crystals to black metallic silver. Film processing is discussed further below.

Both the intensifying screens and the x-ray film commonly are enclosed in a protective, light-proof container called a **cassette** (Figures 1.13 and 1.15). A cassette may contain one screen for use with single emulsion x-ray film or two screens for double emulsion film. It is important to load the cassette with the correct type of film. Cassettes are available in a variety of sizes that correspond to different film sizes. Most cassettes swing open on a hinge, making it easy to remove and load the film. Screens usually are permanently mounted in the cassette, often on a flexible foam pad to maintain close, uniform contact with the film. The back of

the cassette frequently is made of a radiation absorbing material to block backscatter radiation.

Cassettes can last for a long time with proper care. They should be regularly inspected for damaged hinges, loose screens, light leaks, etc. (Box 1.3). Do not store cassettes in prolonged heat as this will damage the intensifying screens. Numbering each cassette is useful to help identify problems that may arise. Use a permanent marker to write the number on an inconspicuous part of a screen, such as near the ID label or in a corner (Figure 1.14). Write the same number on the outside of the cassette. If an artifact appears in a radiograph, you'll know which cassette to inspect for a possible cause.

Film processing

The process of converting the invisible latent image into a visible radiograph involves immersing the film in different chemicals for specific periods of time. Film processing may be performed manually or automatically in a machine. With manual film processing, the chemical solutions are stored in dip tanks and the technician transfers the film from one tank to another, paying special attention to the time intervals (Box 1.4). With automatic processors, the film quickly moves from one solution to the next without human intervention. Automatic processors operate at higher temperatures to speed up the chemical reactions and shorten the processing time. The chemicals used in automatic processors

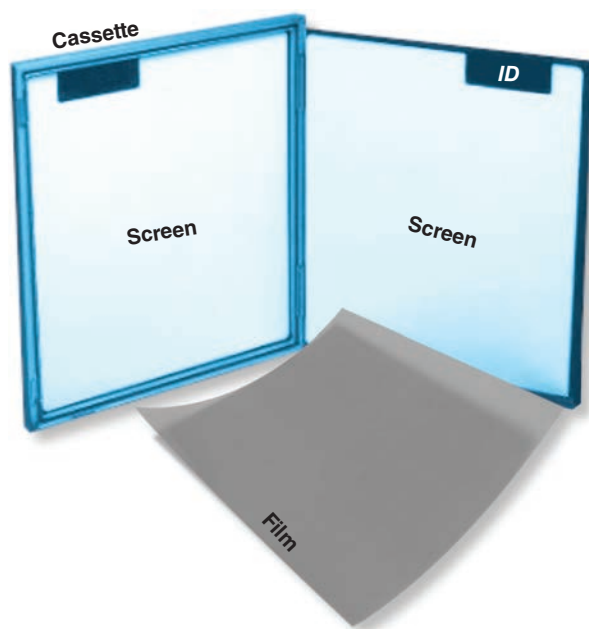


Figure 1.13 Film:screen image receptor. In this picture, the cassette is open and standing on end with a sheet of x-ray film in front. Intensifying screens are mounted inside the cassette, one on each side. The rectangular cutouts along the upper edge of each screen are for labeling the radiograph for identification (ID).

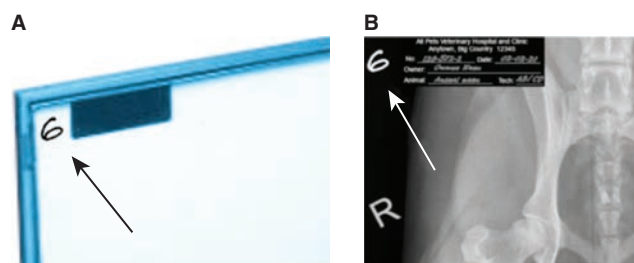


Figure 1.14 Numbered cassette. **A.** The number "6" is written on an intensifying screen (arrow) in the corner of an open cassette. **B.** In a radiograph made using this cassette, the number "6" is visible in the corner (arrow). Numbering the cassettes helps identify those with problems.

Box 1.3 Caring for intensifying screens

1. Screens should be cleaned at least monthly with a gentle, antistatic cleaner and a soft, lint-free cloth. Stand the cassette on end while the screens are drying so dust doesn't collect on them.
2. Handle the screens carefully. They are coated with a thin protective layer but can easily be scratched. Scratches in screens are permanent.
3. Periodically inspect each screen for dirt, stains, scratches, etc., as these can produce radiographic artifacts. Use a UV light or a "black light" in a darkened room to see the damaged areas.

differ from those used for manual processing in that they are specifically designed to work at high temperatures. Whether manual or automatic, film processing involves five basic steps:

1. Develop.
2. Rinse.
3. Fix.
4. Wash.
5. Dry.

The **developer** is a chemical that converts the silver halide crystals to black metallic silver. The altered crystals are converted more quickly than the unaltered crystals, but given enough time the developer will eventually change all of the crystals to black. Therefore, the time the film spends in contact with the developer must be carefully controlled. Development is stopped by **rinsing** the film with water to dilute and remove the chemicals. Automatic processors incorporate rollers to squeeze the rinse water and chemicals away from the film.

The **fixer** is a chemical that dissolves any undeveloped silver halide crystals so they can be washed from the film. Fixer also "hardens" the film emulsion to help preserve the image. After an appropriate length of time in the fixer, the film is **washed** with water to remove the dissolved crystals

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and any chemicals. Automatic processors again use rollers to squeeze away the wash water and chemicals. The film is then **dried** for viewing and storage. Film radiographs should always receive a final viewing after they are dry, as the appearance of the radiograph may differ between wet and dry.

Film processing can significantly affect the quality of the radiograph. It is critical to maintain the developer and fixer at proper concentrations and temperatures per the manufacturer's recommendations. Pay special attention to film immersion times and sequences. Be careful to avoid rough handling of film and splashing of chemicals. All processing equipment should be periodically inspected to ensure peak performance and proper chemical replenishment rates.

Disposal of processing chemicals must be in accordance with environmental protection rules and regulations. Companies and special equipment are available for reclaiming silver from fixer and washer solutions and from x-ray film that is no longer needed. Reclaimed silver can be both environmentally and financially rewarding.

Film fog

Silver halide crystals can be altered by conditions other than light from intensifying screens or x-ray interactions. These include exposure to ambient light, heat, pressure, chemicals, and aging. Crystals that are altered by conditions other than those caused by x-rays add density to the radiograph, but do not provide any useful information. Unwanted blackening of x-ray film is called *fog*. Fog makes it more difficult to identify structures in the image (Figure 1.16 and 1.36). Many of the causes of x-ray film fog can be avoided if we know about them. See the Artifacts section in Chapter 2 for figures depicting film fog.

Light fog is caused by exposing unprocessed x-ray film to visible light. Light fog can be avoided by storing unused

film in light-proof containers. Also, make sure the dark-room is light-proof, and the safelight is the correct color and intensity (see safelight test in the Artifacts section of Chapter 2). Static electricity is a common cause of light fog and can be minimized by electrically grounding darkroom personnel, controlling the humidity, and carefully handling the x-ray film.

Heat fog is caused by prolonged exposure of unprocessed x-ray film to temperatures above 70°F (20°C). Avoid heat fog by storing unused film in a controlled environment.

Pressure fog is caused by physically compressing unprocessed x-ray film. Pressure fog can result from storing boxes of unused film flat instead of upright. Over time the weight of the top sheets will fog the bottom sheets. Pressure fog also

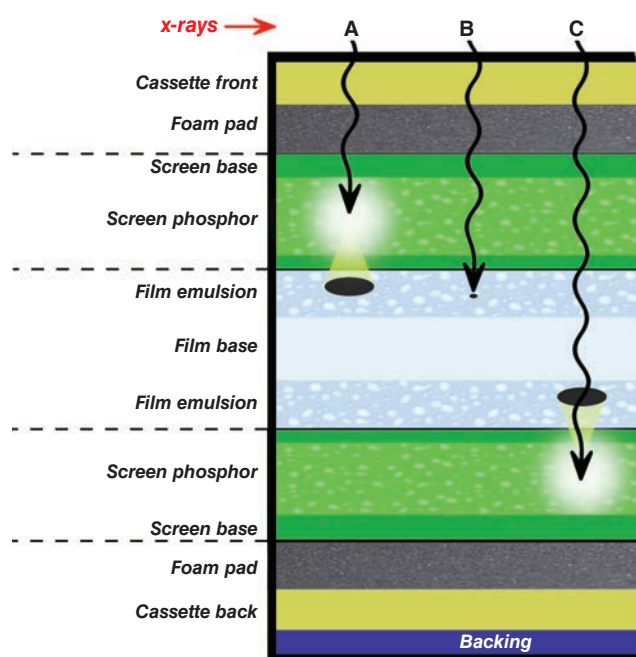


Figure 1.15 *Film-screen system*. This illustration depicts a cross-sectional view of a loaded film cassette. A sheet of x-ray film (colored light blue) is sandwiched between two intensifying screens (colored green). The film is coated with emulsion on both sides. Three x-rays, labeled A, B, and C, are depicted interacting with the film-screen system. **X-ray A** strikes a phosphor in the top screen causing it to emit light in all directions. The light that hits the film alters some silver halide crystals in the upper film emulsion, which will produce a black area when the film is developed (depicted by the black oval). **X-ray B** passes through the top screen without interacting with any phosphors and strikes the film directly. It alters fewer silver halide crystals than light from the screen (thus, a smaller black oval will be produced when the film is developed). **X-ray C** passes through all of the upper layers to strike a phosphor in the bottom screen. The light emitted from the screen travels in all directions and alters some of the silver halide crystals in the bottom film emulsion. This is the advantage of double emulsion x-ray film: it increases the likelihood of x-ray detection so fewer x-rays are needed to produce acceptable radiographic density. The backing in the cassette is designed to block backscatter x-rays from exposing the film.

Box 1.4 TIPS for manual film processing

- Keep the developer and fixer tanks covered to reduce evaporation and chemical breakdown.
- An aquarium heater in the water bath is helpful to maintain proper temperature.
- Thoroughly mix both the developer and fixer solutions prior to each use.
- Agitate the film for a second or two while it is immersed to ensure good contact with the chemicals and to dislodge any air bubbles.
- Do not allow the solution clinging to the film to drain back into the tank for more than 2 seconds; used solution will weaken the remaining chemicals.
- Make sure the darkroom has an exhaust fan that is in use whenever the room is occupied; this is an OSHA requirement in the United States.



Figure 1.16 Fog. Two photographs of a lakeshore depicting the appearance of structures with fog (A) and without fog (B).

is caused by bending or creasing the film (Figure A23) or by sliding it across a table or countertop.

Chemical fog is caused by exposing unprocessed x-ray film to the vapors from cleaning solutions, ammonia, formaldehyde, spare developer, and others. Avoid chemical fog by storing unused film away from these and similar agents and making sure your fingers are clean and dry before handling the film.

Aging fog occurs with expired x-ray film. Over time, the unused film will “self-process” and become darker. Aging fog can be avoided by paying attention to expiration dates and not using expired film for radiography.

Characteristics of film:screen systems

The thickness of the emulsions and the sizes of the *grains* in a film:screen image receptor determine its sensitivity and the amount of radiographic detail it can provide. **Grains** are the silver halide crystals in the film emulsion and the phosphors in the intensifying screens. Both sensitivity and detail are directly related to the sizes of the grains, but they are inversely related to each other. The larger the grains, the more sensitive the system, but the less detail it will provide. The smaller the grains, the better the detail, but more x-rays will be needed to adequately darken the image.

Sensitivity in film radiography sometimes is called **speed**. Speed is a term from the early days of radiography when the x-ray machine controls for mA and kVp were limited and setting the exposure time was the most variable factor. Film:screen systems that could accommodate short exposure times and still produce acceptable density in a radiograph were called *high speed*. Systems that required longer exposure times were called *slow speed*. The term “speed” has persisted even though modern x-ray machines allow intricate mA and kVp manipulations in addition to exposure times.

The **sensitivity** and **detail** of a film:screen system is built-in. The only way to change it is to switch to a system with

different size grains or a different emulsion thickness. A film:screen system with larger grains can produce acceptable radiographic density with fewer x-rays. When fewer x-rays are needed, shorter exposure times are possible and larger body parts can be imaged. However, radiographic detail will suffer because larger grains are more visible in the film radiograph, resulting in a “mottled” or “grainy” appearance. A film:screen system with smaller grains produces better detail, however, higher x-ray exposures will be needed to adequately darken the radiographs. High detail film:screen systems generally are limited to imaging smaller body parts.

Digital image receptors

Digital image receptors convert x-rays into electronic signals. The signals are then digitized and sent to a computer. The computer transforms the digital image into an analog radiograph so it can be viewed and interpreted by humans. Newer technologies are constantly evolving and digital systems are continuing to improve. The currently available systems may be broadly classified as either *CR (computed radiography)* or *DR (digital radiography)*. DR systems can be further divided into three general categories: (1) those that directly convert x-rays into electronic signals, (2) those that indirectly convert x-rays into electronic signals, and (3) CCD, which stands for *charged coupling device*. Whether CR or DR, all-digital image receptors incorporate four basic steps:

1. Capture the x-rays to form a latent image.
2. Translate the latent image into electronic signals.
3. Convert the signals to digital data for the computer.
4. Process the data to form an analog image.

Computed radiography

A computed radiography (CR) image receptor captures x-rays using an imaging plate with *photostimulable storage phosphors (PSP)*. The imaging plate is enclosed in a cassette that is similar in size, look, and feel to the cassettes used in film radiography (Figure 1.17). CR and film:screen sometimes are called “cassette radiography.”

When exposed to x-rays, PSP temporarily stores captured x-ray energy to form a latent image. After exposure, the CR cassette is manually transferred to a machine that removes the imaging plate and scans it with a laser beam (Figure 1.18). The laser light stimulates the PSP to release the stored x-ray energy as light, a process known as *photostimulation*. The amount of light emitted is proportional to the captured x-ray energy. A light guide collects the emitted light and sends it to a photomultiplier tube (PMT). The PMT converts the light into electronic signals, amplifies the signals, and sends them to an analog-to-digital converter (ADC). The ADC digitizes the signals and sends them to a computer for processing. While it is scanning the imaging plate, the laser removes most of the latent image. Any remaining latent image is erased by exposing the plate to bright light, making the

cassette immediately ready for re-use. With proper care, CR cassettes can be used thousands of times.

CR imaging plates are always “on,” meaning they are always sensitive to nearby radiation. Nearby radiation includes that coming from the environment (background radiation) and scatter from radiographing other patients. CR cassettes should not be stored near the x-ray machine and any cassettes that haven’t been used for several days should be placed in the CR reader and erased prior to use.

CR image receptors have been around since the mid-1980s and closely emulate traditional film:screen systems. They are compatible with most existing x-ray facilities and often utilize the same exposure techniques as for film radiography. CR cassettes are portable, enabling them to be used with multiple radiography facilities and different x-ray machines as well as for positional radiography and both table-top and under-the-table techniques. The initial cost of a CR system often is lower than many DR systems. Compared to DR, however, CR requires more user intervention, which means it takes longer to complete a radiographic study and patient throughput is slower.

Digital radiography

Digital radiography (DR) image receptors capture x-rays using either a *photoconductor* or a *scintillator*. A photoconductor translates x-ray energy directly into electrical charge as a one-step process. A scintillator is a crystalline structure that converts x-ray energy into visible light and then channels that light through thin crystals to a photodiode. The photodiode converts the light into electrical charge. This is a two-step process that indirectly converts x-ray energy into electrical charge. Whether formed directly or indirectly, the electrical charge is proportional to the energy of the x-rays and forms the latent image.

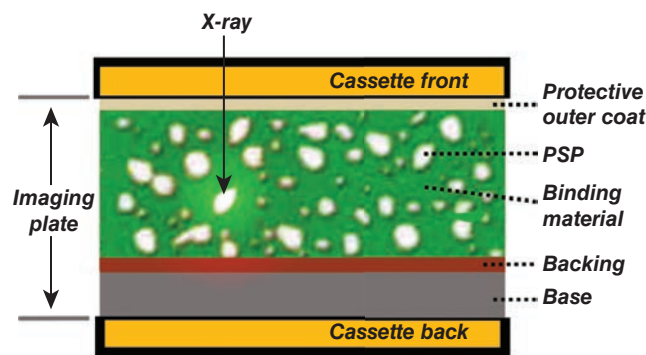


Figure 1.17 CR image receptor. This illustration depicts a cross-sectional view of a CR cassette. The typical CR imaging plate is about 0.3mm thick and composed of a stiff base material coated with a layer of photostimulable phosphors (PSP). The PSP grains are held in a polymer binding material. They absorb and temporarily store x-ray energy. The surface of the imaging plate is protected by a tough plastic coating. The backing material between the base and the PSP layer is designed to block backscatter x-rays.

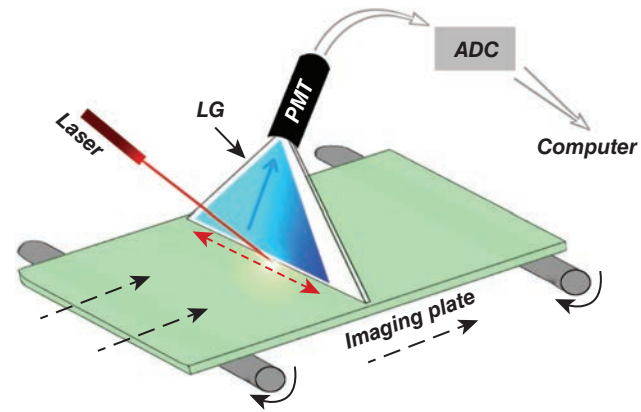


Figure 1.18 CR read-out. This illustration depicts a CR imaging plate inside the CR reader. The plate is transported along rollers as a laser beam sweeps back and forth scanning the latent image. The red laser light stimulates the phosphors to release their stored energy as blue light. The blue light is captured in a light guide (LG) and channeled to a photomultiplier tube (PMT). The PMT converts the light into electronic signals and sends them to an analog-to-digital converter (ADC). The ADC digitizes the signals and sends them to a computer for processing and analog display.

Transistors are used to convert electrical charge into electrical signals. Each transistor contains a detector element that corresponds to a pixel in the digital radiograph. A group of thin-film transistors called a *TFT array* produces the electronic signals, which are then digitized by an ADC and forwarded to a computer (Figure 1.19).

A **charged coupling device** (CCD) is a type of indirect DR image receptor. A CCD uses a scintillator to convert x-rays to light which is collected by an optical system. The optical system may consist of mirrors, lenses, prisms, or fiber optics. It captures the light and sends it to an array of CCD sensors. CCD sensors both convert the light into electrical charge and convert the electrical charge into electronic signals, without the need for photodiodes or a TFT array. CCD sensors are incorporated in many digital cameras.

As opposed to CR systems, DR image receptors convert captured x-rays into electronic signals without the need for user intervention (Figure 1.20). In addition, DR systems are linked directly to a computer which further reduces the time needed to complete a radiography study. Compared to CR and film radiography, DR systems enable faster patient throughput and shorter turnaround times. DR image receptors can be permanently mounted under the x-ray table, or they can remain portable for tabletop use and positional radiography (Figure 1.21).

Characteristics of digital systems

Unlike film:screen systems, digital image receptors do not have a fixed sensitivity. They are able to capture many more of the incident x-rays and successfully convert them into acceptable radiographic density. How accurately and

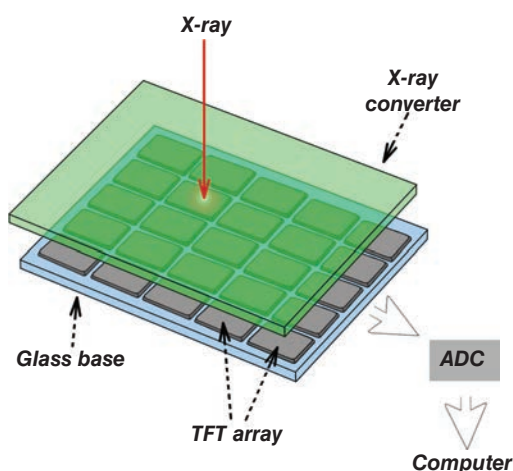


Figure 1.19 DR image receptor. This illustration depicts the key components in a DR image receptor. The x-ray converter either directly or indirectly translates x-ray energy into an electrical charge. It is layered over a thin film transistor array which is mounted on a glass base. The TFT array converts electrical charge to electronic signals and sends them to an analog-to-digital converter, where they are digitized and sent to a computer.

efficiently a digital receptor does this is described by various measurements including the *modulator transfer function* and the *detective quantum efficiency*, both of which are briefly discussed below.

Not all parts of a digital image receptor can convert x-rays into electronic signals. This means that digital sensitivity is affected not only by the number of x-rays that reach the receptor but also where they land. The area of the receptor that is sensitive to x-rays in relation to the total area of the receptor is called the **fill factor**. The larger the fill factor, the fewer x-rays will be needed to produce adequate density.

Modulator transfer function (MTF) measures how accurately a digital receptor transfers the different x-ray intensities into different levels of radiographic density. The higher the MTF, the more different opacities can be displayed. MTF ranges from 0 (no structures in the patient are visible) to 1 (all structures are visible). An MTF of 0.5 means the receptor can display 50% of the differences in opacity as different radiographic densities.

Detective quantum efficiency (DQE) measures how efficiently a digital receptor captures the x-rays it receives and converts them into acceptable radiographic density. The higher the DQE, the fewer x-rays are needed. For example, a DQE of 0.4 means the receptor can utilize 40% of the x-rays it receives. DQE is the ratio of input SNR to output SNR (SNR is signal-to-noise ratio). The higher the DQE, the better the radiographic detail.

The **exposure index (EI)** is a measure of the amount of x-rays the digital image receptor actually received during the making of a particular radiograph. EI was standardized in

2018 and can be used to compare the sensitivities of different digital systems. EI is especially valuable to determine whether the exposure used was too high. In these cases, EI can be used to limit **dose creep**. Dose creep refers to the unnecessarily high radiation doses that frequently are encountered in digital radiography (dose creep is discussed later in this chapter).

EI is directly related to the x-ray exposure dose. With most digital radiography systems, it should be in range of 1700–2100. If the exposure is doubled, the EI increases by 300. If the exposure is cut in half, the EI decreases by 300. Here's how EI can be used: suppose a particular radiographic study resulted in an EI of 2500. An EI of 2500 is above the desired range of 1700–2100. This means the exposure used was too high and the radiograph should be overexposed. However, most digital systems can still display a good quality radiograph from an overexposure, so the image may appear acceptable. But the radiation dose to the patient and nearby personnel was too high. Overexposures need to be discussed with staff and prevented whenever possible. Any future

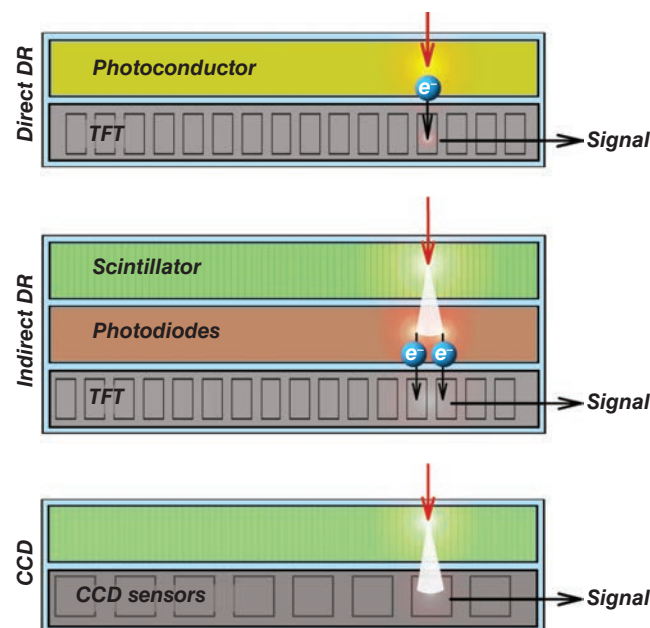


Figure 1.20 Types of DR image receptors. These illustrations highlight the different methods of converting x-ray energy into electronic signals, which are then sent to be digitized and processed by a computer. **Direct DR systems (A)** capture x-ray energy using a photoconductor which directly translate the energy into electrical charge (e^-). The TFT array converts the electrical charge to electronic signals. **Indirect DR systems (B)** use a scintillator to convert captured x-ray energy into visible light. The thin, needle-shaped crystals in the scintillator channel the light to photodiodes, where it is converted into electrical charge (e^-). Electrical charge is then converted to electronic signals by the TFT array. **CCD (C)** also uses a scintillator to convert captured x-ray energy to light, but special sensors then convert both the light into electrical charge and the electrical charge into electronic signals.



Figure 1.21 DR image receptor. This picture depicts a type of portable DR image receptor that can be positioned under the x-ray table or table-top and can be used for positional radiography. It is shown with a cable to link it to a computer, but some models are available as wireless units.

images of this patient should be made using lower exposure. To bring an EI of 2500 into the desired range, we would cut the mAs in half (which reduces the EI by 300 to an EI of 2200) and cut the mAs in half again (which reduces the EI another 300 to an EI of 1900, which is in the desired range).

Sensitivity number (S-number) is inversely proportional to x-ray exposure. As exposure increases, the S-number decreases. For most radiographic studies, the S-number should be between 200 and 300. For example, a radiograph made with an S-number of 100 is below the desired range of 200–300. This means the exposure used was too high (about twice what was needed). Future radiographs should be made with half the mAs, which would make the S-number 200 and in the acceptable range. Another example: a radiograph made with an S-number of 1200 is above the desired range of 200–300, which means the exposure was too low. Future exposures should be higher. Doubling the mAs reduces the S-number to 600 and doubling the mAs again decreases the S-number to 300, which is in the desired range.

The amount of **radiographic contrast** a digital receptor can display is determined by its **pixels**. “Pixel” is short for *picture element*. Pixels are the smallest pieces of a digital radiograph. Each pixel can appear as any of several different shades of gray, depending on its pixel number. Pixel numbers are **binary numbers**. Binary is a “base 2” numbering system that is used by computers, as opposed the “base 10” numbering system we humans use in our everyday lives (i.e., a decimal system). Each digit in a binary number is called a **bit**. “Bit” is short for *binary digit*, and it is the smallest unit of digital data. Each bit is a digit in a binary number and is assigned a value of either 0 or 1. The position of each bit in the binary number represents an exponent of 2. In the

decimal system, the position of each digit represents an exponent of 10.

In digital radiography, the contrast in an image can be adjusted after the radiograph has been acquired. This is a tremendous advantage over film radiography. Digital contrast can be optimized using the large amount of data in each pixel. Computer processing and user manipulations determine the best shades of gray for the pixels to display. However, for a computer to produce quality radiographs, it requires a properly calibrated image receptor and correct input from the user. The method of receptor calibration usually is built into each manufacturer’s system. Calibrations should be done per the manufacturer recommendations prior to radiography to correct any defects, imperfections, and non-uniformities in the image receptor.

Geometry of the x-ray beam

It is essential to understand the effects of x-ray beam geometry and how to use them to make quality radiographs. The radiographic appearance of an object is greatly influenced by its position and orientation relative to the x-ray beam and to the image receptor.

Focal spot

The focal spot is the source of x-rays. It is the point on the anode target that is struck by the high-speed stream of electrons. The focal spot, however, is not a tiny dot without length or width; it has dimensions. X-rays are emitted from all parts of the focal spot and travel in straight lines in all directions. The x-rays coming from one side of the focal spot overlap with those coming from the other side. This overlap makes the edges of objects appear blurry in the radiograph (Figure 1.22). Edge-blurring is called **geometric unsharpness**. Geometric unsharpness reduces radiographic detail, making it more difficult to distinguish two objects that are close together. Edge-blurring sometimes is called *penumbra*, which is an astronomy term used to describe the fuzzy-edged shadow cast by the moon during a partial eclipse. The term *umbra* refers to the more sharply defined center part of the shadow.

The degree of geometric unsharpness depends on the ratios of the distances between the focal spot, the object, and the image receptor and on the size of the focal spot. These ratios are explained in greater detail when we discuss distortion later in this chapter. The size of the focal spot may be altered by changing its physical size or its effective size. The smaller the focal spot, the less geometric unsharpness and the better the radiographic detail.

The **physical size of a focal spot** is fixed by the x-ray tube manufacturer. Most x-ray tubes are equipped with two focal spots; a small one and a large one (Figure 1.23). A small focal spot typically is about 0.5–1.0 mm in size and a large

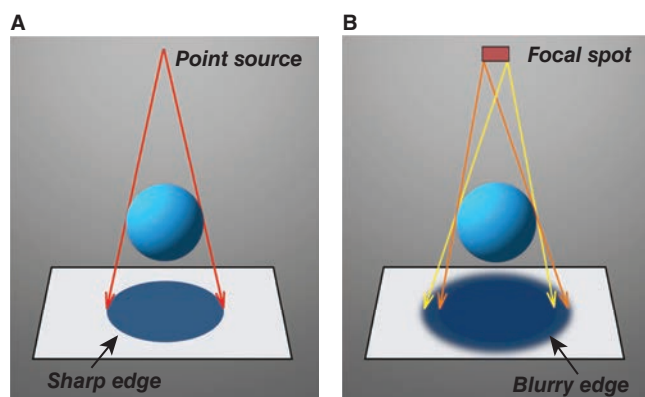


Figure 1.22 Geometric unsharpness. If x-rays originated from a point source, as shown in illustration **A**, the margin of the object would appear sharp in the image below. However, x-rays originate from a focal spot with dimensions, as shown in illustration **B**. X-rays coming from one side of the focal spot (orange arrows) overlap with those coming from the other side (yellow arrows), which causes the object margin to appear blurred in the image below.

focal spot is about 1.0–2.0 mm. The small focal spot produces better radiographic detail than the large focal spot, but it is less tolerant of heat. Fewer and less energetic x-rays can be produced at the small focal spot, which generally limits its use to body parts measuring less than 10–12 cm thick. The large focal spot is more heat tolerant and allows greater x-ray exposures, shorter exposure times, and imaging of larger body parts, but there is less radiographic detail.

The **effective size of the focal spot** can be altered using the *line focus principle*. The line focus principle states that viewing a sloped surface at an angle reduces its apparent size. This is why the anode disc is beveled. Tilting the anode target makes the focal spot appear smaller (Figure 1.24).

The effective focal spot size also can be reduced by moving the x-ray tube further away from the image receptor. With increased distance, the focal spot more closely resembles a point source. The greater the distance, the better the radiographic detail. However, as you increase the distance, the x-ray exposure must be increased exponentially to maintain radiographic density. If you double the distance, you will need four times the x-ray exposure due to the *inverse square law*.

Inverse square law

The inverse square law occurs because x-rays diverge as they move away from the focal spot, spreading out over larger and larger areas (Figure 1.25). This means that there are fewer x-rays per unit area (less x-ray beam intensity) as you move further away from the source. The decrease in x-ray beam intensity is inversely proportional to the square of the distance from the focal spot: $I = 1/D^2$, where “*I*” is the intensity of the x-ray beam and “*D*” is the distance from the focal spot.

Heel effect

Because the target is angled to create a smaller effective focal spot, the bottom of the anode is thicker. The thicker bottom commonly is called the “heel” of the anode. The heel absorbs some of the x-rays produced at the focal spot. Absorption results in fewer x-rays being emitted from the anode side of the x-ray tube and more x-rays coming from the cathode side. The heel effect can produce as much as a 40% difference in x-ray beam intensity between the anode side and the cathode side of the x-ray tube (Figure 1.26). This difference is useful when imaging body parts that vary in thickness or opacity. The thicker or more opaque end of the body part is positioned toward the cathode side where the x-ray beam is more penetrating. Remember, the word “cathode” has more letters than the word “anode” and more x-rays come from the cathode side than the anode side. The heel effect is most noticeable when using a:

1. Large FOV.
2. Lower x-ray exposure.
3. Short distance between the x-ray tube and the image receptor.

Distortion

Distortion occurs when an object’s actual size or shape is misrepresented in the radiograph. It is caused by equal or unequal magnification of the object. Equal magnification makes an object appear overall larger in the radiograph than its actual size. Unequal magnification causes *foreshortening* or *elongation* distortion which are described below. Any form of

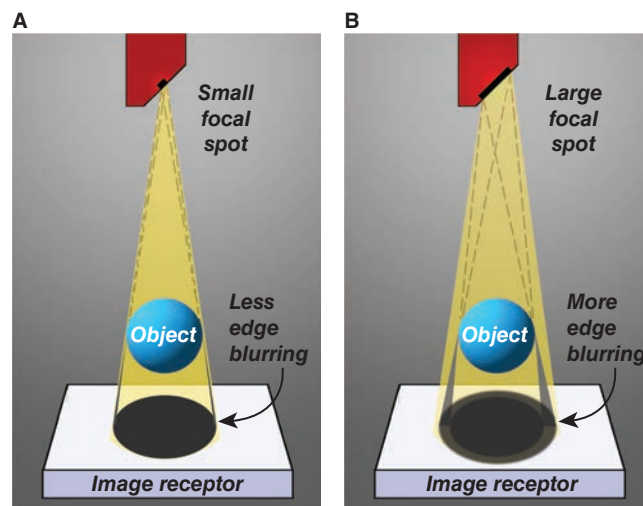


Figure 1.23 Physical size of the focal spot. These illustrations depict the effect of focal spot size on edge sharpness. The anode is shown in red at the top of each image. **(A)** With a small focal spot there is less overlap of the x-rays and less blurring of the object margin in the image below. **(B)** With a larger the focal spot there is greater overlap of x-rays and more geometric unsharpness.

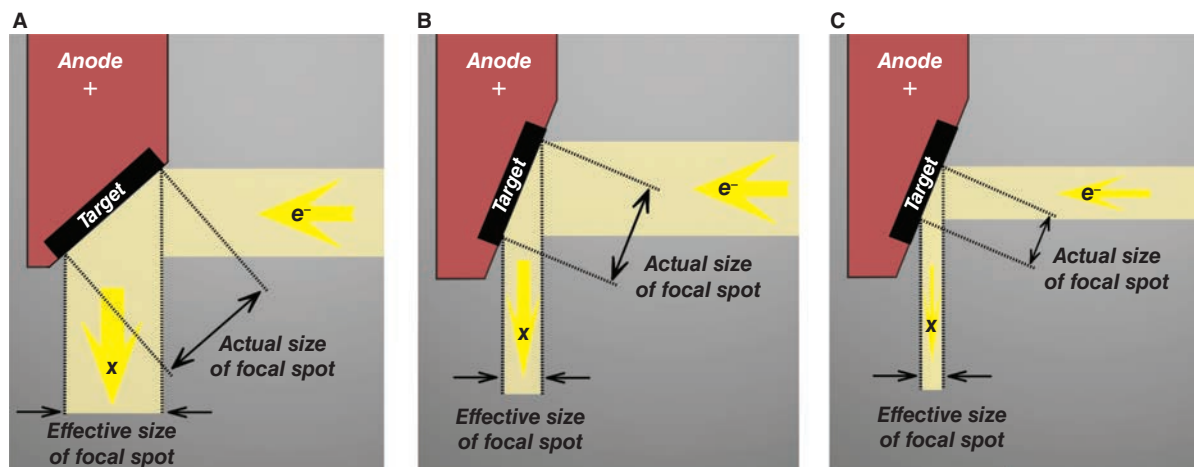


Figure 1.24 Line focus principle. These illustrations depict variations in the effective size of the focal spot due to the slope of the anode target and the size of the electron beam (e^-). The steeper the slope, the smaller the effective size of the focal spot. The actual size of the focal spot is determined by the size of the electron beam. (A) 45° slope. (B) 20° slope. (C) The actual size of the focal spot is smaller because the stream of electrons is smaller (due to using a smaller filament). x = x-rays.

magnification also causes geometric unsharpness. There are three distances that are important in distortion:

1. Distance between the focal spot and the image receptor: the **source-to-image distance** or **SID**.
2. Distance between the focal spot and the object: the **source-to-object distance** or **SOD**.
3. Distance between the object and the image receptor, the **object-to-image distance** or **OID**.

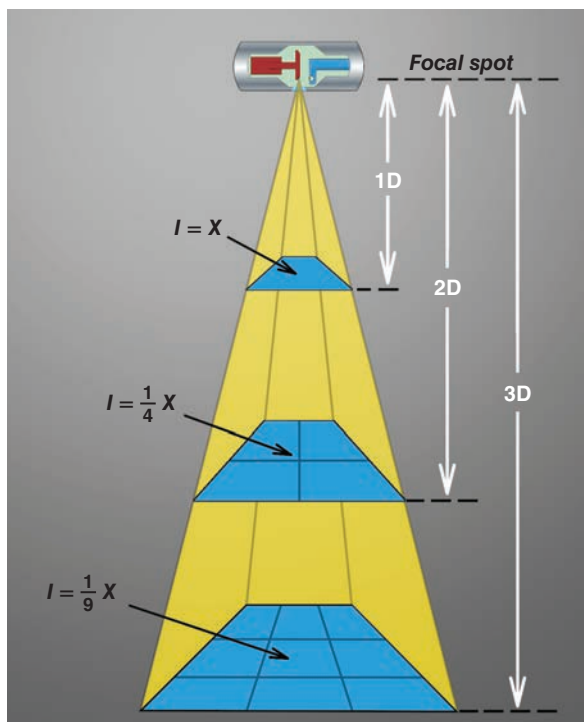


Figure 1.25 Inverse square law. This illustration depicts the divergence of x-rays as they move further from the focal spot. The x-rays spread out over a larger and larger area as the distance (D) increases. The intensity (I) of the x-ray beam is inversely proportional to the square of the distance from the focal spot: $I = 1/D^2$. Doubling the distance decreases the x-ray beam intensity to $1/4$ x.

It is the ratio between any two of these distances that is important when dealing with magnification and geometric unsharpness (Figure 1.27). To minimize magnification, we want the object to be as close to the image receptor as possible (OID as short as practical). The degree of magnification is determined by the ratio of SID to SOD (SID/SOD is called the *magnification factor*). The closer the ratio is to 1.0, the less magnification (Figure 1.28).

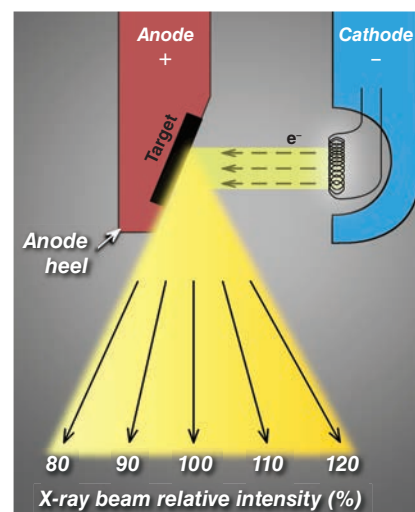


Figure 1.26 Heel effect. More x-rays are emitted from the cathode side of the x-ray tube than from the anode side because some of the x-rays produced are absorbed in the anode heel. There is a gradual loss of x-ray beam intensity moving from cathode side to anode side.

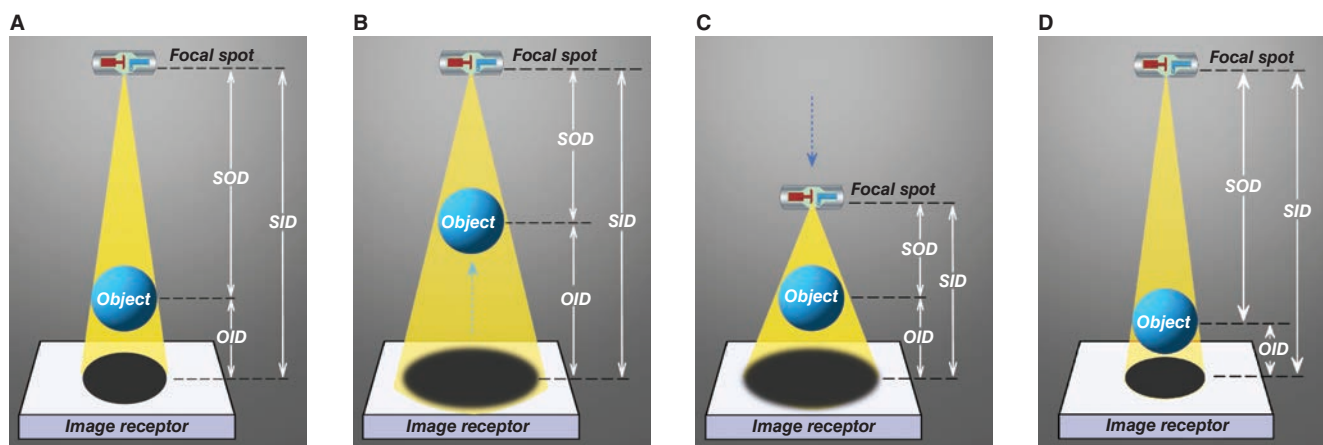


Figure 1.27 Magnification. (A) Source-to-image distance (SID), source-to-object distance (SOD), and object-to-image distance (OID) are illustrated. (B) Increasing the OID (moving the object further away from the image receptor) while maintaining the same SID creates magnification distortion and geometric unsharpness. (C) Shortening the SID (moving the x-ray tube closer to the image receptor) while maintaining the same OID also creates magnification. (D) Shortening the OID (placing the object closer to the image receptor, such as using a table-top-technique instead of under the table) and using the same SID will lessen magnification.

Sometimes magnification is desired. Here is an example: suppose we want to make a radiograph that displays the magnification of a paw. We position the patient's foot on a cardboard box or foam block to move it further away from the image receptor, increasing the OID. This will cause magnification and geometric unsharpness. To minimize the edge-blurring, we use the small focal spot and move the x-ray tube further away, increasing the SID. Because the SID is longer, the x-ray exposure needs to be increased due to the inverse square law, so we increase the mAs.

Both **elongation distortion** and **foreshortening distortion** occur when the position of an object is oblique to the x-ray beam or image receptor. Oblique positioning makes the OID longer on one side of the object than the other. When one side is magnified more than the other the shape of the object is distorted. Unequal magnification also can occur if the image receptor is obliquely positioned in relation to the object or x-ray beam. Foreshortening distortion makes an object appear overall shorter in the radiograph than its actual size (Figure 1.29). Elongation distortion makes an object appear overall longer than its actual size (Figures 1.30 and 1.31).

Elongation also occurs in objects that are off-center to the x-ray beam. Because x-rays diverge, objects nearest the center of the beam will be imaged more accurately than objects at the periphery of the beam. All peripheral objects in a radiograph suffer some degree of elongation distortion. For example, when imaging the spine only the vertebrae and intervertebral disc spaces in the center of the image are accurately depicted for size and shape. The peripheral vertebrae are more elongated than actual, which may be mistaken for

narrowed intervertebral disc spaces. Another example is the appearance of the cardiac silhouette at the periphery of an abdominal radiograph. Due to elongation distortion the cardiac silhouette is distorted and may appear longer than actual, which may be mistaken for cardiomegaly.

To minimize distortion, position the object of interest as close to the image receptor as practical (minimize the OID), align the object with the center of the x-ray beam, and collimate the FOV as small as practical. When interpreting radiographic findings, be aware of distortion artifacts.

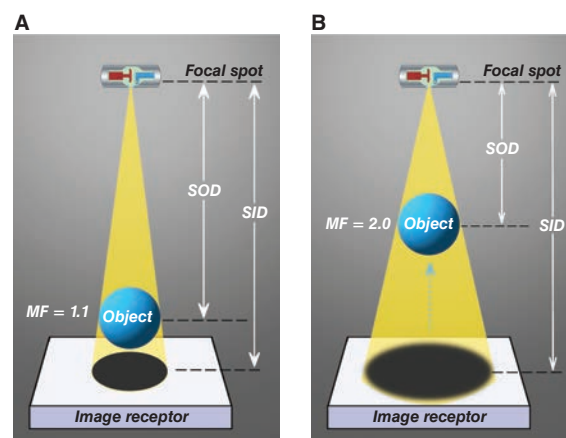


Figure 1.28 Magnification factor. Illustration A depicts an object close to the image receptor with little magnification. The SID and SOD are similar resulting in a magnification factor (MF) near 1.0. In illustration B, the object is further from the image receptor, resulting in magnification and geometric unsharpness. The object appears twice as large in the image as actual size (magnification factor is 2.0).

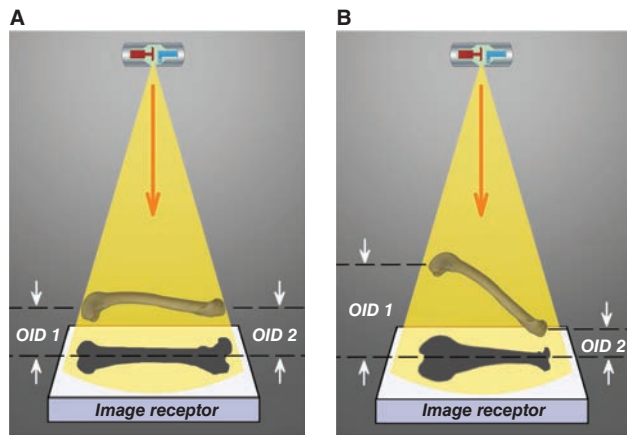


Figure 1.29 Foreshortening distortion. **A.** The long axis of the bone is parallel with the image receptor and perpendicular to the x-ray beam and therefore its size and shape are accurately represented in the image below. **B.** The bone is tilted in relation to the image receptor and oblique to the x-ray beam so that it appears shorter than actual and abnormal in shape due to unequal magnification.

X-ray interactions with matter

Each x-ray that enters a material will result in one of three possible outcomes: (1) it may be *transmitted*, passing through the material unchanged, (2) it may be *absorbed* and completely disappear inside the material, or (3) it may be *scattered*, becoming a secondary x-ray that travels in a new direction with less energy (Figure 1.32). Transmitted and absorbed x-rays are valuable to make radiographs. Scatter x-rays provide no useful diagnostic information, degrade image quality, and expose the patient and nearby personnel to unnecessary radiation.

Transmitted x-rays that reach the image receptor and are recorded produce useful blackened areas in a radiograph (i.e., radiographic density). The more transmitted x-rays, the darker that part of the image. The likelihood of an x-ray being transmitted is affected by its energy. The higher the energy, the more penetrating the x-ray. Recall that x-ray energy is controlled by kVp, so the higher the kVp, the more transmitted x-rays and the darker the radiograph. The likelihood of x-ray transmission also is affected by the opacity of the material. The greater the opacity, the fewer x-rays will pass through. Higher opacity materials appear lighter in radiographs. The mAs also will affect the number of transmitted x-rays because mAs affects the overall quantity of x-rays produced. The higher the mAs, the greater the number of transmitted x-rays.

Absorbed x-rays never reach the image receptor and therefore produce no radiographic density. The more absorbed x-rays, the lighter or brighter that part of the image. In medical radiography, x-ray absorption primarily occurs via the *photoelectric effect*.

The **photoelectric effect** is similar to characteristic x-ray production (Figure 1.5) except it is an x-ray instead of a

high-speed electron that collides with an atom. The x-ray transfers all of its energy to an inner level electron, ejecting the electron and ionizing the atom (Figure 1.33). The ejected electron travels a short distance before losing its energy to other atoms along its path. Both the ionization of the atom and the energy from the ejected electron can damage or destroy living cells. Ionized atoms are unstable so a higher energy outer level electron drops down to fill the void in the inner level. The outer level electron loses energy in the form

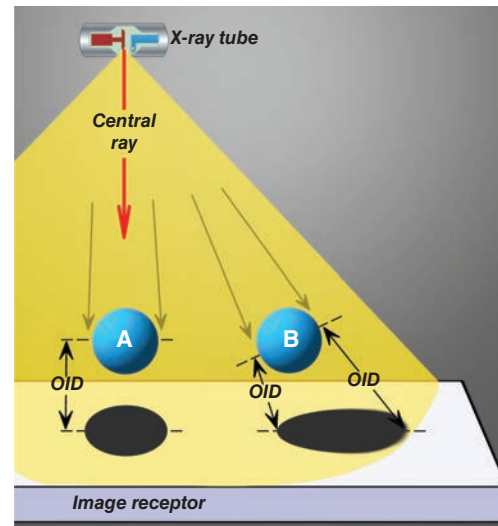


Figure 1.30 Elongation distortion. Object **A** is aligned with the central ray (red arrow) resulting in an accurate depiction of its size and shape in the image below. Object **B** is off-center, peripheral to the central ray, making it appear longer and more ovoid than actual in the image below.

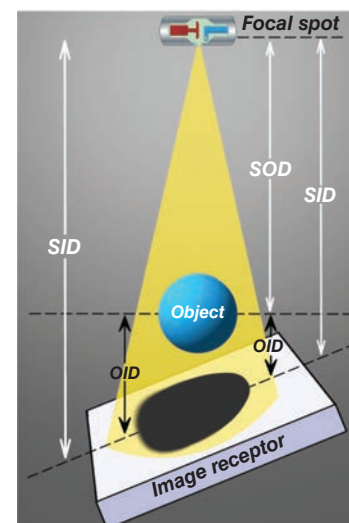


Figure 1.31 Elongation distortion. In this illustration the image receptor is tilted in relation to the x-ray beam and object resulting in unequal magnification. The OID is longer on one side than the other, which distorts the shape of the object and causes geometric unsharpness on that side.

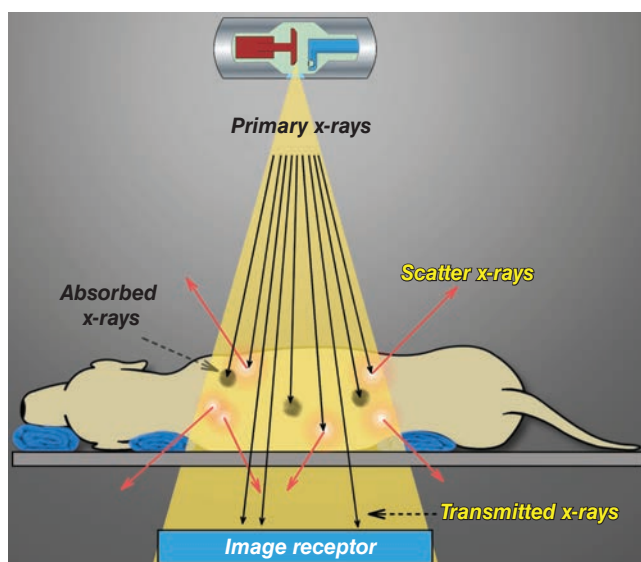


Figure 1.32 X-ray interactions. X-rays produced in the x-ray tube are called *primary* x-rays. Primary x-rays that pass through the patient unchanged are *transmitted* x-rays. Primary x-rays that interact with atoms in the patient may either be *absorbed* (lose all of their energy and completely disappear) or be *scattered* (become a secondary x-ray that travels in a new and unpredictable direction, as shown by the red arrows).

of an x-ray. This new x-ray is a *secondary* x-ray because it was not produced in the x-ray tube (primary x-rays are emitted from the tube). The energy of the secondary x-ray is equivalent to the difference between the outer level and inner level binding energies. In most cases, the secondary x-ray is absorbed without further damage to cells.

The likelihood of x-ray absorption via the photoelectric effect depends on the size of the atoms (i.e., atomic number) and the energy of the x-rays (i.e., kVp). The larger the atomic number, the greater the chance of x-ray absorption. Doubling the atomic number increases the probability of x-ray absorption by a factor of 8. The higher the kVp, the less the chance of x-ray absorption because the x-rays are more penetrating. Doubling the x-ray energy decreases the probability of absorption by a factor of 8.

The photoelectric effect is a major cause of x-ray attenuation up to about 70 kVp. Above 70 kVp the probability of x-ray absorption decreases, except at the **K-edge**. At the K-edge there is a sudden increase in absorption. K-edge refers to the binding energy of an inner level electron (the inner level of an atom is called the *K-shell*). X-rays with energy that is slightly greater than the binding energy of an inner level electron (just above the K-edge) are more likely to be absorbed (Figure 1.34).

The K-edge can be quite useful when imaging materials with a high atomic number, particularly positive contrast media (e.g., barium, iodine). Materials with high atomic numbers contain more K-shell electrons and therefore are

more likely to absorb x-rays with a certain energy. The K-edge is not nearly as useful in soft tissues because these materials are composed of elements with smaller atomic numbers and fewer K-shell electrons (e.g., carbon, hydrogen, oxygen, nitrogen). The K-edges of barium and iodine are around 35 keV which is about the effective energy of a 70 kVp x-ray beam. This is one reason why barium and iodine are such ideal positive contrast agents for medical radiography. When performing a positive contrast study try to keep the kVp as close to 70 as practical to enhance x-ray absorption. The greater the x-ray absorption, the greater the contrast.

Scatter x-rays are secondary radiation. During most radiographic procedures, about 95% of scatter comes from the patient and the rest from the x-ray table, image receptor, floor, or any other material that happens to be in the x-ray beam FOV.

Scatter x-rays result from the incoherent or coherent interactions of primary x-rays. Most scatter is caused by **incoherent scattering**, which is also called the *Compton effect*, *inelastic scattering*, or *modified scattering*. Incoherent scattering occurs throughout the range of x-ray energies used in medical imaging (40–125 kVp). An x-ray collides with an atom, ejecting an electron and ionizing the atom (Figure 1.35). The x-ray energy is transferred to the ejected electron and to a secondary x-ray. The secondary x-ray travels in a completely random

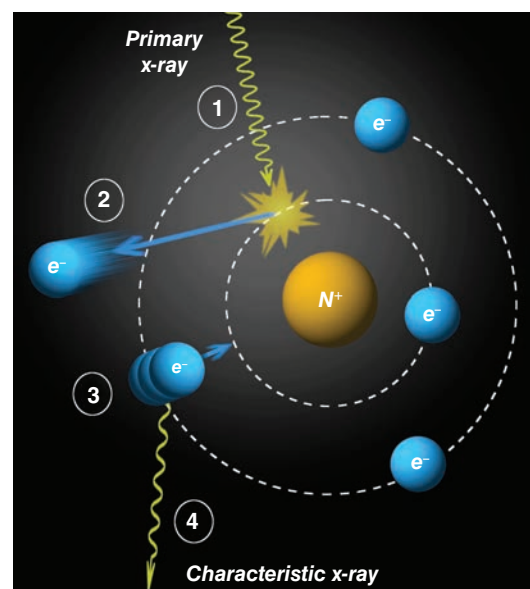


Figure 1.33 Photoelectric absorption. 1. A primary x-ray collides with an atom in the patient. The x-ray transfers all of its energy to an inner shell electron and completely disappears. 2. The electron now has enough energy to escape the atom and is ejected. 3. An outer shell electron drops down to fill the void in the inner shell. 4. The outer shell electron emits excess energy as a new x-ray. The energy of the secondary x-ray is less than that of the primary x-ray (note the longer wavelength). N^+ = nucleus; e^- = electrons.

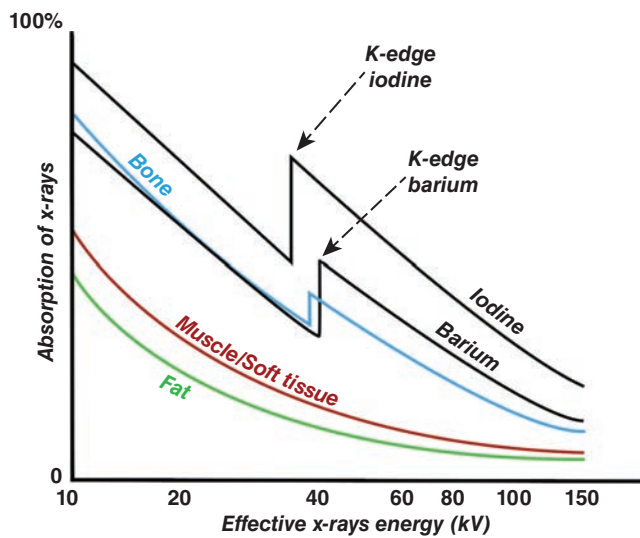


Figure 1.34 K-edge absorption. This graph depicts the relative abilities of fat, soft tissue, bone, barium, and iodine to absorb or attenuate x-rays. The differences in attenuation are highest with lower x-ray energies (lower kV). Radiographic contrast between the different tissues is greatest with lower energy x-rays. As x-ray energy increases, the x-rays are more penetrating and less likely to be absorbed. There is less contrast between tissues when using higher energy x-rays. However, at the K-edges of iodine, barium, and to a lesser extent bone, there is a sudden increase in x-ray absorption. At these parts of the graph there is a significant difference in attenuation between tissues which can be used to maximize radiographic contrast.

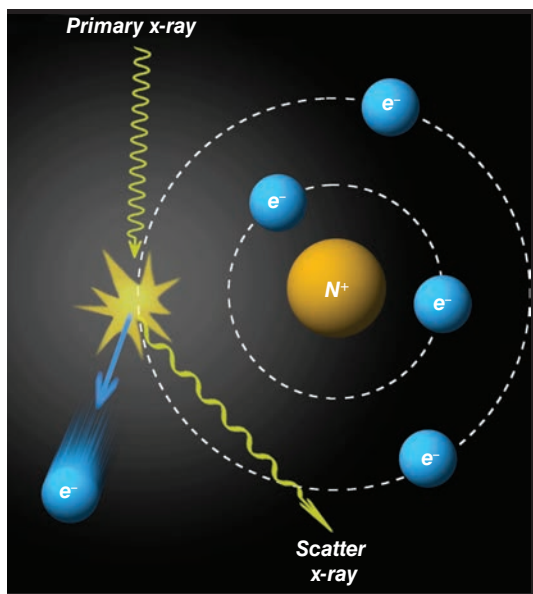


Figure 1.35 Incoherent scattering. A primary x-ray collides with an atom, ejecting a loosely bound outer level electron. The x-ray energy is transferred to the electron and to a secondary x-ray. The scatter x-ray travels in a new and unpredictable direction with less energy (longer wavelength) than the primary x-ray.

Box 1.5 To lessen the amount of scatter fog, minimize the x-ray interactions with tissue

- Make the FOV as small as practical
- Use grid when body part is thicker than 10–12 cm
- Use air gap when grid is not available

direction, anywhere from 0° to 180° from the original x-ray's path. Most incoherent scattering tends to be in the forward direction, but it is wholly unpredictable. If the secondary x-ray travels back toward the x-ray tube it is called **back-scatter** radiation.

Coherent scattering occurs with low-energy x-rays, less than 10 kVp. It is also called *elastic*, *unmodified*, *classical*, *Rayleigh* scattering, or the *Thompson effect*. Coherent scattering usually is not significant in medical imaging, but it is mentioned here because even very low energy x-rays can interact with atoms in the body. This type of scattering occurs when an x-ray collides with a firmly bound electron in an atom and creates a secondary x-ray. The electron is not ejected and the atom is not ionized. The secondary x-ray continues in a new direction with the same energy.

Scatter x-rays are a major cause of poor quality radiographs. Most of the x-rays emerging from the patient are scatter (50%–90%). Unfortunately, an image receptor cannot distinguish between scatter and transmitted x-rays. Scatter increases radiographic density, but does not represent anything inside the patient. Scatter causes a diffuse, unwanted darkening of the radiograph which is called **fog**. Fog reduces contrast and visible detail. We discussed earlier some of the causes of film fog as it pertains to film:screen radiography. Scatter fog is a much bigger problem and affects both film and digital systems. In addition to reducing image quality, scatter x-rays also expose the patient and nearby personnel to unnecessary radiation. Though less energetic than primary x-rays, many scatter x-rays retain sufficient energy to ionize atoms and damage cells.

The primary contributor to the amount of scatter radiation is the volume of tissue being irradiated (Box 1.5). Tissue volume is determined by the **field-of-view** (FOV) and the **thickness of the body part**. The larger the FOV, the more scatter (Figures 1.36 and 1.37). The thicker the body part, the more scatter. Some of the most effective and practical methods to limit the amount of scatter include making the FOV as small as possible and using a grid whenever body part thickness exceeds 10–12 cm.

Grids

A **grid** is one of the most effective and practical methods of minimizing scatter fog and enhancing radiograph quality.

Grids are devices that are placed between the patient and the image receptor to absorb scatter x-rays (Figures 1.38

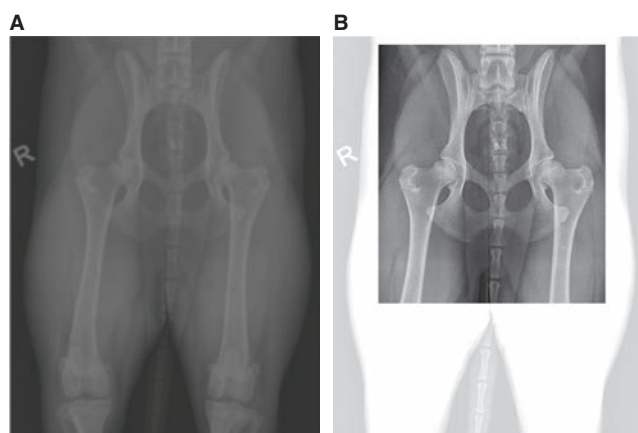


Figure 1.36 Scatter fog. **A.** VD view of a dog pelvis depicting the effect of scatter fog. There is an overall increase in radiographic density with poor contrast and less visible detail due to detail. **B.** VD view of the same dog made with a smaller FOV. The radiograph is better quality due to less scatter fog.

and 1.39). A grid consists of alternating strips of an x-ray absorbing material, typically lead, and a non-absorbing material such as carbon fiber or aluminum. The strips may be arranged in a parallel or cross-hatched pattern, or they may be focused toward the x-ray tube so they align with the x-ray beam. A focused grid is more likely to absorb x-rays that are not aligned with the primary beam (scatter x-rays) and less likely to absorb x-rays that are aligned with the beam (transmitted x-rays). Focused grids must be positioned in the center of the x-ray beam and used at a specific distance from the focal spot (Figure 1.40).

Grids are most useful when larger volumes of tissue are irradiated. They should be used whenever body part thickness exceeds 10–12 cm. The effectiveness of a grid (the amount of scatter radiation it can absorb) is affected by its *grid ratio* and the *grid frequency* (Figure 1.41).

The **grid ratio** is the height of a lead strip divided by the distance between two strips. The higher the ratio, the more scatter will be absorbed (Figure 1.42). High ratio grids can absorb up to 90% of scatter radiation. Unfortunately, high ratio grids also absorb as much as 30% of transmitted x-rays. A grid ratio must be high enough to remove most of the scatter but not so high that there aren't enough transmitted x-rays to make a quality radiograph. In most practice situations, a grid ratio of 8 : 1 is appropriate for x-ray exposures in the range of 70–90 kVp.

Grid frequency is the number of lead strips per unit distance. Frequency generally is expressed as *lines per cm (lpc)* or *lines per inch (lpi)*. The higher the grid frequency, the more scatter will be absorbed. Low-frequency grids (40–50 lpc or 100–120 lpi) generally are used in x-ray systems with a moving grid (see Bucky-Potter mechanism below). Medium frequency grids (50–60 lpc or 120–150 lpi) are suitable for most veterinary practices and can be used in stationary grid

holders. High-frequency grids (more than 60–70 lpc or 150–170 lpi) may be preferred for some digital imaging systems to help reduce grid artifacts.

Because the grid is physically located between the patient and the image receptor, its lead strips sometimes are visible in the radiograph. The strips appear as thin, parallel, white lines that are regularly-spaced across the entire radiograph (see Artifacts in Chapter 2). These lines are called **grid lines**. They tend to be less noticeable with higher frequency grids. Grid lines can be blurred using a Bucky-Potter mechanism, which makes them less distracting in the radiograph.

The grid was invented in 1913 by a German radiologist named Gustav Bucky. Shortly thereafter, an American radiologist named Hollis Potter designed a device to blur the grid

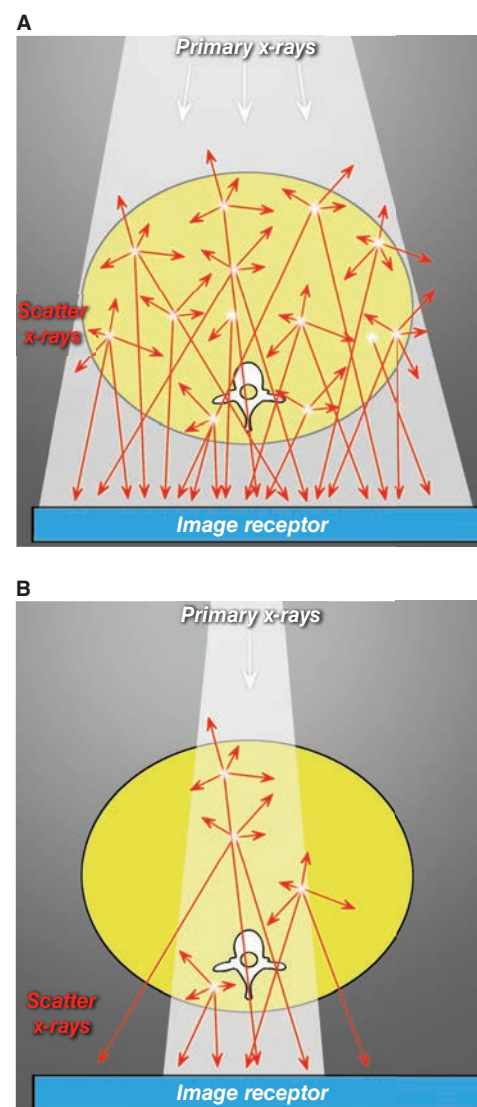


Figure 1.37 Scatter and FOV. These illustrations depict the amount of scatter radiation created by using a larger FOV (**A**) and a smaller FOV (**B**) with the same body part thickness. The size of the FOV is a primary determinant of the amount of scatter radiation produced.

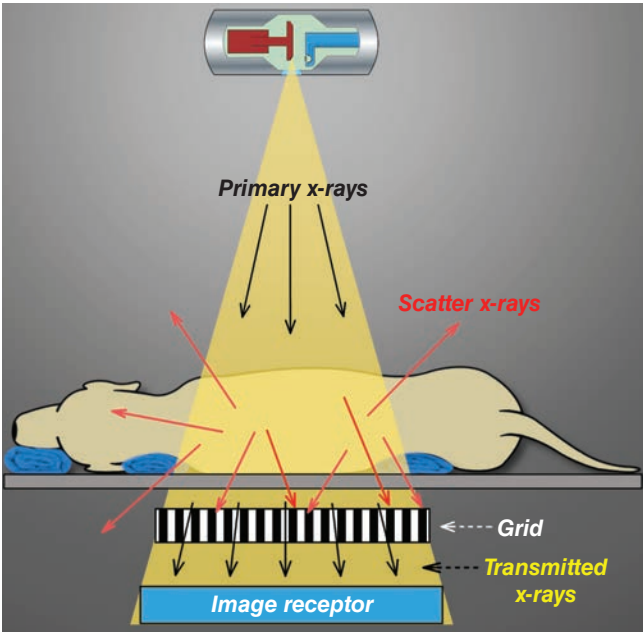


Figure 1.38 Grid. This illustration depicts a grid positioned between a patient and the image receptor. The purpose of the grid is to block unwanted scatter x-rays (red arrows) from reaching the image receptor while allowing transmitted x-rays (black arrows) to pass through.

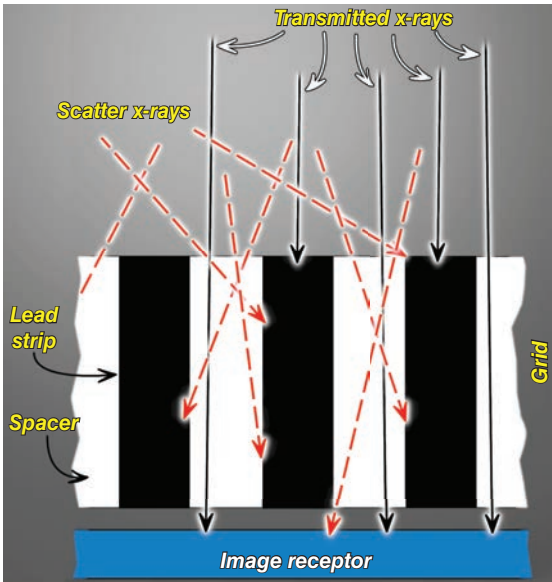


Figure 1.39 Grid close up. This illustration depicts the method by which a grid reduces scatter fog. Scatter x-rays travel in various directions and most are not incident to the image receptor. Many scatter x-rays are absorbed by the lead strips in the grid, but some will pass through the spacers to reach the receptor. Transmitted x-rays travel in straight lines incident to the receptor. These ideally pass through the spacers, but some will strike a lead strip and be absorbed.

lines. The device, now known as the **Bucky-Potter mechanism** or the *Potter-Bucky diaphragm*, moves the grid during the x-ray exposure, oscillating it at right angles to the lead strips. The speed at which the grid moves determines the shortest exposure time possible to avoid grid lines. Moving the grid during exposure is most valuable when using a low ratio grid or a low-frequency grid. However, the movement may startle some patients and create movement artifacts in the radiograph.

Ideally, all transmitted x-rays pass through a grid to reach the image receptor. Unfortunately, the grid absorbs many of them. The amount of transmitted x-rays that reach the grid compared to the number that pass through the grid is called the **grid factor** (also known as the *Bucky factor*). The grid factor determines the additional x-ray exposure needed when using a grid compared to not using a grid. For example, a grid factor of 2 means that for every two x-rays that reach the grid, one is absorbed and one passes through. A grid factor of 2 means the x-ray exposure must be doubled when using the grid, usually twice the mAs. The grid factor increases as the grid ratio or grid frequency increases. In digital radiography, software is available to compensate for scatter fog which may reduce the need for a grid.

Air gap

The air gap technique takes advantage of the fact that scatter x-rays are lower energy and more divergent than transmitted x-rays. Moving the image receptor further away from the patient creates a space that allows more of the scatter x-rays to dissipate in the air gap or miss the receptor (Figure 1.43). A 10–15 cm spacing between patient and image receptor has been shown to be effective in reducing scatter fog. The tradeoff with using an air gap is the object-to-image distance (OID) is increased which results in magnification and geometric unsharpness.

Radiographic density

Density in a radiograph is the amount of black in the image. The darker areas in the radiograph are high density and the whiter areas are low density. Each shade of gray represents a level of density.

The term “density” may be a source of confusion. In the early days of radiology, “density” was used to describe both the physical density of the material being imaged (i.e., its weight per unit volume) and the concentration of black metallic silver on the film (Box 1.6). Physical density was called *subject density* and it produced the whiter areas in the radiograph. The amount of black in the film image was called *radiographic density*. However, subject density is only one factor to consider when evaluating the lighter areas in a radiograph. The ability of a



Figure 1.40 Grid. A picture of the front of a grid depicting a typical label that lists the characteristics of the grid including type, composition, ratio, frequency, focal distance, and others.

material to attenuate x-rays is based on more than its physical density alone. Therefore, the more correct term for subject density is radiopacity or simply opacity. Opacity is the ability to attenuate x-rays. Opacity produces the whiter areas in the radiograph and density is the blacker areas. Opacity and density are inversely related: the greater the opacity of a material, the less density it will produce in the radiograph.

Box 1.6 Density in film radiography

In film radiography, radiographic density sometimes is called *optical density* or *transmitted density* because with film, the amount of density is determined by the amount of light that can be seen through the film. A film radiograph with high density permits little light to shine through.

The levels of density in a radiograph are determined by the number of x-rays that are recorded by the image receptor. The number recorded depends on the number that reach the receptor and the **sensitivity** of the receptor. Sensitivity is the minimum quantity of x-rays a receptor must receive to form acceptable levels of density. The number of x-rays that reach the image receptor is determined the number produced (mAs), their energy (kVp), and the opacities of the objects and tissues being imaged. The SID also affects the number because a shorter SID will result in more x-rays reaching the image receptor than a longer SID due to the inverse square law (see Box 1.7, SID and density).

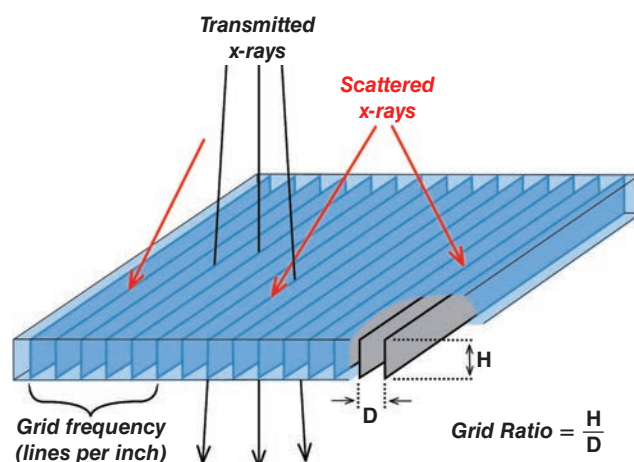


Figure 1.41 Grid. This illustration depicts a cutaway view of a parallel grid with alternating lead and carbon fiber strips. Grid frequency is the number of lead strips per inch. Grid ratio is the height of a lead strip (H) divided by the space between two strips (D).

The larger the quantity of x-rays produced, the more will reach the receptor. Recall that the number of x-rays produced is determined by the **mA** control and the **exposure time**, which together are called **mAs** (*milliamp-seconds*).

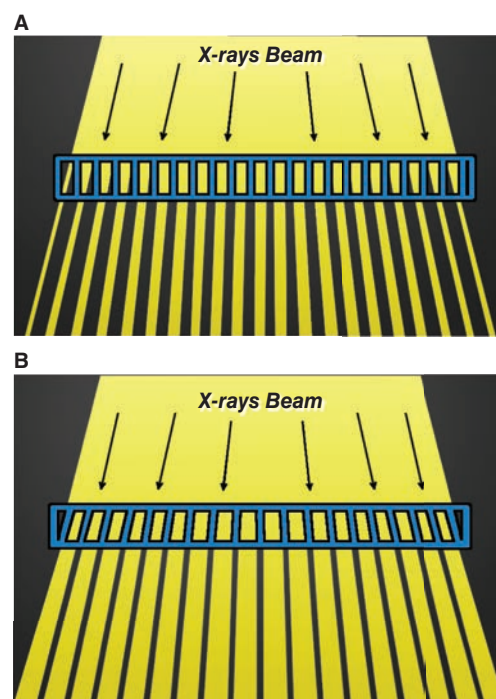


Figure 1.42 Focused grid. These illustrations depict (A) a parallel grid with vertical strips and (B) a focused grid with strips aligned to the x-ray beam. The grids are colored blue and the black streaks below each grid depict areas where the primary x-rays are absorbed by the strips. The parallel grid absorbs significantly more of the primary x-rays than the focused grid.

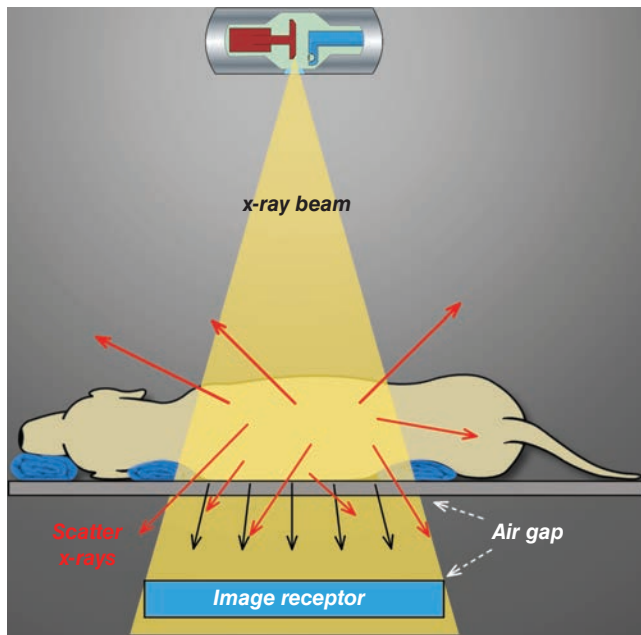


Figure 1.43 Air gap technique. Scatter x-rays (red arrows) are more divergent than transmitted x-rays (black arrows). An air gap is present between the patient and the image receptor to allow more scatter x-rays to miss the receptor.

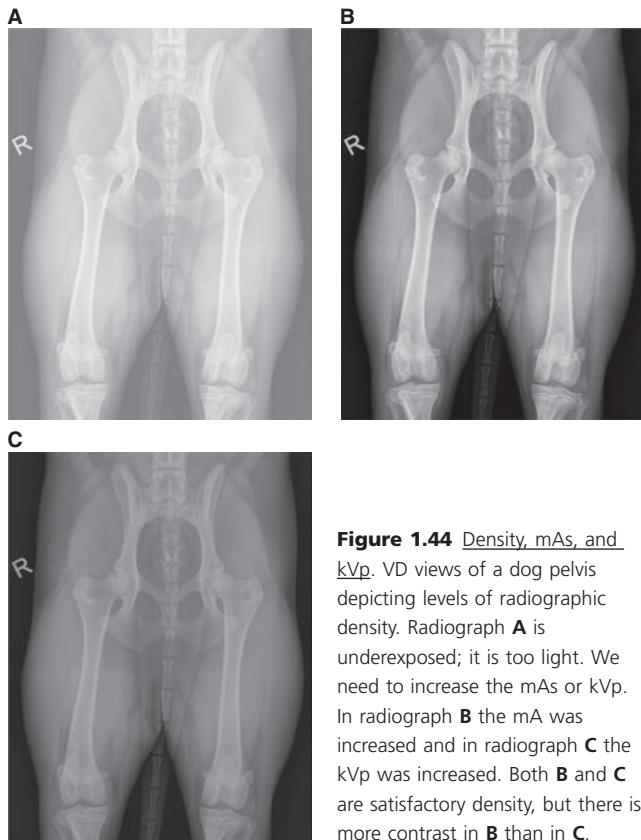


Figure 1.44 Density, mAs, and kVp. VD views of a dog pelvis depicting levels of radiographic density. Radiograph **A** is underexposed; it is too light. We need to increase the mAs or kVp. In radiograph **B** the mAs was increased and in radiograph **C** the kVp was increased. Both **B** and **C** are satisfactory density, but there is more contrast in **B** than in **C**.

The kVp control also affects the number of x-rays, but primarily determines the energy of the x-rays.

When you need to increase or decrease the density in a radiograph, **mAs** is the first adjustment to consider (Figure 1.44). To make the radiograph darker, double the mAs (double either the mA or the exposure time, not both). To make the radiograph lighter, reduce the mAs by 50% (cut the exposure time in half). Adjusting the mAs less than by a factor of 2 usually doesn't cause a visible change in the image.

The higher the energy of the x-rays, the more will reach the receptor. As **kVp** increases the x-rays become more penetrating and more will be transmitted. Sometimes you need to adjust the kVp rather than the mAs to produce acceptable radiographic density. This is particularly true when the exposure time needs to be shortened. A short exposure time is desirable because it helps minimize motion artifacts. A **20% increase** in kVp will darken the radiograph as much as doubling the mAs; a **16% decrease** in kVp will lighten the image as much as halving the mAs.

The greater the **opacity** of the material being imaged, the fewer x-rays will reach the receptor, and the less density will be produced. More opaque materials appear whiter in the radiograph. As an x-ray beam passes through a patient, it encounters objects and tissues with many different opacities. The x-rays that exit the patient exhibit many different x-ray intensities (numbers of x-rays per unit area). The emerging x-ray intensities can differ by 1000-fold or more. To effectively record all of these different intensities, an image receptor needs a wide *dynamic range*.

Box 1.7 SID and density

The SID is inversely related to radiographic density. The longer the SID, the less the density. The shorter the SID, the greater the density. The standard SID in most veterinary practices is 100 cm (40 in.). In some situations the SID needs to be altered, such as when performing mobile radiography or when using a low output x-ray machine. Changing the SID often requires an adjustment in the x-ray exposure to maintain the same density in the radiograph. The following formula is helpful:

$$\left(\frac{\text{SID1}}{\text{SID2}} \right)^2 = \frac{\text{mAs1}}{\text{mAs2}}$$

For example, if the SID must be shortened from 100 to 70 cm and the mAs at 100 cm was 3.0, the mAs at 70 cm must be reduced to 1.5 to achieve the same radiographic density, as shown below:

$$\begin{aligned} \left(\frac{100 \text{ cm}}{70 \text{ cm}} \right)^2 &= \frac{3 \text{ mAs}}{X \text{ mAs}} \\ 2.0 &= \frac{3}{X} \\ 1.5 \text{ mAs} &= X \end{aligned}$$

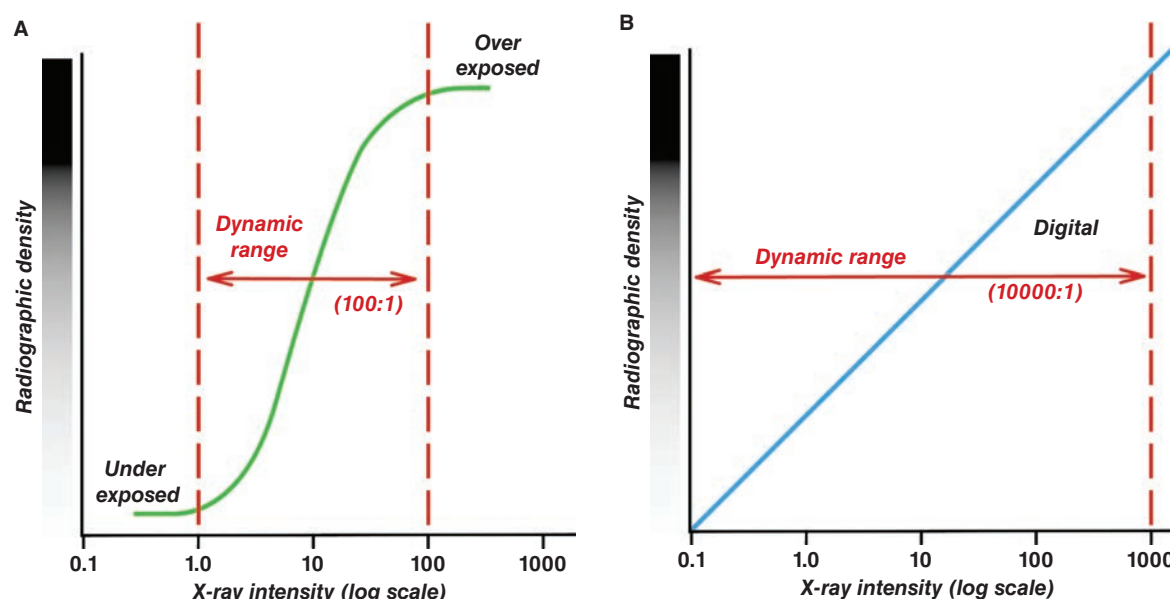


Figure 1.45 Dynamic range of film vs. digital. Characteristic curves for a film:screen image receptor (A) and a digital image receptor (B). Radiographic density is along the vertical axes. These are the visible shades of gray, from minimum to maximum, that the image receptor can produce. X-ray intensity is along the horizontal axes and represents the amount of x-rays the image receptor received. X-ray intensity is displayed as a logarithmic scale because the amounts of x-rays emerging from the patient can vary by a 1000-fold or more and log numbers are easier to work with when dealing with such magnitudes. The curve for the film:screen image receptor (A) is flat along the left bottom, which means no significant density can be formed with this amount of x-rays. This part of the film radiograph would appear too light or underexposed. The top right part of the film:screen curve again is flat, which means no more density can be formed above this amount of x-rays. This part of the film radiograph would be too dark or overexposed. The part of the curve with slope is the area where an image with acceptable density can be formed. The slope is the *dynamic range*. The dynamic range for the film:screen system is 100 : 1, as shown between the two red dotted lines. Notice that the film:screen characteristic curve does not begin at 0. This is because in all x-ray film there is some inherent radiographic density due to the tint in the base and some unexposed emulsion that typically remains. The curve for the digital image receptor (B) is linear because digital systems can accept a much wider range of x-ray intensities and still form visible shades of gray. The dynamic range of this digital image receptor is 10,000 : 1, which is 100 times greater than the film:screen system in (A).

Dynamic range refers to all of the different quantities of x-rays – minimum to maximum – that an image receptor can receive and still form acceptable levels of density. The wider the dynamic range, the more different shades of gray the receptor can display. The more shades of gray, the more different opacities it can display. The dynamic range of an image receptor is described by a *characteristic curve*. **Characteristic curves** are useful to illustrate and compare the capabilities of different image receptors (Figure 1.45).

The dynamic range of digital radiography systems is much wider than the range in film:screen systems, which is why digital receptors are able to produce acceptable radiographs over a wide range of x-ray exposures. A typical film:screen system, for example, has a dynamic range of about 100, whereas the dynamic range of many digital systems is over 10,000.

Sometimes it is useful to have a graph that displays all of the different levels of density and the areas in the radiograph associated with each density. Such a graph is called a **histogram** (Figure 1.46). Histograms reveal the number of pixels with each shade of gray in the image. The wider the histogram, the more shades of gray. The histogram must fit within

the dynamic range of the image receptor or parts of the radiograph will be too light or too dark.

Opacity

The only reason we can see anything in a radiograph is because of differences in opacity. It is the **relative opacity** of one material compared to an adjacent material that allows us to distinguish it or not distinguish it. Differences in opacity are determined by the sizes of the atoms in the materials, how tightly the atoms are packed together (Table 1.1) and the thickness of each material.

In medical radiography, the opacity of each tissue or object is categorized into one of five different groups. Every shade of gray you see in a radiograph is either **gas, fat, fluid/soft tissue, mineral, or metal** opacity. When in doubt, the opacity is decided by comparing it to the opacity of a known material. A material may be known by its anatomical shape and location (Figure 1.47).

Gas opacity materials attenuate very few x-rays and produce the most radiographic density (Figure 1.48). Gas is the

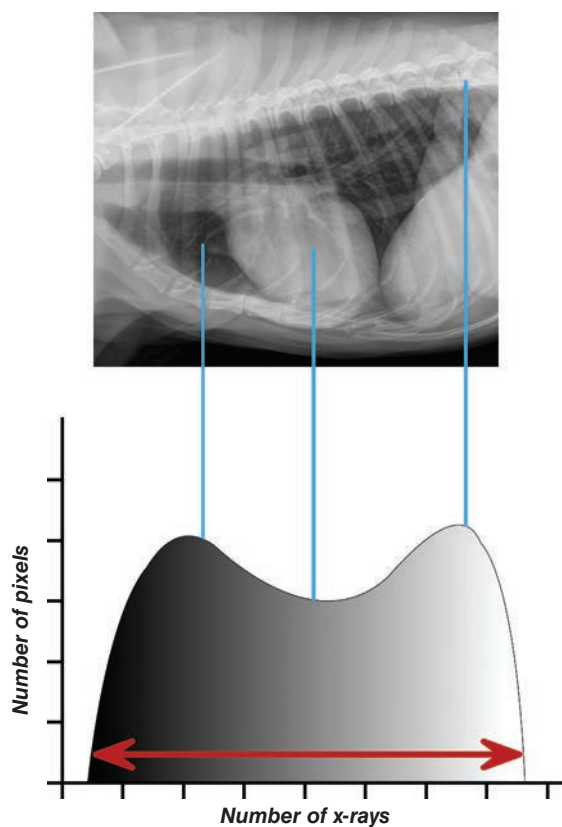


Figure 1.46 Histogram. The histogram displays all of the different radiographic densities (visible shades of gray) in the radiograph. Each shade of gray is determined by the number of x-rays the image receptor receives. The number of pixels with each shade of gray is represented by the height of the graph. On the left side of the graph, lower opacity tissues such as lung allow more x-rays to pass through to the image receptor. These appear as darker areas in the radiograph. Higher opacity tissues such as spine attenuate more x-rays so fewer pass to the image receptor. These appear as lighter areas in the radiograph. To display all of the dark and light areas, the width of the histogram (red double-headed arrow) must fit within the dynamic range of the image receptor. Shades of gray that fall outside the dynamic range will not be visible in the radiograph.

material that appears darkest or blackest in a radiograph. Examples include lungs, any collection of gas inside the body, and the air outside the body.

Fat opacity materials attenuate more x-rays than gas and produce shades of gray that are lighter than gas but darker than fluid/soft tissue. Examples include fascial planes, omentum, falciform ligament, mediastinal fat, retroperitoneal fat, and lipomas.

Fluid/soft tissue opacity materials attenuate more x-rays than fat and produce lighter shades of gray than fat, but not as white as bone. The soft tissues and fluids in the body are the same relative opacity. Examples include muscles, organs, cartilage, blood, urine, and body cavity effusions.

Mineral opacity materials attenuate more x-rays than soft tissue/fluid and produce whiter areas in the radiograph. Examples include skeletal structures, many types of calculi,

Table 1.1 Relative opacities of various materials

Material	EAN	SG	Relative Opacity
Air	7.6	0.001	0.01
Foam rubber	5.0	0.15	0.8
Lung	7.4	0.32	2.4
Plastic ^a	5.8	0.05–1.06	0.03–6.1
Cellulose (wood, fabric)	7.0	0.20–1.50	1.4–10.5
Fat	6.0	0.91	5.5
Solid rubber	5.0	1.10	5.5
Ice	7.5	0.93	7.0
Water	7.5	1.00	7.5
Seawater	7.5	1.03	7.7
Muscle	7.6	1.04	7.8
Blood	7.6	1.06	7.9
Graphite	6.0	2.30	13.8
PVC	13.9	1.30	18.1
Diamond	6.0	3.50	19.5
Bone	13.8	1.85	25.5
Glass	10.3	2.50	25.8
Calcium	20.0	1.55	31.0
Aluminum	13.0	2.70	35.1
Concrete	17.0	2.30	39.0
Barium	53.0	3.50	122.0
Stainless steel	25.8	7.85	202.5
Zirconium	40.0	6.49	259.6
Iodine	56.0	4.93	276.0
Lead	82.0	11.34	930.0
Tungsten	74.0	19.25	1424.0
Gold	79.0	19.30	1524.7

The opacity of a material is its ability to attenuate x-rays. The more opaque the material, the fewer x-rays will pass through. The opacity of a material is affected by the sizes of its atoms (atomic number) and how tightly the atoms are packed together (specific gravity). The higher the atomic number (Z), the greater the opacity. As the atomic number increases, the likelihood of x-ray absorption grows exponentially (Z^3). In medical imaging, we frequently deal with materials that are made of a number of different elements, each with its own atomic number. Soft tissues in the body, for example, contain hydrogen, oxygen, carbon, nitrogen, and others. When dealing with composite materials, we use an approximate or effective atomic number (EAN). Specific gravity (SG) is a comparison of the number of atoms in a given volume of a material to the same volume of water. The larger the specific gravity, the more tightly the atoms are packed together. The higher the specific gravity, the greater the opacity because more atoms in the path of the x-rays and more x-rays are likely to be attenuated. Specific gravity has a greater effect on opacity than atomic number. Consider the opacities of air and muscle. Air attenuates far fewer x-rays than muscle, yet the effective atomic number for each is the same. They both are composites of hydrogen, oxygen, carbon, and nitrogen. However, the atoms in muscle are 1000 times more tightly packed together than in air. In this table, the relative opacity for various materials is calculated by multiplying $\text{EAN} \times \text{SG}$. These are approximate values for illustration purposes only.

^aPolystyrene and polyethylene.

and dystrophic mineralization. Although they are all mineral opacity, the relative opacity in different parts of bones differ, particularly in cortical and cancellous bone (Figure 1.49).

Metal opacity materials attenuate a large number of the x-rays used in medical imaging and produce the lightest, whitest areas in a radiograph. Examples include foreign objects, body markers, orthopedic implants, and positive contrast media such as barium and iodine compounds.

Categorizing a material as one of these five opacities does not mean it will produce the same shade of gray in every radiograph. It means that the material appears as one of these opacities relative to a known material. The actual gray shade a material will produce in a radiograph depends on its thickness, summation of other materials, and the intensity of the x-rays.

The **thicker** the material, the more atoms in the path of the x-rays and more x-rays are more likely to be attenuated. As thickness increases, opacity increases. A thick piece of material is more opaque than a thin piece of the same material (Figure 1.50). The thicker piece will attenuate more x-rays and appear lighter in the radiograph than the thinner piece, even though both are part of the same material with the exact same composition. The thicker material will be relatively more opaque than the thinner piece.

Thickness also can affect the relative opacities of two different materials with different compositions. Consider a thick piece of fat and a thin piece of bone (Figure 1.51). Fat is inherently less opaque than bone, but thick fat may produce the same shade of gray in a radiograph as thin bone because they both will attenuate the same number of x-rays and therefore appear as similar opacities in the radiograph.

Similar to alterations in thickness, **summation** can lead to an increase or decrease in the relative opacity of a material.

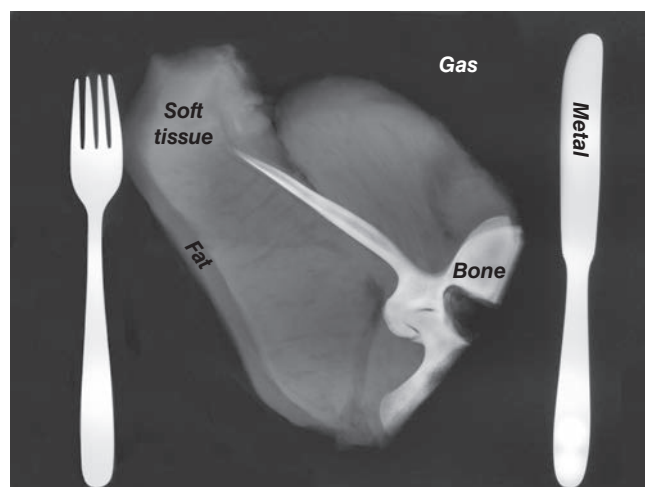


Figure 1.47 Five opacities. In this radiograph, all five relative opacities are represented by structures that are readily identified. Gas opacity in the surrounding air; fat opacity in the fascial planes and along the edge of the steak; soft tissue opacity in the steak muscle; mineral opacity in the bone, and metal opacity in the eating utensils.

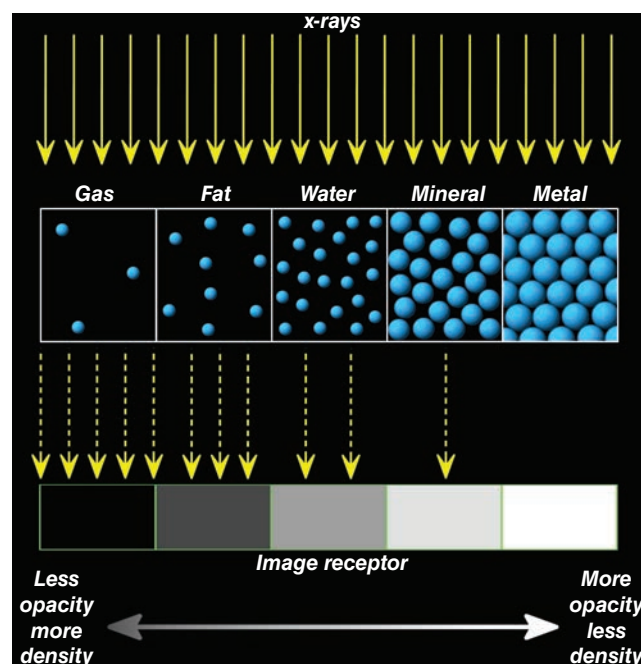


Figure 1.48 Five relative opacities. This illustration depicts equal thicknesses of gas, fat, water, mineral, and metal. The opacity of each is determined by the sizes of the atoms and how tightly they are packed together. The greater the opacity, the greater the x-ray attenuation and the fewer transmitted x-rays (dotted arrows). The fewer x-rays that reach the image receptor, the less the radiographic density. Opacity is inversely related to density.

Summation occurs when two or more materials in different planes overlap. The resulting opacity is the sum of all of the superimposed opacities (Figure 1.52). For example, soft tissue summated with gas will appear darker (less opaque), whereas soft tissue summated with soft tissue will appear lighter (more opaque). Because radiographs are two-dimensional images of three-dimensional structures, there is a lot of summation in most images. Summation can alter the opacity of a material to the point that it resembles the opacity of a different material in the radiograph. When this occurs, we differentiate the materials based on their location, shape or pattern in the radiograph (Figure 1.53).

So far, we have determined that the opacity of a material, and conversely the amount of radiographic density it will produce, is affected by its composition, thickness, and summation with other materials. The appearance of a material in the radiograph (shade of gray it will produce) is also affected by the **x-ray beam intensity**. The greater the number or energy of the x-rays, the more will pass through the material to reach the image receptor. A material will appear whiter in a radiograph made with low exposure and darker with high exposure. However, the intensity of the x-ray beam affects all materials in the field of view and all parts of the radiograph. Even though a specific material may appear lighter or darker based on x-ray beam intensity, its relative opacity compared to the other materials remains the same (Figure 1.54).

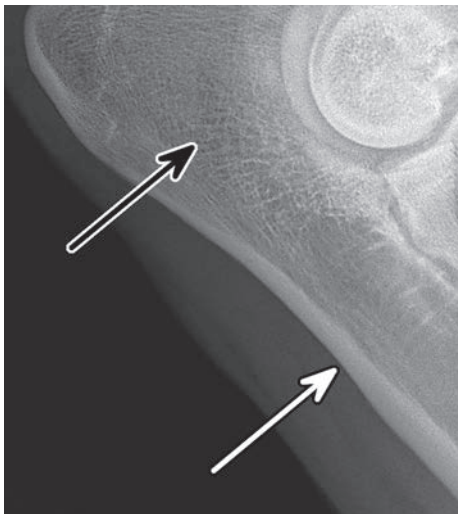


Figure 1.49 *Relative opacity.* Close up lateral view of a bone depicting the opacity of cortical bone (white arrow) compared to cancellous bone (black arrow). Both are mineral opacity, but cancellous is a mixture of bone and soft tissue, whereas cortical is mostly bone.

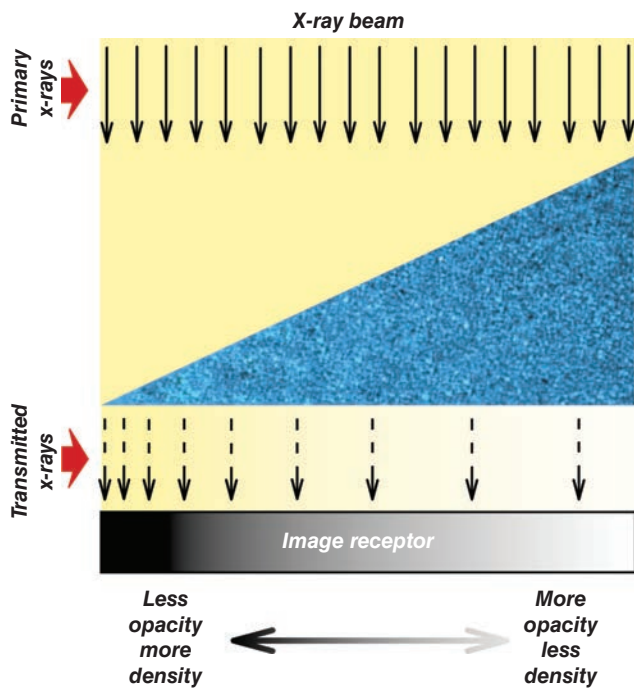


Figure 1.50 *Thickness.* An x-ray beam passing through a wedge-shaped piece of material will penetrate the thin part more easily than the thick part. The image receptor will receive more transmitted x-rays from the thin side than the thick side and will form more radiographic density on the thin side than on the thick side. The opacity of the thicker part is greater than the opacity of the thinner part, even though it is the same material with exactly the same composition throughout.

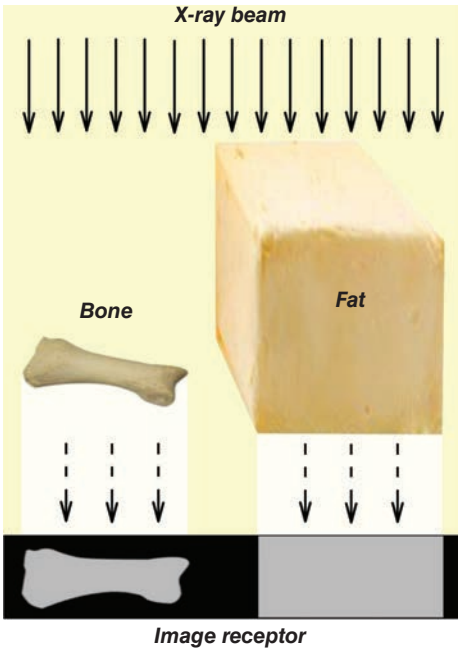


Figure 1.51 *Thickness and contrast.* This illustration depicts an x-ray beam passing through a thin piece of bone (A) and a thick piece of fat (B). The opacity of fat is inherently less than that of bone, but because it is very thick, the fat will attenuate an amount of x-rays equivalent to the thin bone. Both the bone and the fat produce the same radiographic density in the image below.

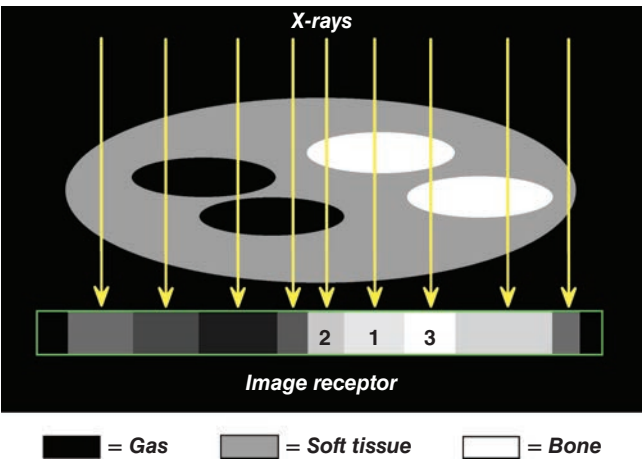


Figure 1.52 *Summation.* This illustration depicts x-rays passing through a structure that contains gas, soft tissue, and bone opacity materials. The opacity of each material and the amount of density it produces in the radiographic image varies due to summation. Bone, for example, appears as one shade of gray when summated with soft tissue (1), a darker shade when summated with gas (2), and a lighter shade when summated with bone (3).

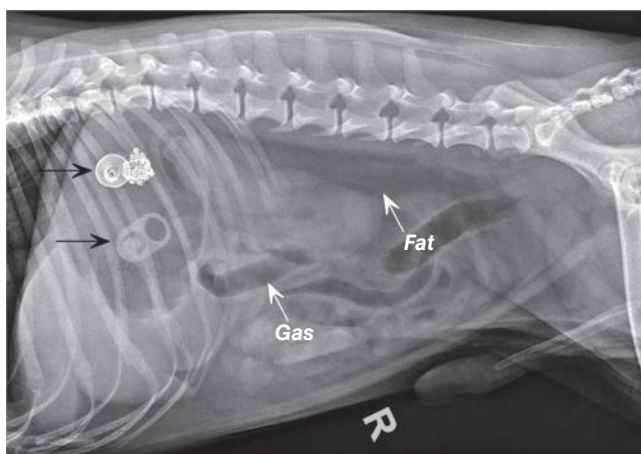


Figure 1.53 Relative opacity. Lateral view of a dog abdomen depicting similar radiographic densities of materials with different opacities. An area of bowel gas and an area of retroperitoneal fat produce the same shade of gray (white arrows), but they are readily distinguished based on their locations and adjacent structures. Gas is inherently less opaque than fat, but summation with soft tissue results in a radiographic density that is similar to fat. There are two metal opacity objects in the stomach (black arrows). Both are metal, but they differ in radiographic density. The density of the earring (closer to the spine) is clearly metal opacity. The density of the pop tab is similar to bone or soft tissue, but we recognize the shape and know it is metal.

The terms “radiolucent” and “radiopaque” have long been used to describe the opacity of materials, but they are actually misnomers. For example: mineral opacity calculi will appear white in urine and black in contrast medium (Figure 1.55). These calculi could be described as both “radiopaque” and “radiolucent”. Soft tissue opacity calculi will not be visible in urine (commonly called “radiolucent”), will appear black in contrast medium (also “radiolucent”), and will appear white when surrounded by gas (“radiopaque”). All materials are “radiopaque” to some degree and any material may appear “radiolucent” adjacent to a more opaque material. Labeling a radiographic finding as radiolucent or radiopaque does not convey nearly as much useful information as descriptions such as “mineral opacity”, “gas opacity”, “increased in opacity”, “decreased in opacity”, “more opaque than”, “less opaque than”, etc. For example, diamonds are less opaque than cubic zirconia (Figure 1.56).

Radiographic contrast

Contrast is the amount of difference between the dark areas and the light areas in a radiograph. It is the degree of change from one density (shade of gray) to the next. The **level of contrast** is determined by the number of shades of gray that are displayed. The more grays between black and white, the lower the contrast because there is less difference

between one shade and the next. A high contrast radiograph is mostly black and white with few shades of gray in between. High contrast sometimes is called *short scale* and low contrast sometimes is also called *long scale* (Figure 1.57).

The goal in making radiographs is to balance contrast with density so all areas of interest are visible. If contrast is too low, there will be no visible distinction between similar radiographic densities and we won’t be able to distinguish structures. If contrast is too high, some opacities will not be visible because the shade of gray needed to display them is not available (some adjacent shades of gray will be grouped together). The key factors that determine the level of contrast in a radiograph are the:

1. Latitude of the image receptor.
2. Energy of the x-rays (kVp).
3. Differences in opacities in the subject (patient).
4. Amount of fog.

Latitude is the range of x-ray intensities, from minimum to maximum, that an image receptor can receive and still form acceptable radiographic contrast. The latitude of a



Figure 1.54 Exposure and opacity. Three radiographs of the same dog foot made with increasing x-ray exposures. As exposure increases, the overall opacity of the structures decreases, but relative opacity remains constant. Notice that any point in one radiograph is a different shade of gray compared to the same point in another radiograph, but its *relative* opacity to other points in the foot remains the same. We recognize the different parts of the foot in each image because of the differences in relative opacity. It would be incorrect to describe any particular shade of gray as “bone” or “soft tissue” just because it is a certain level of radiographic density. We know it is soft tissue or bone because of its relative opacity to the tissue next to it.

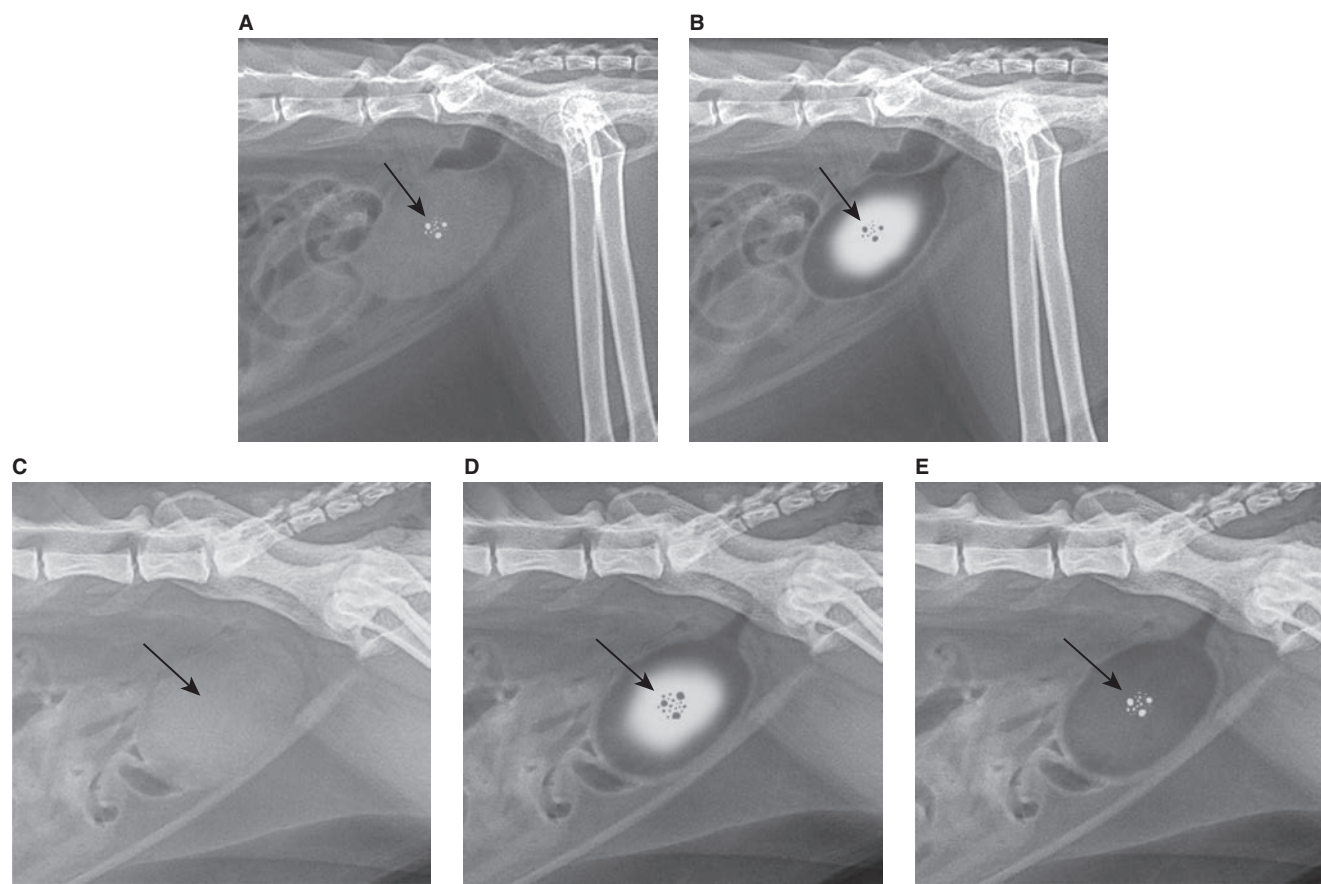


Figure 1.55 Radiolucent vs. radiopaque. Lateral views of a dog abdomen depicting mineral opacity and soft tissue opacity calculi in the urinary bladder. (A) Mineral opacity calculi in urine appear white or “radiopaque”. (B) The same mineral opacity calculi in contrast medium appear black or “radiolucent”. (C) Soft tissue opacity calculi in urine are not visible, but they commonly are called “radiolucent”. (D) The same soft tissue opacity calculi in a thin layer of positive contrast medium appear black or “radiolucent”. (E) The same soft tissue opacity calculi surrounded by gas appear white or “radiopaque”. As you can see, the words radiolucent and radiopaque are not stand-alone descriptors and convey less meaningful information than terms such as mineral opacity and soft tissue opacity.

receptor is represented by a characteristic curve (Figure 1.58). Latitude is inversely related to contrast: the wider the latitude, the less the contrast. Wide latitude receptors are able to form many shades of gray over a large range of x-ray intensities, which means they can display many different levels of opacity. Wide latitude is low contrast so there will be less distinction from one density to the next and less contrast between one opacity and the next. Narrow latitude receptors form fewer gray shades over a smaller range of x-ray intensities, which means fewer different levels of opacity can be displayed. Narrow latitude is high contrast so the radiograph will be more black and white. The latitude of the image receptor needs to be wide enough to display all of the structures of interest (levels of opacity) and narrow enough so there is adequate contrast to identify each structure in one radiograph. X-ray exposures that fall outside the latitude of an image receptor will result in little or no radiographic contrast.

In **film radiography**, the latitude of the image receptor is built-in and cannot be changed. The only way to increase or

decrease contrast is to switch to a film:screen system with a different emulsion formulation. Contrast in film radiography also is affected by processing. Over-development or under-development of the film can significantly and permanently reduce contrast.

In **digital radiography**, the latitude of the image receptors is very wide. Digital receptors are able to form acceptable contrast over a broad range of x-ray exposures.

The **higher the kVp, the lower the contrast**. Higher energy x-rays are more penetrating, which means there is less differential absorption between different materials. When there is less difference in x-ray attenuation, the different shades of gray that result from passing the x-ray beam through tissues of different opacities will be more similar in the radiograph and less distinguishable (Box 1.8). To quote a common expression used in medical radiography: “kVp kills contrast.”

Radiographic contrast can be altered to a certain degree by manipulating kVp and mAs. To increase contrast and maintain radiographic density, decrease the kVp by 16% and

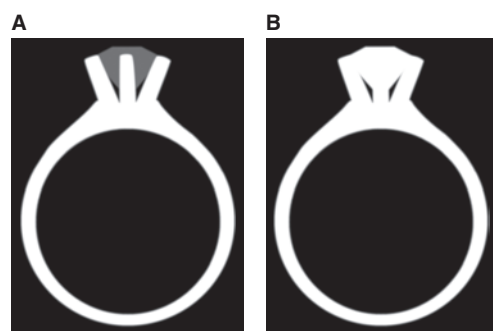


Figure 1.56 Diamond vs zirconium. The above images depict close-up radiographs of two similar engagement rings, one containing a diamond (A) and the other a cubic zirconium crystal (B). The relative opacity of the zirconium crystal is much greater than that of a diamond and it appears whiter in a radiograph. Diamonds are made of carbon which is essentially soft tissue opacity.

double the mAs. In most cases, you will need to repeat this adjustment more than once to make a visible difference in the radiograph (which may result in very large changes in kVp and mAs). To decrease contrast and maintain density, increase kVp 20% and cut the mAs in half. Again, repeat these adjustments as needed to achieve the desired effect. You may wonder why we say increase kVp by 20% and decrease it by 16%. If you do the math, you will see that we end up changing the kVp by nearly the same number each way. Some simply say change the kVp by 15% up or down.

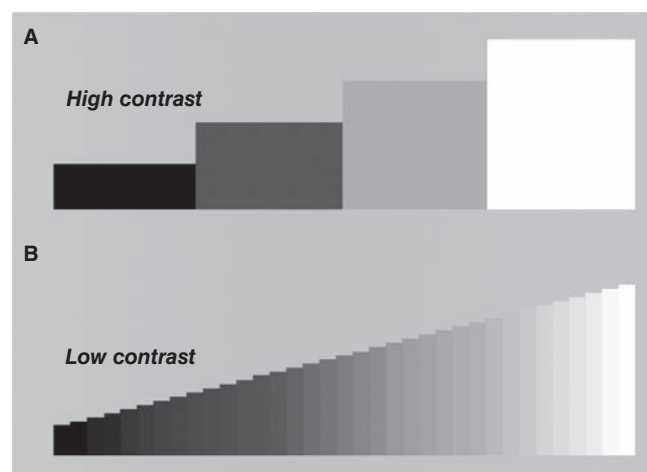


Figure 1.57 Contrast. (A) High contrast. There is a big difference moving from one shade of gray to the next and fewer steps between black and white. There are only a few gray shades available so only a few different levels of opacity can be displayed. (B) Low contrast. There is little distinction from one gray shade to the next and many steps between black and white. There are more shades of gray available to display more different levels of opacity. In this illustration, roughly 10 different shades of gray in the low contrast example are grouped into each shade of gray in the high contrast example. None of the 10 shades that are displayed in the low contrast image can be distinguished from each other in the high contrast image.

Always try to keep the exposure time as short as possible. In general, kVp above 70 results in relatively low radiographic contrast. Higher kVp techniques can be useful to help distinguish tissues with similar opacities because more shades of gray are available, such as imaging the lungs, abdomen, and thicker body parts. Lower kVp techniques (below 70) generally result in more radiographic contrast because there are fewer shades of gray. X-ray exposures near 70 kVp take advantage of the K-edges of bone and positive contrast agents to optimize their radiographic contrast (Figure 1.36).

Fog can significantly reduce radiographic contrast, as we discussed with scatter x-rays.

Displayed contrast

A tremendous advantage of digital radiography over film radiography is the ability to change contrast after the image has been acquired. In film radiography, contrast may be altered slightly by adjusting the intensity of the light that is transmitted through the film using variable intensity view-boxes and “hotlights.” Digital contrast, however, can be optimized using the numerous shades of gray available in each pixel. Post-processing in digital radiography allows us to evaluate both thin and thick structures in the same radiograph.

The amount of contrast a digital receptor can display is determined by the number of shades of gray available to each pixel, which is called **bit depth**. Bit depth (represented by “N”) is inversely related to contrast. As bit depth increases (N gets larger), the number of gray shades available to each pixel increases exponentially (2^N). For example, a pixel with a bit depth of 4 can display any of 2^4 or 16 different shades of gray. A pixel with bit depth of 8 can display any of 2^8 or 256 different shades (Figure 1.59). Recall that a bit is a digit in a binary number. Each shade of gray a pixel can display is represented by a binary number (Table 1.2). As bit depth increases, pixel numbers grow larger and larger and commonly are expressed in **bytes**. A byte is 8 bits which means a byte represents 2^8 or 256 shades of gray. A kilobyte (Kb) is

Box 1.8 Radiographic contrast

With high contrast, it is easier to see the differences, but there is less to see.

With low contrast, there are more things to see, but it is more difficult to see the differences.

2^{10} shades and a megabyte (Mb) is 2^{20} shades.

The computer automatically configures each pixel to display the best contrast by using a *look-up table*.

A **look-up table** (LUT) is a set of algorithms designed to optimize radiographic contrast for the specific body part

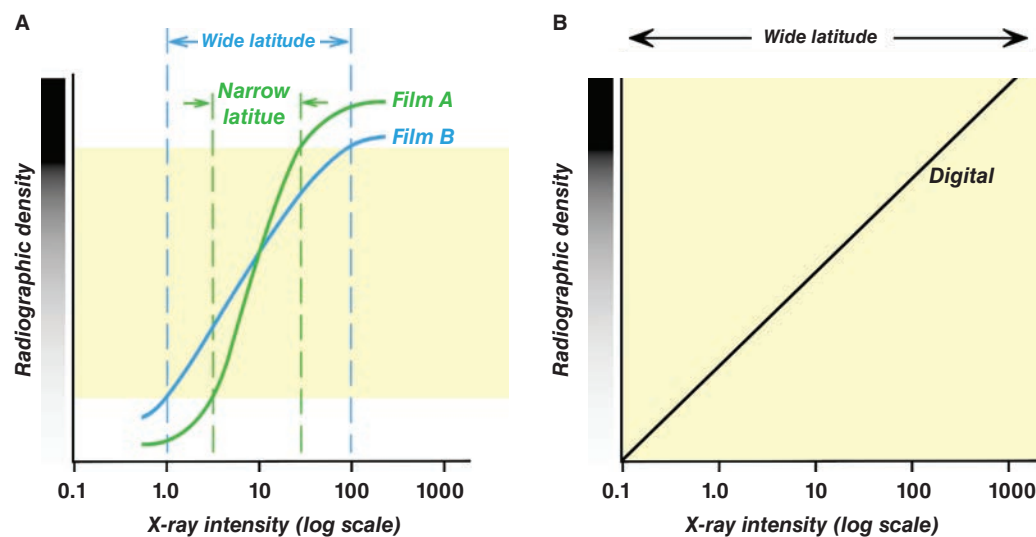


Figure 1.58 Latitude and contrast. These illustrations depict characteristic curves for film radiography (A) and digital radiography (B). Along the vertical axis (x-axis) of each graph are the acceptable levels of radiographic density (visible shades of gray). The horizontal axes (y-axes) is a logarithmic scale displaying the amount of x-rays the image receptor receives (the different x-ray intensities emerging from the patient). The yellow region across each graph depicts the shades of gray the image receptor can form. The slope of each curve represents the level of contrast the receptor can produce; the steeper the slope, the bigger the difference from one gray shade to the next. Latitude ranges from the minimum amount of x-rays the receptor needs to make any discernible shades of gray to the maximum amount of x-rays, above which no more contrast can be made. Any amount of x-rays below or above the latitude of a receptor will result in little or no contrast. **Film A** is narrow latitude. It can form contrast over the range of x-ray intensities between the two green dotted lines. The slope of the green curve is steep, which means small changes in the amount of x-rays will produce big changes in density and lots of contrast between shades of gray. **Film B** is wide latitude. It can form contrast over the range shown between the two blue dotted lines. The slope of the blue curve is not as steep as the green curve, which means more different levels of density can be formed over a larger range of x-ray intensities. There is less distinction from one gray shade to the next.

being imaged (e.g., thorax, abdomen, spine, extremity). The user selects the body part and the computer compares the histogram of the just acquired radiograph with its table of stored histograms. It “looks up” and selects the best histogram using specific pixel values of interest to properly display the radiograph. The LUT process permanently alters the raw data to improve contrast in the final radiograph. After computer processing, the user can use computer software to further alter the displayed contrast by *leveling* and *windowing* the image.

Leveling makes all of the pixels lighter or darker, effectively decreasing or increasing the radiographic density. **Windowing** allows the user to select a certain range of pixel values (shades of gray) from the numerous values available to increase or decrease contrast (Table 1.3). The range of selected values is called the **window width**. Any pixel values outside the window width will either be all white or all black. Window width is inversely related to contrast; the wider the window, the more shades of gray (less contrast). The narrower the window, the more black and white the image. The *window center* is the part of the radiograph where contrast will be focused. Leveling and windowing do not permanently alter the data; any changes made can be undone.

Radiographic detail

Detail is the sharpness of change from one radiographic density (one shade of gray) to the next. It determines how clear the edges appear when two structures are close together.

Radiographic detail is not the same as *opacity interface*. Detail refers to the quality of the radiographs whereas *opacity interface* relates to issues with the patient. Key factors that affect radiographic detail are listed below and in Box 1.9.

1. Motion.
2. Geometric unsharpness.
3. Noise.
4. Image receptor characteristics.

Motion during x-ray exposure is a major cause of poor detail in veterinary radiology. Motion may result from a struggling patient, an unstable x-ray tube, or inadvertent bumping of the image receptor. Motion leads to blurry margins in the radiograph. To minimize motion artifacts, use as short an exposure time as practical and appropriate patient restraint. Keep in mind that sedation may be less stressful to the patient than struggling with it.

Periodically inspect all equipment to make sure it is secure.

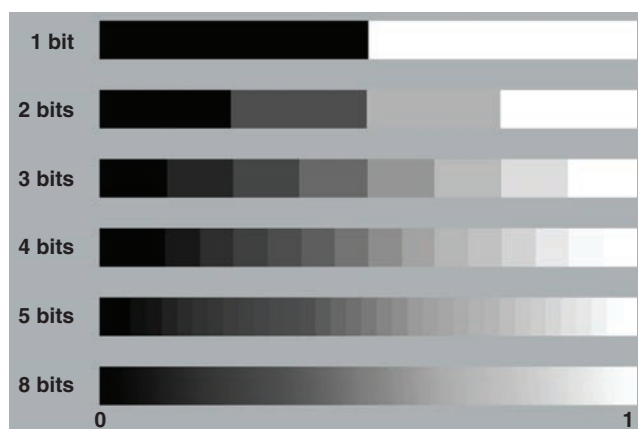


Figure 1.59 Bit depth. This illustration depicts the number of different shades of gray each pixel in an image is able to display as determined by its number of bits. An image made with 1-bit pixels would display the highest contrast because there are only 2^1 gray shades available: black or white. An image made with 4-bit pixels could display up to 2^4 or 16 different shades of gray. With 8 bit pixels, any of 2^8 or 256 shades of gray could be displayed.

Avoid hand-holding of portable x-ray tubes or image receptors. Not only does this lead to poor-quality radiographs, but it exposes the operator to more radiation than necessary.

Minimize geometric unsharpness to maximize detail. Use the small focal spot whenever practical

Table 1.2 Pixel numbers with 4-bit depth^a

Pixel Number				Pixel Value	Shade of Gray
0	0	0	0	0	
0	0	0	1	1	
0	0	1	0	2	
0	0	1	1	3	
0	1	0	0	4	
0	1	0	1	5	
0	1	1	0	6	
0	1	1	1	7	
1	0	0	0	8	
1	0	0	1	9	
1	0	1	0	10	
1	0	1	1	11	
1	1	0	0	12	
1	1	0	1	13	
1	1	1	0	14	
1	1	1	1	15	

^a A pixel with bit depth of 4 means it can display any of 16 different shades of gray. Each value of gray is represented by a 4-bit (4 digit) pixel number (a binary number).

(Figure 1.23) and the longest source-to-image distance (SID) that is feasible (Figures 1.27 and 1.28). Position the body part of interest as close to the image receptor as practical and align the x-ray beam, body part, and image receptor to minimize distortion (Figures 1.29–1.31).

Noise diminishes radiographic detail. Noise is random variations in density that produce a “mottled” or “grainy” pattern in the radiograph. The mottling does not represent anything in the patient and makes the edges of structures less distinct. The more noise, the less detail.

Noise is present in all radiographs. It ranges from non-distracting to severe enough to obscure all detail and render the radiograph useless. The mottled pattern created by noise tends to be more noticeable in underexposed and high-contrast radiographs. The cause of noise may be inherent in the image receptor itself or it may be due to the naturally uneven manner in which x-rays strike the image receptor. The latter is called *quantum noise* or *quantum mottle*.

Inherent noise or “background” noise occurs in all imaging systems and cannot be eliminated. In film radiography, it is associated with the sizes of the silver crystals and phosphors; the bigger the grains, the more noise. In digital radiography, inherent noise results from random electrical currents. These currents may be generated by thermal activity within the system or they may be caused by interference from other electronics in the area (e.g., fluorescent lights, nearby motors).

Quantum noise is caused by unequal numbers of x-rays striking the image receptor. In physics, a *quantum* refers to a discrete amount of energy. The term quantum commonly is used to describe an x-ray photon, thus the name “quantum noise.” The patchy pattern created by quantum noise is similar to that created by the first few drops of rain that fall on a paved walkway. In some areas, there are clusters of raindrops (x-rays) and in others there are only a few. The variation is due to the non-uniform nature in which the raindrops emerge from the raincloud or the x-rays emerge from the patient. Quantum noise becomes less noticeable as more and more x-rays strike the image receptor (more raindrops hit the pavement).

To assess the amount of detail an image receptor can provide, we look at the minimum distance between two lines that allow them to be distinguished as separate in the radiograph. This can be accomplished using a test pattern that contains progressively thinner and more numerous lines (Figure 1.60). The more lines that can be discriminated over a certain distance, the better the detail. The number of visible lines typically is expressed as line pairs per millimeter (lpmm) or line pairs per inch (lpi). The human eye can discriminate about 10–14lpmm. Film:screen systems can resolve about 7–10lpmm. Many current digital systems are capable of about 5–6lpmm, but some can resolve 10lpmm. Although detail in digital imaging systems typically is less than in film radiography, the wide dynamic range and latitude of digital

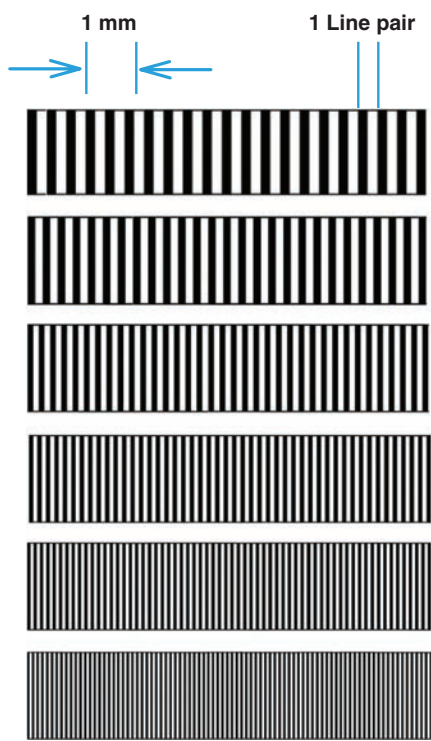


Figure 1.60 Line-pairs. Pictured here is a test pattern that can be used to evaluate the level of radiographic detail an image receptor can provide. The more line pairs that can be resolved over a distance of 1 mm, the better the detail. In this example, the top test pattern resolves 2 line pairs per mm. A line pair consists of a black line and a white line (the white lines are the spaces between the black lines). Radiographic detail is expressed as line pairs per millimeter or lpm.

receptors more than makes up for the difference. Digital detail likely will continue to improve **with advancing technology**.

Radiographic detail is also affected by **line spread** (Figure 1.61). Line spread refers to the lateral diffusion of light that occurs in image receptors that convert x-rays to visible light as an intermediate step (i.e., film:screen systems and indirect digital systems). The light emitted by a phosphor or scintillator spreads out to produce an area of light that is larger than the size of the x-ray itself. The more line spread, the less detail. Line spread does not occur in

Box 1.9 Maximize radiographic detail

- Use smallest focal spot that is practical.
- Make SID and SOD as long as feasible.
- Make OID as short as possible.
- Position central ray in middle of area of interest.
- Make area of interest parallel to image receptor.
- Use shortest exposure time practical.
- Utilize appropriate patient restraint.

direct digital radiography or in film radiography where no screens are used.

Displayed radiographic detail

The radiographic detail provided by a film:screen image receptor cannot be changed. Detail is limited by the sizes of the silver crystals in the film and the phosphors in the screen. The larger the grains, the more noise mottling and the less detail.

The amount of detail in a digital system is determined by **spatial resolution**. Spatial resolution is affected by pixel size, pixel pitch, and the signal-to-noise ratio. Pixel size is not a permanent, fixed, or defined number. There is no standard physical size to a pixel. Pixel size is determined by the size of the FOV and the size of the matrix used to display the digital image. A matrix is a rectangular array arranged in rows and columns (Figure 1.66). The FOV determines the overall length and width of the matrix. The size of the matrix (number of pixels it contains) is determined by its number of rows and columns. The smaller and more numerous the pixels, the better the detail (Box 1.10). Pixels generally are considered to be square-shaped therefore they are measured along one side. Pixel size and pixel pitch usually are expressed as length in microns (μ).

Pixel pitch is the distance between two adjacent pixels. Pitch is measured from the middle of one pixel to the middle of the next. The shorter the pixel pitch, the better the detail (Figure 1.62). In current digital radiography systems, the distance between pixels is about the same as the pixel size and the two generally are considered to be equivalent. Pixel pitch in current systems is around 100–150 μ . The human eye can distinguish about 25 μ . The width of a human hair is about 40 μ . Smaller pixel pitches are possible, but they would lead to much higher x-ray exposures.

Signal-to-noise ratio (SNR) affects the detail in digital radiographs. SNR is a comparison of the amount of useful information generated by a digital image receptor to the amount of non-useful information. In other words, it is the number of electronic signals produced by x-rays compared to the amount of noise. The higher the SNR, the better the detail. Low SNR most often is due to an insufficient amount of x-rays reaching the image receptor. Many digital imaging systems automatically try to compensate for low exposures by amplifying the few electronic signals. Unfortunately, amplification also increases noise.

The type of **monitor** used for viewing digital radiographs contributes greatly to visible detail. The monitor must be able to display the same size matrix and number of shades of gray that the digital receptor provides, otherwise some detail will be lost.

The Art of radiography. The quality of a radiograph is affected by four characteristics: **density**, **detail**, **distortion**, and **contrast**. There are multiple technical factors that

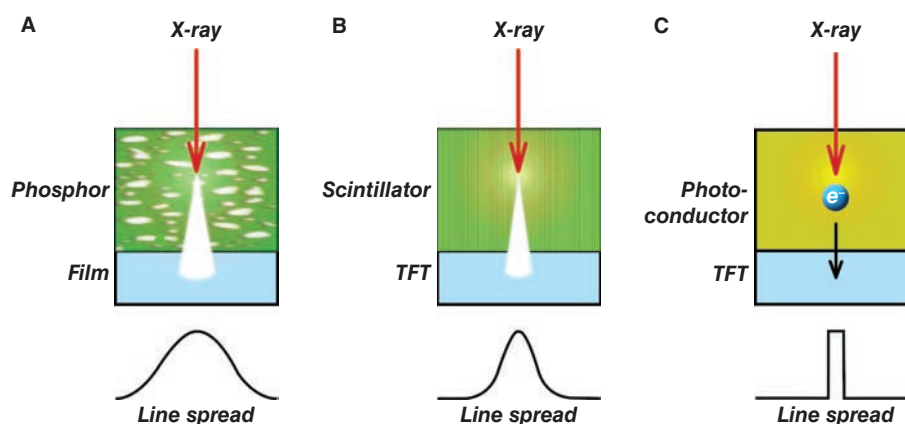


Figure 1.61 Line spread. These illustrations depict light being emitted by (A) a phosphor and (B) a scintillator in response to an x-ray interaction. The light spreads as it travels to the film or TFT array. Light spread is less with the scintillator due to the linear crystals that channel the light to reduce lateral diffusion. C. Image receptors that convert x-rays directly to electronic signals eliminate line spread.

influence these four characteristics, including mAs, kVp, focal spot size, SID, and others. In Table 1.4, we summarize the effects on radiographic quality caused by increasing or decreasing various factors. As you can see, altering a particular factor may increase one characteristic of radiographic quality

while decreasing another. Often there is a compromise. Knowing how to adjust different factors is part of the Art of radiography. The more skilled you are at image acquisition, the better the radiographic quality and the more diagnostic the images. The Art of radiography also includes proper

Box 1.10 Pixel size and radiographic detail

Increasing the matrix size (number of rows and columns) while maintaining the same FOV (length and width of the matrix) will reduce the sizes of the pixels and increase radiographic detail.

$$\text{Pixel size} = \frac{\text{FOV}}{\sqrt{\text{Matrix size}}}$$

For example: a FOV that is 20 × 20 mm will produce a radiograph that is 400 mm². If the radiograph is displayed with a resolution of 512 × 512 (a matrix size of 512 rows × 512 columns), there will be 262,144 pixels in the image (512 × 512). The size of each pixel is calculated as follows:

$$\frac{400}{\sqrt{(512 \times 512)}} = \frac{400}{512} = 0.78 \text{ mm} = 780 \mu$$

If we double the resolution, increasing the matrix size to 1024 × 1024, the number of pixels increases to 1,048,576, which means the size of each pixel must decrease by half to fit in the same FOV:

$$\frac{400}{\sqrt{(1024 \times 1024)}} = \frac{400}{1024} = 0.39 \text{ mm} = 390 \mu$$

If we double the resolution again, to 2048 × 2048, pixel size decreases by another 50%:

$$\frac{400}{\sqrt{(2048 \times 2048)}} = \frac{400}{2048} = 0.19 \text{ mm} = 190 \mu$$

Smaller and more numerous pixels leads to greater radiographic detail.

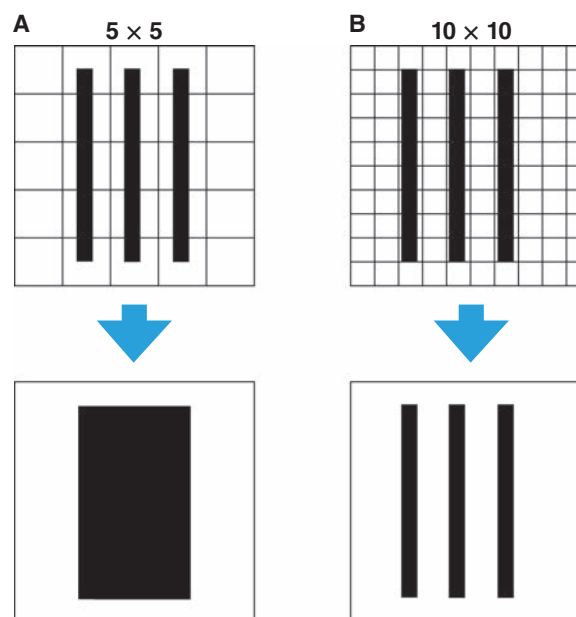
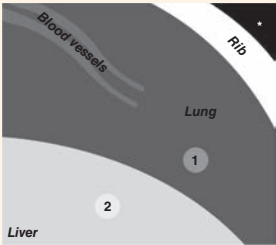


Figure 1.62 Pixel pitch. In this illustration, a 3-line test pattern is used to evaluate detail with two different matrix sizes. Each square in each matrix represents one pixel. To display the 3 lines as separate structures the sizes of the pixels must be smaller than the spaces between the lines. In matrix A, the resolution is 5 × 5 and the pixel pitch is larger than the spaces between the lines, which means there aren't enough pixels to distinguish the individual lines as separate. The test pattern is displayed as one thick line in the image below. In matrix B, the resolution is doubled to 10 × 10, which reduces both pixel size and pixel pitch. This allows a column of pixels between each line. The test pattern is displayed as three separate lines in the image below.

Table 1.3 Radiograph quality^a



This illustration depicts structures in a close-up VD view of the thorax. Included are soft tissue opacity blood vessels, liver, and lung nodules. Nodule 1 is in the lung field and nodule 2 is summated with the liver. The rib is mineral opacity. The lung is gas opacity and there is gas opacity outside the rib (*)



High contrast
High contrast narrow latitude The visible structures are mostly black and white. There are not enough shades of gray to display all of the different opacities in the thorax.



Poor detail
The margins of structures are indistinct, the edges appear hazy and ill-defined



Large pixels
Detail is less because the image is displayed with large pixels. The image is "pixilated," making the margins of structures appear irregular and indistinct



Good quality
Good detail the margins of the structures are distinct and well-defined
Good contrast all structures are displayed with acceptable density and contrast so they are visible in the image



Low contrast
Wide latitude There are many shades of gray between black and white, but little distinction from one shade to the next. Structures with different opacities are visible but not very distinct.



Noise
The background in the image appears mottled or grainy, which makes the margins of structures less distinct.



Small pixels
Detail is better because the image is displayed using smaller pixels. The matrix is larger, which means more pixels are available.

^aIn this table are several illustrations that depict the same area in a thoracic radiograph. The structures in this area are labelled in the first image (upper left corner). A good quality radiograph, with adequate density, contrast and detail, is depicted in the upper right corner. Poor quality radiographs due to unacceptable contrast, detail, noise, and pixel size are depicted in the remaining images.

Table 1.4 Table of compromise for radiography^a

Factor	Increase Decrease	Effect on Radiograph Quality			
		Density	Detail	Distortion	Contrast
mAs	↑	↑	—	—	—
	↓	↓	—	—	—
kVp	↑	↑	—	—	↓
	↓	↓	—	—	↑
SID with static OID	↑	↓	↑	↓	—
	↓	↑	↓	↑	—
OID with static SID	↑	↓	↓	↑	↑
	↓	↑	↓	↓	↓
Sensitivity of image receptor	↑	↑	↓	—	—
	↓	↓	↑	—	—
Focal spot size	↑	—	↓	↑	—
	↓	Limits exposure	↑	↓	—
Fog grid (reduce fog)	↑	↑	↓	—	↓
	↓	↓	↑	—	↑
Pixel size	↑	—	↓	—	—
	↓	—	↑	—	—
Noise	↑	—	↓	—	—
	↓	—	↑	—	—
Latitude	↑	—	—	—	↓
	↓	—	—	—	↑

^aThis table lists various technical factors that can affect radiographic quality. For example: increasing or decreasing mAs affects radiographic Density. Increasing or decreasing kVp affects both Density and Contrast. Increasing OID with static SID will decrease Density and increase Contrast due to an air gap (object will be further away from the image receptor) and will increase magnification distortion.

patient positioning and accurate interpretation of the images, which are discussed in the next chapter.

Technique chart

A chart of reliable x-ray exposures is essential to making consistent quality radiographs. A technique chart must be customized to the individual radiography system, whether digital or film. Technique charts commonly are **kVp variable**, which means the mAs is fixed according to the body part being imaged and the kVp is adjusted for thickness. A variable mAs chart can be made if the kVp settings are limited.

Prior to creating your technique chart, all equipment should be installed, properly calibrated, and in good working order. The x-ray machine should be connected to a dedicated line of electricity with a voltage compensator to minimize fluctuations in power. Set your **source-to-image distance**. The standard SID in most veterinary practices is 100 cm (40 in.). A shorter SID may be needed if the x-ray machine is low output, perhaps 75 cm (30 in.). Low output units such as portable x-ray machines may limit the flexibility of a technique chart.

The **grid** you plan to use should be installed or otherwise available. Grids are recommended when body part thickness exceeds 10–12 cm. When using a grid, the x-ray exposure needs to be increased from the non-grid technique. Some digital imaging systems incorporate software to compensate for scatter radiation, which may reduce the need for a grid.

With film radiography, make sure the **processing equipment**, whether manual or automatic, is installed and in optimal condition. The chemicals should be relatively fresh, well-mixed, and at the proper temperatures. However, brand new chemicals sometimes are “hot” and it may be preferable to make your technique chart about 2 weeks after changing chemistry, depending on how often it is changed.

The first step in the creation of your technique chart is to make a **mAs table**. Begin by listing all of the available mA settings across the top and all of the available exposure times down the left side. Then multiply each mA setting by each exposure time to determine all of the possible mAs settings (Table 1.5). This table will be used to choose the shortest exposure times available to help reduce motion artifacts.

The next step is to make some **test radiographs**. Select an animal that is typical of the type of patient seen at your clinic or hospital. In general, a medium-sized dog that is not overweight is a good candidate. Select a dog less than 20 kg (50 lb), perhaps with thickness of about 15 cm at its widest part. Make three lateral views of the abdomen. We image the abdomen because it contains many soft tissue opacity structures.

Make one abdominal radiograph using 5 mAs, another with 10 mAs, and the third with 20 mAs. Use your mAs chart to select the shortest exposure time for each. To select a kVp, measure the width of animal’s abdomen in centimeters and

Table 1.5 mAs table ^a					
mA	50 mA	100 mA	200 mA	300 mA	
Time (seconds)					
1/60 (0.016)	0.83	1.67	3.33	5.00	
1/30 (0.033)	1.67	3.33	5.00	10.0	
1/20 (0.050)	2.50	5.00	10.0	15.0	
1/15 (0.066)	3.33	6.67	13.3	20.0	
1/10 (0.100)	5.00	10.0	20.0	30.0	
2/10 (0.200)	10.0	20.0	40.0	60.0	
3/10 (0.300)	15.0	30.0	60.0	90.0	
4/10 (0.400)	20.0	40.0	80.0	120	
1/2 (0.500)	25.0	50.0	100	150	
3/5 (0.600)	30.0	60.0	120	180	
4/5 (0.800)	40.0	80.0	160	240	
1 (1.000)	50.0	100	200	300	

^aFor this particular x-ray machine, the available mA settings are listed across the top of the table and the available exposure times are listed down the left side. Times are displayed as both fractions and decimals. Multiplying each mA by each time gives all of the possible mAs settings for this particular x-ray machine.

the SID in inches. The kVp is calculated from these two factors. It should be twice the abdomen thickness plus the SID. For example, if the abdomen is 15 cm and the SID is 40 in., the kVp would be 70; (2 × 15) + 40 = 70. Use this kVp for all three radiographs.

After all three radiographs have been made, select the one with the best radiographic density (Figure 1.63). The mAs used to make this radiograph will be the **base mAs** for the technique chart. Be sure to dry completely all hand-processed film radiographs before reviewing them.

If all of the radiographs are underexposed (too light), either the kVp or the mAs needs to be increased. If there is too much contrast in the radiographs (too black and white), increase the kVp. If the contrast is acceptable, increase the mAs. If increasing the mAs makes the exposure time exceed 1/30 second, increase the kVp instead. If all of the test radiographs are overexposed (too dark), cut the exposure time in half to reduce the mAs by 50%.

Variable kVp chart

The third step in creating your technique chart is to determine the **variable kVp** settings for different thicknesses and body parts. Let’s begin with abdomen. Suppose from our test radiographs we determined that the exposure needed to make an acceptable lateral view of a 15 cm thick abdomen was 10 mAs and 70 kVp. We can then determine the exposure needed for every other thickness of abdomen. For each 1 cm decrease in abdomen thickness, we subtract 2 kVp. For each 1 cm increase in thickness, we add 2 kVp. We can do this up to 80 kVp. Above 80 kVp, we add 3 kVp for each 1 cm increase

in thickness until we get to 100 kVp. Over 100 kVp, we add 4 kVp for each 1 cm increase in thickness.

For **thoracic** radiography we use the same kVp settings as for abdomen, but decrease the mAs by half. The thoracic cavity is mostly air-filled and generally requires less exposure. Lower mAs means shorter exposure times which is important to minimize cardiac and respiratory motion blur. Make a lateral thoracic radiograph of your test animal to confirm this thoracic technique is satisfactory. If needed, increase or decrease kVp to darken or lighten the test image. Then extrapolate the rest of the kVp values for the other thicknesses (Table 1.6).

For **orthopedic** imaging we use the same kVp settings as for abdomen, but double the mAs. Make a lateral view of the lumbar spine of your same test animal to confirm this

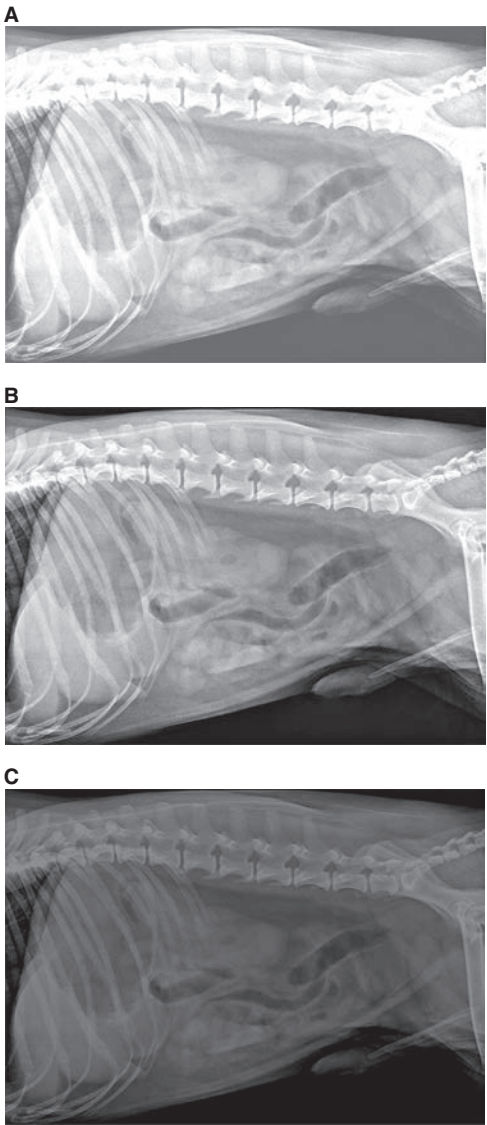


Figure 1.63 Test radiographs. Three lateral views of a dog’s abdomen depicting (A) too little radiographic density due to underexposure, (B) acceptable density, and (C) too much density due to overexposure.

Table 1.6 Variable kVp chart.^a

Thickness	Thorax		Abdomen		Skeleton	
cm	kVp	mAs	kVp	mAs	kVp	mAs
1	50	2.5	42	5.0	36	5
2	52	2.5	44	5.0	38	5
3	54	2.5	46	5.0	40	5
4	56	2.5	48	5.0	42	5
5	58	2.5	50	5.0	44	5
6	60	2.5	52	5.0	46	5
7	62	2.5	54	5.0	48	5
8	64	2.5	56	5.0	50	5
9	66	2.5	58	5.0	52	5
10	68	2.5	60	5.0	54	5
Grid						
11	70	5.0	62	10	56	10
12	72	5.0	64	10	58	10
13	74	5.0	66	10	60	10
14	76	5.0	68	10	62	10
15	78	5.0	70	10	64	10
16	80	5.0	72	10	66	10
17	83	5.0	74	10	68	10
18	86	5.0	76	10	70	10
19	89	5.0	78	10	72	10
20	92	5.0	80	10	74	10
21	95	5.0	83	10	76	10
22	98	5.0	86	10	78	10
23	101	5.0	89	10	80	10
24	105	5.0	92	10	67	20
25	109	5.0	95	10	69	20
26	113	5.0	98	10	71	20
27	117	5.0	86	20	73	20
28	121	5.0	89	20	75	20
29	125	5.0	92	20	77	20
30	110	10	95	20	79	20

^aThis chart was made for a particular x-ray machine. It displays the mAs and kVp settings needed to image a wide range of body part thicknesses. The test radiographs were made using a 15 cm thick patient. A grid is used when body part thickness exceeds 10 cm. Notice the mAs is doubled when using the grid. In this example chart, the kVp used for thorax was increased 20% over the abdomen technique because the test radiograph for thorax was underexposed. The kVp for musculoskeletal radiographs was decreased 16% from abdomen technique because the test radiograph of the spine was overexposed.

technique is acceptable. Then extrapolate the rest of the kVp values. Double the mAs again if the radiograph needs to be darker or decrease kVp by 16% if the image needs to be lighter. Try to keep the kVp below 80 to take advantage of the K-edge of bone and enhance contrast (Table 1.6).

Adjustments to the technique chart

Patients presented for veterinary radiography vary significantly in size, shape and conformation. In addition, the reasons that bring them into the clinic or hospital often necessitate adjustments to the exposure technique (e.g., fluid or gas in a body cavity). The mAs or kVp suggested by the technique chart may need to be altered for a particular patient, particularly with film radiography (Boxes 1.11 and 1.12). The choice of which factor to adjust often is based on experience and equipment limitations.

Changes to mAs typically are made to increase or decrease radiographic density. If you don't like a radiograph that is well positioned with good detail and is not fogged, think density first. If the radiograph is too light, double the mAs. If it is too dark, cut the mAs in half. Altering the mAs by less than a factor of 2 probably won't make a visible change in the radiograph. Changing the mAs can be repeated as needed to achieve the desired density. If you correct the density and the radiograph still is not acceptable, then consider changing the contrast.

Adjustments to kVp alter the penetrating abilities of the x-rays and affect the radiographic contrast. The minimal change to kVp to produce a visible difference in the radiograph is either a 20% increase or a 16% decrease. These adjustments also can be repeated as needed to achieve the desired result. To increase the contrast in a radiograph, but maintain the same radiographic density, decrease the kVp 16% and double the mAs. Do this at least twice. To decrease contrast and maintain density, increase kVp 20% and cut mAs in half. Again, do this at least twice.

Box 1.11 Conditions that typically require higher exposures

1. Obesity.
2. Pleural or peritoneal effusion.
3. Positive contrast studies.
4. Overlying bandaging or cast material.

When using a low-power x-ray machine, a good rule of thumb is to increase kVp when the radiograph is too light (if possible), and decrease the mAs when it is too dark. This allows you to use the shortest exposure time possible.

Box 1.12 Conditions that typically require lower exposures

1. Emaciation.
2. Immature animals.
3. Pneumothorax.
4. Negative contrast studies.
5. Compression radiography.

Radiograph storage and distribution

Radiographs are part of the medical record and must be safeguarded. Film radiographs must be physically stored in a safe environment where they are protected from theft, abuse, chemicals, and extremes in temperature and humidity for the legal life of the records. Digital radiographs must be stored in computer files with reliable backup copies both onsite and offsite.

Federal laws in the United States mandate that all radiographs be securely maintained and readily accessible both for the short term and the long term. All states in the US have enacted laws that govern the retention of medical records. In general, veterinarians are required to keep records at least 5 years past their last contact with the patient. Consult your local governmental guidelines and malpractice statutes of limitations for specific requirements.

Radiographs usually are regarded as the property of the hospital or clinic. However, in the US most state laws guarantee that your clients have access to them. It is a good idea to have a written policy regarding the release of any radiographs. The policy should state whether the originals will be loaned out or if copies will be made and how the radiographs will be tracked to ensure their safe return. Signed documents are recommended.

Most veterinary practice laws require specific medico-legal documentation be attached to all radiographs (Box 1.13). This information must be permanently embedded in the emulsion of film radiographs using photo flashing equipment, lead markers, or lead-based writing tape (Figure 1.64). Body markers also should be visible (e.g., Right and Left markers). In some practices, each technologist has his or her own R and L body markers with their initials on them. Digital radiographs must be saved in a format that securely attaches the required information to the image file. This most often is done using DICOM.

DICOM is an acronym that stands for **D**igital **I**maging and **C**ommunication **I**n **M**edicine. It is a worldwide standardized method for storing and transmitting digital radiographs.

DICOM is not a type of image, rather it is a type of file format. With DICOM, various types of data can be permanently attached to the radiographic image, including patient information, the type of digital equipment that was used to make the radiograph, pixel information, facts about transmission, image compression, and more. The data is protected and cannot be altered without destroying the file. A major advantage with DICOM is its worldwide acceptance. Radiographs created on one type of digital imaging system can be viewed on a different system without loss of quality. Radiographs saved in DICOM format typically are stored, retrieved, transmitted, and viewed using a PACS.

Box 1.13 Required information on all radiographs

- Name of veterinarian or hospital/clinic.
- Name of animal.
- Name of owner.
- Date radiograph was made.
- Initials of persons making the radiograph.

PACS is an acronym that stands for **P**icture **A**rchiving and **C**ommunications **S**ystem. It refers to a network of hardware and software designed to securely store and transmit radiographs (Figure 1.65). Most PACS can handle multiple imaging modalities in addition to radiography, such as ultrasoundography, computed tomography and magnetic resonance imaging. The PACS provides individual points of access to the images and associated data and to both short term and long term archives. Another name for a PACS is DIMS which stands for Digital Image Management System.

Within a PACS network is a group of two or more computers that are connected to each other. Linked computers that are close together (e.g., in the same building) is called a *local area network* or LAN. Computers that are further apart and linked by telephone lines, cables, or radio waves across

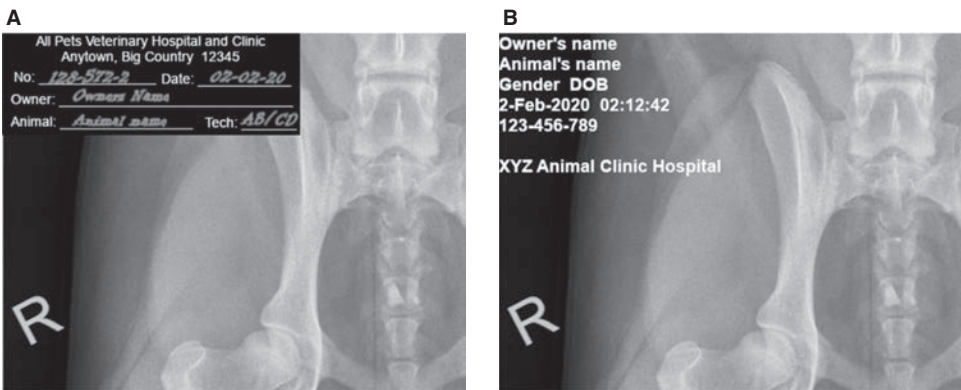


Figure 1.64 Radiograph identification. **A.** Film: photo flash. **B.** Digital. Identifying information should be clearly visible and permanent in each radiograph.

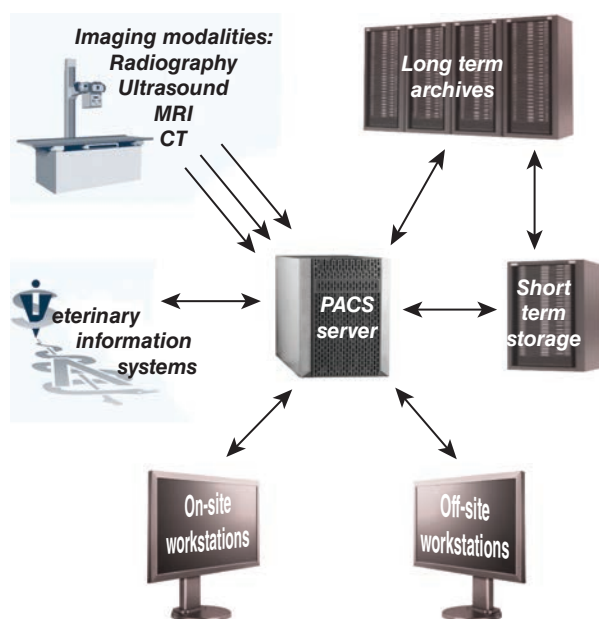


Figure 1.65 PACS. This illustration depicts the various components of a typical PACS. Each component is linked to a central server in a geometric arrangement called *star topology*. Other types of topology are also possible. Imaging modalities send radiographs and other data to the server, which in turn communicates with the hospital information system, onsite and offsite workstations, and both long-term and short-term storage archives.

a city, country, or the world is called a *wide area network* or WAN.

The **server** computer in a PACS manages the resources, runs the operations, and stores the data. The **client** computers request information from one or more servers and are part of the *workstations*. Workstations are used to display, process, and review radiographs and to link them with patient information and other data. A workstation may either be onsite, physically located in the hospital or clinic, or offsite, set up in a location that is remote from the medical facility. Many workstations are equipped with a diagnostic grade monitor, PACS viewing software, and a teleradiology service.

A great benefit of a PACS is the ease of transmitting digital radiographs, medical reports, and patient information among its various components and to the hospital medical systems (e.g., Hospital Information System, Radiology Information System, Electronic Medical Records). There are many complex components that need to operate well together for a PACS to function properly. Software and network connections must be compatible with multiple users, various storage sites, and the practice management system. It is essential to have good technical support and to train your technicians and staff so they have a basic understanding of the PACS. A well-managed PACS is mostly invisible, operating behind the scenes to minimize the need for repeated input of data,

providing ready access to the radiographs, speeding up the turnaround times for radiographic studies, and improving patient care. A PACS is only as good as its weakest link.

Digital radiographs contain vast amounts of information, which makes for very large file sizes (Box 1.14 and Figure 1.66). The larger the file, the more computer memory is needed and the longer it takes to process and transmit the data. In addition, the files must be backed up, which doubles or triples the amount of storage space required. To make file sizes more manageable they usually are **compressed**.

Various computer algorithms are used to compress digital radiographs. Some of these are proprietary and therefore not compatible with other applications. It is important to determine whether your PACS is using a DICOM compliant algorithm. If not, the type of compression you are using may lock you in with your current PACS vendor. There are two basic types of compression: *lossless* and *lossy*.

Lossless compression is reversible, which means the original radiograph can be reconstructed from the compressed data without any loss of image quality. The amount of compression or the *compression ratio* with the lossless method is about 2 : 1.

Lossy compression is not reversible. The original radiograph cannot be completely reconstructed from the compressed file, so there will be some loss of image quality. The clinical significance of the lost quality varies with the type of algorithm used. The compression ratio with the lossy method is greater than with the lossless method and can be as high as 20 : 1.

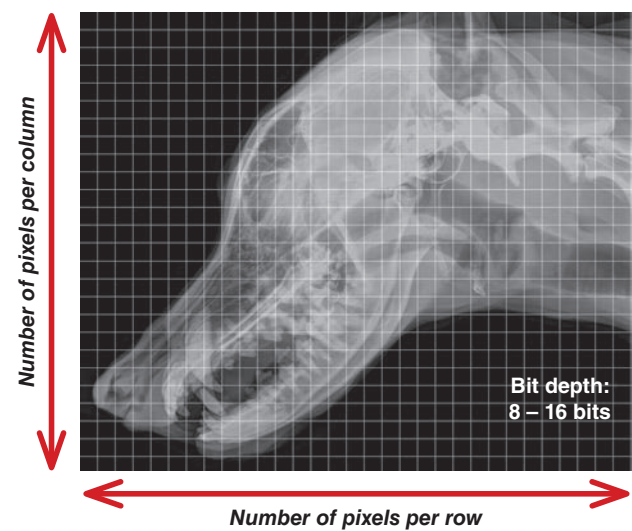
As mentioned earlier, back-up copies of digital radiographs must be safely stored both onsite and offsite. The back-ups need to be readily accessible at all times, either through a direct link to your PACS or via a network. **Onsite storage** requires something in addition to just backing up to the computer hard drive. **Reliable redundancy is crucial.** Backup copies must be dependable so they can be used to replace your data should it be lost for whatever reason. Many options are available to make spare copies of your data onsite, including extra computer hard drives, portable

Box 1.14 File size of a digital radiograph

The file size of a digital radiograph is calculated by multiplying the total number of pixels in the image by the pixel bit depth. For example: a radiograph made on a 12-bit digital system with a 1024×1024 resolution results in a file size of 1.5 Mb.

The total number of pixels in the image is

$$\begin{aligned} 1024 \times 1024 &= 1,048,576 \text{ pixels} \\ 1,048,576 \times 12\text{-bit} &= 12,582,912 \text{ bits} \\ 12,582,912 \text{ bits} \div 8 &= 1,572,864 \text{ bytes} \\ 1,572,864 \text{ bytes} \div 1024 &= 1536 \text{ Kb} \\ 1536 \text{ Kb} \div 1024 &= 1.5 \text{ MB} \end{aligned}$$



Matric size (R × C)	Number of Pixels	Radiograph size in Mb
512 × 512	262,144	0.25 – 0.50
1024 × 1024	1,048,576	1.0 – 2.0
2048 × 2048	4,194,304	4.0 – 8.0

Figure 1.66 File size of digital radiographs. A matrix is superimposed on a lateral view of a dog’s head. The file size of the radiograph is determined by multiplying the total number of pixels in the matrix by the bit depth of the digital imaging system. In this example, three different matrix sizes are listed (512, 1024, and 2048). The total number of pixels in each matrix is determined by multiplying the number of pixels in a row (R) by the number of pixels in a column (C). The file sizes of radiographs made using digital systems ranging from 8–16 bit depth are listed in megabytes.

external disc drives, optical drives, flash storage, tape, and others. A redundant array of independent disks (RAID) can be configured to mirror the data on the computer so that if one hard drive fails the data remains safe on the others. Portable media can be used off-line while not connected to the network and also can be transferred off-site.

Offsite storage means the back-up data is moved away from the medical facility. This commonly includes sending it to one or more remote servers, known as *cloud* storage. **Cloud storage** may provide some advantages, including technical support and a large amount of storage space, the latter usually is expandable to meet future needs. Radiographs stored in the cloud often are easily shared with clients, colleagues, radiologists, and other specialists. When selecting a cloud storage platform, be sure your data is stored in a vendor-neutral and DICOM compliant format.

Radiographs are stored for different lengths of time in different locations. Initially, they are sent to the PACS server, where they are stored for a few months up to 2 years, depending on

whether the facility is in-patient or out-patient. A copy typically remains in the digital system where the radiographs were made for a few days. The server sends copies to be archived in long-term storage. In the archives, the radiographs that are not being actively used are stored for the legal life of the study.

Radiation safety

Exposure to x-radiation can damage or destroy living cells and increase the risk of developing cancer. Although the health risk is relatively small, there is no “safe dose” of radiation. Any amount of exposure can cause damage (Box 1.15). The level of health risk depends on the dose of radiation received, the time over which the dose was received, and the body parts exposed. The higher the dose and the shorter the time to receive that dose, the greater the risk. Some body parts are more sensitive to radiation damage than others (Table 1.7).

We are exposed to radiation every day, from both natural and man-made sources (Figure 1.67). **Natural radiation** is ever-present, coming from outer space, the earth, and even our own bodies. Cosmic radiation originates in the sun and stars and continuously impacts the earth. It is more intense at higher elevations. Radiation from the earth originates in the low levels of radon, uranium, radium, and others that are ubiquitous in the soil, water, and vegetation. We frequently ingest this type of radiation with our food and water and also inhale it, the latter primarily as radon. Radiation within our bodies comes from isotopes of carbon and potassium present since birth. The average dose of natural radiation to persons living in the United States is about 0.3 rem/year.

Man-made radiation mostly comes from medical procedures such as diagnostic imaging and radiation therapy. It also comes from consumer products such as televisions, smoke detectors, building materials, airport x-ray security systems, and to a lesser extent from the fallout of nuclear weapons, nuclear accidents, and the fuel used in nuclear reactors. The average dose of man-made radiation to persons living in the United States is about 0.3 rem/year.

Box 1.15 Radiation safety

- Any amount of x-ray exposure can damage cells.
- There is no threshold dose of x-rays below which it is certain that an adverse effect cannot occur.
- The effects of radiation are cumulative and may not become apparent until later in life.
- Any exposure to x-rays should be As Low As Reasonably Achievable (**ALARA**).

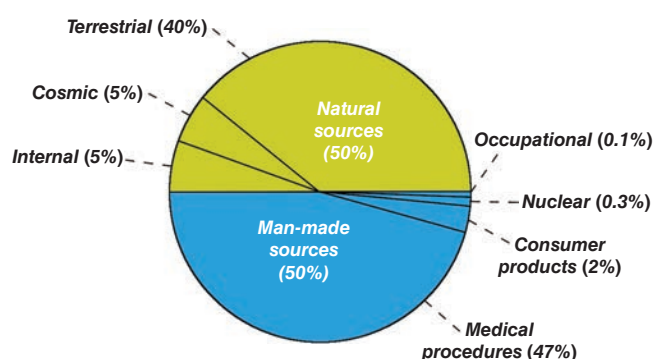


Figure 1.67 Sources of radiation exposure. The average dose of radiation received by persons living in the United States is around 0.6 rem/year. About half of this dose comes from natural sources and half from man-made sources (this illustration is modified from NCRP Report No. 160.) The dose can be much higher for persons receiving radiation therapy.

Biologic effects of radiation

Radiation is one of the most thoroughly investigated causes of disease. More is known about the effects of x-rays on cells and organ systems than is known about many other health risks. The damage caused by x-rays is due to ionization. When electrons are lost chemical bonds can break and molecules can fall apart. Ionization can directly or indirectly injure or kill cells.

Direct cellular damage occurs when vital molecules in the cell wall or in the DNA are destroyed by ionization. Indirect damage results from the free radicals that are generated by the ionization of water. Most living cells are nearly 80% water. Free radicals are highly reactive molecules that can severely alter cell functions. The susceptibility of a cell to the damaging effects of x-rays is known as *radiosensitivity*.

Radiosensitivity is relative (Table 1.7). Some cells are more sensitive to radiation than others. The most radiosensitive cells are those with high rates of division, high rates of metabolism, and immature or non-specialized cells. When a cell is damaged by ionizing radiation there are three possible outcomes:

1. The damage may be completely repaired and the cell returns to normal function.
2. The damage may kill the cell or make it unable to reproduce, either of which results in complete loss of cell function. This is known as **deterministic** damage and becomes clinically evident when enough cells lose function and cause the organ to fail.
3. The damage may be poorly repaired, resulting in mutation of the cell and possible future malignancy. This is known as **stochastic** damage and clinical signs are less predictable. The severity of a mutation is not dependent on the dose of radiation received; any dose can lead to the development of cancer.

Measuring radiation exposure

Several different measurements are used to describe the effects of radiation on living tissue. The *exposure dose* is the amount of radiation energy delivered to an area. The *absorbed dose* is the amount of energy deposited in tissue. The *effective dose* is the amount of biological damage that was caused by the absorbed dose. When dealing with x-rays, each of these dose measurements is equivalent. They are all affected by the x-ray beam intensity, the distance from the source of x-rays, and time during which the tissue is irradiated. We currently are measuring radiation doses using units from two different systems (Table 1.8): the *British system* with its "conventional units" and the *System International* (SI).

The **exposure dose** is a measurement of the amount of ionizations a given intensity of x-rays can cause in air. The conventional unit of measure is the Roentgen (R). The SI unit is coulomb/kilogram (C/kg). Specifically, 1 R is the amount of x-ray energy needed to create 1.6×10^{12} ion pairs/g in dry air at 0°C. This amount of energy is very small in everyday life, but at the cellular level it is quite large, able to cause thousands of trillions of ionizations. The conversion

Table 1.7 Relative radiosensitivity of various tissues in the body

Level of Radiosensitivity	Tissue
High	• Lymphoid organs, thymus • Bone marrow, blood • Spleen • Testes, ovaries • Small intestine
Fairly high	• Skin, hair follicles • Cornea • Oral cavity, esophagus • Rectum • Urinary bladder, ureters • Vagina, uterine cervix
Moderate	• Optic lens • Stomach • Growing cartilage and bone • Fine vasculature
Fairly low	• Mature cartilage and bones • Salivary glands • Lungs, kidneys, liver • Pancreas • Thyroid, adrenal, pituitary glands • connective tissue
Low	• Muscle, • brain, spinal cord, nerves

factor to change conventional units to SI units is $1\text{ R} = 2.58 \times 10^{-4}\text{ C/kg}$.

The **absorbed dose** relates the exposure dose to the amount of damage the x-rays could cause in living tissue. The conventional unit of measure is the rad, which is short for **radiation absorbed dose**. The SI unit is the gray (Gy). Specifically, $1\text{ rad} = 100\text{ ergs of energy/g of material}$ and $1\text{ Gy} = 1\text{ J of energy/kg}$. The conversion factor to change conventional units to SI units is $100\text{ rad} = 1\text{ Gy}$.

The **effective dose** is a measure of the biological effects of different types of radiation. It is sometimes called the **dose equivalent**. Some forms of radiation are more damaging than others and the effective dose can be greater than the absorbed dose. However, with x-rays the effective dose and the absorbed dose are equivalent. The conventional unit of measure is the rem, which is short for **roentgen equivalent in man**. The SI unit is the sievert (Sv). The conversion factor for changing conventional units to SI is $1\text{ rem} = 0.01\text{ Sv}$.

In the United States, the Nuclear Regulatory Commission (NRC) has set limits for the **maximum allowable radiation doses** a person can receive in a year (Table 1.9). These limits are in addition to any exposure received from natural and other man-made sources of radiation, which is about 0.6rem/year. People who work with radiation and are occupationally exposed are allowed up to 5rem (0.05Sv) per year. This includes medical personnel. Occupational workers who are pregnant or under the age of 18 are allowed only 10% of the annual dose limit or 0.5rem (0.005Sv) per year. Non-medical personnel and the general public are allowed no more than 2% of the whole body occupational dose in a year (0.1rem or 0.001Sv). In a veterinary practice, no one under the age of 18 or pregnant should be exposed to x-rays.

Table 1.8 Units of radiation measurement

Measurement	Conventional Unit	SI Unit
Exposure dose	Roentgen (R)	Coulomb/kg (C/kg)
Absorbed dose	rad (r)	Gray (Gy)
Effective dose	rem	Sievert (Sv)
Radioactivity	Curie (Ci)	Becquerel (Bq)
Conversions		
Exposure dose	$1\text{ C/kg} = 3876\text{ R}$, $1\text{ R} = 258\mu\text{C/kg}$	
Absorbed dose	$1\text{ Gy} = 100\text{ rad}$	
Effective dose	$1\text{ Sv} = 100\text{ rem}$	
Radioactivity	$1\text{ mCi} = 37\text{ mBq}$	

In this table, System International (SI) units and conventional units are listed along with the conversion factors needed to move between systems.

Table 1.9 Maximum allowable radiation exposures per year^a

Tissue	Dose Limit
Whole body (10 mm)	5 rem (0.05 Sv)
Lens of the eye (3 mm)	15 rem (0.15 Sv)
Skin of whole body or any extremity (0.07 mm)	50 rem (0.50 Sv)
Any organ	50 rem (0.50 Sv)
Declared pregnant	0.5 rem (0.005 Sv)
Declared under 18	0.5 rem (0.005 Sv)
General public (non-radiation workers)	0.1 rem (0.001 Sv)

^aThis table lists the maximum allowable radiation exposures per year for various body parts. Whole body refers to deep exposure, which is 10 mm tissue depth. The lens of the eye is 3 mm tissue depth. Skin is considered shallow exposure, 0.07 mm tissue depth.

Protection from x-rays

X-rays cannot be detected by human senses and the damage caused by x-rays is not immediately apparent. Not only is protection from radiation a good idea, it is required by law. Per the US federal government; all reasonable efforts must be made to keep exposure to ionizing radiation as low as practical. Law also requires that all individuals working with radiation receive proper training in the safe use of any device that produces radiation.

The guiding principle of radioprotection is **ALARA**, which stands for **As Low As Reasonably Achievable**. Exposures to ionizing radiation must be as far below dose limits as practical while accomplishing the desired diagnostic or therapeutic goal. The three major tenets of ALARA are *time*, *distance*, and *shielding*.

Time: minimize the amount of time spent near x-rays. The damaging effects of radiation are cumulative. The more time spent in the presence of x-rays, the greater the health risk. Radiology personnel should be on rotating shifts to avoid any individual receiving excessive exposure. Only persons necessary to the imaging procedure should be in the x-ray room during an exposure. When possible, avoid manual restraint of patients during radiography. Minimize the need to repeat radiographs by ensuring the x-ray machine settings are correct and the patient is properly positioned. Every repeat radiograph increases the exposure to x-radiation by another 100%.

Distance: maximize the distance from x-rays. The greater the distance from the source of x-rays, the lower the exposure. Recall the inverse square law; doubling the distance reduces the x-ray beam intensity by a factor of four (Figure 1.28). Distance is perhaps the most effective way to reduce x-ray exposure. Just taking two steps back, away

from the x-ray machine, can reduce your exposure by 75% or more. Try to avoid manual restraint of patients whenever possible.

Shielding: use a barrier to block exposure to x-rays.

A barrier made of a material that absorbs x-rays and placed in the path of the x-rays will decrease exposure. Shielding should be used to protect against both primary and secondary or scatter radiation. Lead is a commonly used material because of its large, densely packed atoms (high atomic number and high weight per volume). Other materials such as concrete and certain metals also are used for shielding. Different materials attenuate x-rays to different degrees.

The attenuating ability of material often is compared to lead and expressed as **lead equivalents**. A lead equivalent is the thickness of lead that would block the same amount of x-rays as the thickness of the material being used. Shielding materials are assessed by **half-value layer**. The half-value layer (HVL) is the thickness of a material that would be needed to reduce the x-ray beam intensity by half. HVL, however, is dependent on x-ray energy. The higher the energy of the x-rays, the thicker the material needed (Table 1.10). For example: the HVL of lead for a 50kVp x-ray beam is 0.06mm. The HVL needed for a 100kVp beam is 0.27mm, more than four times the thickness needed for the 50kVp beam.

Three types of shielding are used in radiography: *structural*, *equipment*, and *personal*. **Structural shielding** is embedded in the walls and doors of the x-ray room. In the United States, structural shielding must limit radiation exposure to persons outside the x-ray room to less than 1mGy/year (less than 0.02mGy/week). **Equipment shielding** surrounds the x-ray tube. There must be enough shielding to prevent the emission of x-rays other than through the tube window. Mobile shields are commercially available and provide protection to those behind the shield. It is important to know the rating of the shielding material you are depending on. Some are meant for secondary radiation, while others may protect against primary radiation (up to a stated limit of energy).

Personal shielding is protective clothing. Personal shielding includes protective aprons, gloves, glasses and thyroid shields (Figure 1.68). These are essential components of radiation safety. Protective clothing is designed to protect against scatter radiation, NOT primary x-rays. Protection against primary x-rays would require much thicker and heavier equipment. Personal shielding should provide at least 0.25mm lead equivalents. Lead aprons, thyroid shields and gloves typically supply 0.25–0.50mm lead equivalents. Lead glasses generally provide 0.75mm lead equivalents of shielding (eyes are radiosensitive). The life expectancy of lead aprons and gloves is about 10 years, depending on their use and storage. Periodically inspect and radiograph all items for cracks in the shielding material. Personal shielding ONLY is protective when it is worn. Modern radiation protective

Table 1.10 Approximate HVLs of lead and concrete^a

X-ray Energy (kVp)	HVL Lead (mm)	HVL Concrete (mm)
50	0.06	4.32
100	0.27	15.10
150	0.30	22.32

^aThis table lists the approximate half-value layers (HVLs) of lead and concrete for various x-ray energies.

clothing is much more functional than in the past and gets easier to use with practice.

Radiation safety controls are designed to provide a safe working environment. Physical safety controls include shielding around the x-ray room and equipment, making sure protective clothing is available to all radiology personnel, and restricting access to radiation areas.

Administrative safety controls include radiation warning signs that are clearly posted at all radiology facilities (Figure 1.69). All radiology personnel must receive documented training and must be monitored on a regular basis for radiation exposure. Protocols for normal operating procedures and emergency situations should be written down, readily available, and strictly followed. This is the job of the radiation safety officer.

All organizations that are licensed to use x-rays must have a **Radiation Safety Officer** (RSO). The RSO is the point of contact for all activities associated with the use of x-rays and ensures that radiation safety protocols are in place and that people adhere to them. The RSO makes sure all individuals that use the x-ray equipment are appropriately trained, supervised, and formally authorized. He or she is responsible for knowing the appropriate governmental radiation safety regulations and ensures that all rules and procedures for the safe use of x-rays are observed. Lack of knowledge is indefensible according to most government agencies. The RSO must keep accurate records of the use and maintenance of the x-ray equipment and ensure that it is protected from unauthorized access or removal. In the US, veterinary x-ray machines may be inspected every 3 years by the state department of health. These inspections frequently are random and unannounced.

Dose creep

There is a tendency among radiographers to use higher than necessary x-ray exposures to avoid repeat radiographs. Digital image receptors are far more forgiving of too many x-rays than too few x-rays. The higher the x-ray exposure, the greater the signal-to-noise ratio (SNR) and the less the chance of a poor quality radiograph. Increases in exposure usually start small and may initially go unnoticed. Over time, however, the exposures gradually get higher and higher, which is



Figure 1.68 Radiation protective clothing. This image depicts the typical appearance of a lead apron, lead gloves, thyroid shield, and protective glasses.

why the problem is called *dose creep*. Dose creep is not an issue in film radiography because a higher than needed exposure results in a visibly overexposed radiograph. Dose creep is a growing problem in digital radiography and can lead to significant increases in radiation exposure to the patient and nearby personnel and a shortened life of the x-ray tube.

In general, there is an optimum x-ray exposure that produces a good balance between radiograph quality and radiation dose. Regular monitoring of the exposure techniques used at a hospital or clinic can help identify these optimum exposures. Monitoring can be done using the EI and S-numbers as discussed earlier with digital image receptors. The Optimum EI and S-numbers will differ between different digital systems and different clinical procedures. You will need to determine the acceptable EI and S-numbers for your particular radiography facility.

Dose creep is less likely to occur with higher quality digital image receptors because they typically produce less noise and a higher SNR at standard exposures. Digital noise reduction software also can be used to increase the SNR, especially at low exposures. Automatic exposure control timers (AEC timers) and anatomical programming software can help standardize x-ray exposures with less user input. Control of dose creep is best managed when

exposure records are accurately maintained and periodically reviewed.

Dosimetry

In the US, federal law requires that any individual who is likely to receive more than 10% of any annual occupational dose limit be monitored for radiation exposure. The Occupational Safety and Health Administration (OSHA) standard 1910.1096 states, "Every employer shall supply appropriate personnel monitoring equipment, such as film badges, pocket chambers, pocket dosimeters, or film rings, and shall require the use of such equipment." Other countries and jurisdictions have similar regulations about monitoring x-ray exposure.

Monitoring typically involves a personal radiation detection device called a **dosimeter**. Dosimeters are attached to the users clothing. They must be worn outside of protective clothing to accurately measure the radiation dose. Whole-body dosimeters commonly are attached somewhere between the neck and the waist.

Dosimeters generally are worn for a specific period of time and then processed. Most are examined about every 3 months to determine what dose of radiation the user received. The two most common types of dosimeters currently in use are *film badges* and *thermoluminescent dosimeters* (Figure 1.70).

Film badges, as the name implies, measure exposure to x-rays using a piece of radiation-sensitive film. The film is packaged in a tightly sealed envelope that prevents light, moisture and chemicals from affecting it. The envelope is enclosed inside a plastic holder called a "badge." Inside the badge are a series of filters that differentially attenuate x-rays. The filters are used to calculate the energy of the x-rays based on the amount of radiographic density under each one. The film is double-coated so the emulsion on one side contains large grains which are sensitive to low x-ray exposures and

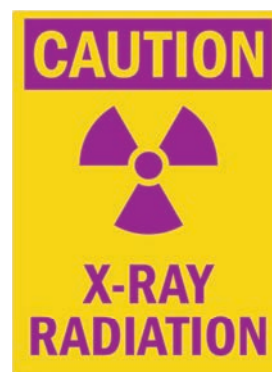


Figure 1.69 Radiation warning sign. This is an example of a standard international sign used to warn people about onsite radiation. It typically is yellow with purple lettering and includes the symbol for radiation.



Figure 1.70 Types of dosimetry. Pictured are a film Badge (A) and a Thermoluminescent dosimeter (B). The plastic badge is propped open with the envelope containing the film sitting outside it. The series of filters is visible.

the other side contains small grains to detect high exposures. The film must be sent to a lab for processing. If the large grain emulsion is overexposed, the film was exposed to higher

x-ray intensity and the user's exposure dose is computed using the small grain emulsion. Film badges are simple, inexpensive, and very reliable. They provide a permanent record of exposure and are quite accurate for x-ray exposure doses greater than 0.1 rem. In addition, different x-ray energies can be distinguished. The disadvantages of film badges include: the film can only be processed once, the film is susceptible to heat and pressure fogging, and the film must be sent away for processing, which is more time-consuming than using an onsite reader.

A **thermoluminescent dosimeter** (TLD) contains phosphors that absorb x-ray energy. When heated, the phosphors release the captured energy as light, which is called thermoluminescence. The amount of light released is proportional to the amount of x-rays absorbed. Instead of reading the amount of darkening in a piece of film, a TLD reader measures the amount of light emitted by the phosphors. TLDs are able to measure a greater range of x-ray exposure doses than film badges. The exposure dose is easily obtained and can be read onsite. TLDs can be reused many times, but each exposure dose can be read out only once, so TLDs do not provide a permanent record of exposures.

