CHAPTER 4

MEDICAL IMAGING MODALITIES: X-RAY IMAGING

As introduced in Chapter 1, medical imaging is an essential aspect of radiological sciences for visualization of anatomical structures and metabolic information of the human body. Structural and functional imaging of a human body is important for understanding human anatomy, function of organs, and associated physiological processes. Imaging also becomes a critical source of information to study the physiological behavior of an organ or tissue under a treatment. Thus, imaging is an essential tool for diagnosis as well as evaluation of the treatment of an illness.

Medical imaging modalities can be broadly classified into two classes: (1) anatomical or structural and (2) functional or metabolic (1–4). The characterization of a specific anatomical imaging modality depends on its ability to discriminate different constituents of the body such as water, bone, soft tissue, and other biochemical fluids. Examples of major anatomical imaging modalities include X-ray imaging (X-ray radiography, X-ray mammography, X-ray computer tomography [CT]), ultrasound, and magnetic resonance imaging (MRI). The characterization of a specific functional imaging modality depends on its ability to discriminate different levels of metabolism caused by specific biochemical activity that may be generated by the uptake of a radiopharmaceutical substance. The biochemical activity describing the functional behavior of the tissue or organ can be caused by any internal or external stimulation. Major functional imaging modalities include functional MRI (fMRI), single photon emission computed tomography (SPECT), positron emission tomography (PET), and fluorescence imaging. For example, fMRI methods can be used to measure the blood flow or oxygenation level in brain tissue. The changes in blood flow or oxygen level in the tissue reflect neural activity in the brain caused by stimulation such as sound or light. In nuclear medicine imaging modalities, blood flow in the tissue or an organ can be measured through an emission process of a radioactive tracer that is administered in the blood. For example, a PET image obtained through the administration of flurodeoxyglucose (FDG) may show blood flow and glucose metabolism in the tissue that may be affected by a disease such as a tumor or epilepsy (1–8).

Medical Image Analysis, *Second Edition*, by Atam P. Dhawan

4.1. X-RAY IMAGING

X rays were discovered in 1895 by Conrad Roentgen, who described them as a new kind of ray that could penetrate almost anything. He described the diagnostic capabilities of X rays for imaging human body and received the Nobel Prize in 1901. X-ray radiography is the simplest form of medical imaging with the transmission of X rays through the body, which are then collected on a film or an array of detectors. The attenuation or absorption of X rays is described by the photoelectric and Compton effects, with more attenuation through bones than soft tissues or air (1–5).

X rays are a part of the electromagnetic spectrum with smaller wavelengths than the ultraviolet or visible light spectrum. Because of their smaller wavelength, X rays have high energy providing excellent capability of straight penetration and transmission in human body. Soft X rays are identified as having wavelengths from 10 nm to 0.10 nm (or 100 Å to 1 Å), corresponding to 120 eV to 12.3 keV or 3×10^{16} Hz to 3×10^{18}, respectively. Wavelengths shorter than 0.1 nm (1 Å) up to 0.001 nm (0.01 Å) are considered hard X rays. For medical applications, it is important to select X rays with wavelengths that provide linear attenuation coefficients for human body. X-ray radiation photons can be used for both diagnostic and therapeutic applications. Lower energy X rays are used for diagnostic imaging, while higher energy photons are utilized in radiation therapeutic applications. Usually, X rays between 0.1 nm and 0.01 nm (1 Å to 0.1 Å, with corresponding 12.3 keV to 123 keV energy range) are used for diagnostic purposes. In this range the attenuation is quite reasonable to discriminate bones, soft tissue, and air. In addition, the wavelength is short enough for providing excellent resolution of images, even with submillimeter accuracy. Wavelengths shorter than those used for diagnosis provide much higher photon energy and therefore less attenuation. Increasing photon energy makes the human body transparent for the loss of any contrast in the image. The diagnostic X-ray wavelength range provides sufficient energy per photon with a refractive index of unity for almost all materials in the body. This guarantees that the diffraction will not distort the image and rays will travel in straight lines (1–8).

For medical imaging applications, an X-ray imaging tube is used as an external ionized radiation source to generate an X-ray radiation beam that is transmitted through human body in a scanning mode. As the radiation beam passes through a specific location, X rays undergo absorption and scattering as described in Chapter 3. In other words, the radiation beam is attenuated due to the mass attenuation coefficients of physiological structures in the body. The attenuation of radiation intensity is determined at each scan location by measuring the difference of intensity between the source and detector. A two-dimensional (2-D) attenuation map, as obtained through scanning the object in the respective geometry, can be recorded on a radiographic film for an X-ray (film) radiograph. Alternatively, the attenuation map can be stored digitally using X-ray detectors and electronic instrumentation to display a digital X-ray radiograph. Furthermore, the attenuation measurements can be obtained through a three dimensional (3-D) scanning geometry to define projections all around the object. Attenuation projection data are then used in image reconstruction algorithm (described in Chapter 8) to reconstruct and display 3-D CT images. More details are provided later in this chapter.

4.2. X-RAY GENERATION

X rays are generated as the result of interactions of high-speed electrons with heavy target atoms such as tungsten or molybdenum. In principle, an accelerated electron loses energy in interaction with an atom and the loss of energy emits X-ray photons in a scattered direction. The basic principle is explained in detail below.

An atom is comprised of a nucleus and electrons that revolve around the nucleus in orbits called shells. In a nucleus, protons carry positive charge while neutrons possess no charge. Protons and neutrons basically provide mass to an atom, while the number of electrons defines the atomic number. As an atom is neutral in charge, the number of protons is equal to the number of electrons. Each electron shell has a characteristic energy level that represents the binding energy of the electron to the corresponding shell. Also each shell has a maximum energy level depending on its position from the nucleus for the stability of the atom. The K-shell that is closest to the nucleus has greater binding energy than the next L-shell. An electron can be ejected or transferred to another shell from its original place depending on the energy exchange that is caused by an interaction of the atom with a quantum. In each interaction, the total energy is preserved according to the principle of energy preservation. The energy lost by an electron due to change in path is emitted in the form of X-ray photon. For example, if an incident electron with energy greater than the binding energy of K-shell interacts with an atom, the K-shell electron is ejected, leaving an ionized atom with a vacancy in the K-shell. This vacancy is filled by an electron from the outer L-shell. Because the total energy has to be preserved, the difference between the binding energies of K- and L-shells causes the release of a characteristic X-ray photon as a result of the transition of the electron from the L- to K-shell.

Since each element has its own shell-binding energy levels that are specific to its atomic structure, the energy of the X-ray photon emitted in the interaction process is a characteristic feature of that element. For example, in the case of tungsten, a commonly used element for X-ray generation, the specific K- and L-shell binding energy levels are, respectively, 69.5 and 10.2 keV. Thus, an interaction of incident electron of energy greater than 69.5 keV with an atom of tungsten yields an emission of X-ray photon of 59.3 keV (Fig. 4.1). In the case of tungsten, an incident electron, also called an accelerating electron, must have energy greater than 69.5 keV. It is apparent that in such interactions, electrons striking the target material can go through interactions with several nuclei before they are stopped, resulting in electrons with different energy levels. Therefore, emitted X-ray photons also produce a distribution of energy called white radiation or Bremsstrahlung radiation spectrum. Along with the white radiation, characteristic emissions occur due to specific transitions of electrons from different shells such as L to K, M to K, or N to K, as shown in Figure 4.2. The energy of X-ray photons emitted from transitions of electrons to regions other than the K-shell are not used in diagnostic radiology because they are usually of low energy that can be easily absorbed in the medium.

In an X-ray generation tube electrons are released by the source cathode and are accelerated toward the target anode in a vacuum under the potential difference ranging from 20,000 to 150,000 V. Depending on the radiological application,

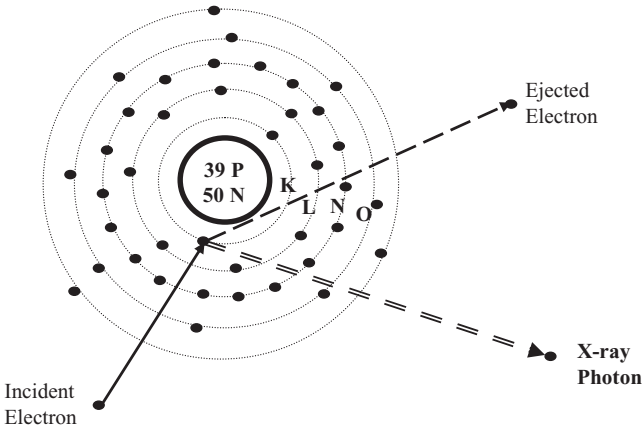


Figure 4.1 Atomic structure of a tungsten atom. An incident electron with energy greater than K-shell binding energy is shown striking a K-shell electron for the emission of an X-ray photon.

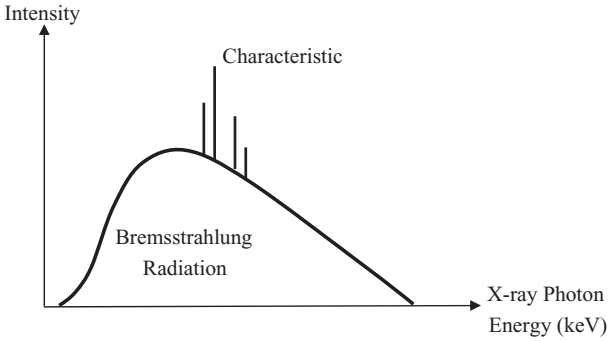


Figure 4.2 A typical X-ray radiation distribution spectrum with characteristic radiations.

specific element materials are used in the X-ray tubes for building the source cathode and target anode in a suitable configuration to maximize X-ray emission. The desired energy level of X-ray photons as a focused monochromatic beam is achieved through appropriate filtration using attenuation mediums absorbing the undesired low-energy emissions. Lead-based collimators are used to define an adequate field dimension of the radiation beam suitable for the specific area of the patient to be radiated for imaging.

Important parameters for selection of X-ray tube for diagnostic imaging include the size of the focal spot, spectrum of X-ray energies, and the X-ray intensity. It is obvious that the focal spot size defines the potential spatial resolution in the X-ray imaging system such as digital radiography, mammography, and CT. Since the attenuation coefficients of X rays and their penetration capabilities in human tissue are wavelength-dependent, it is important that the X-ray source system provides a monochromatic or short band spectrum of appropriate energy at the suitable intensity level. The X-ray intensity level directly defines the signal-to-noise ratio

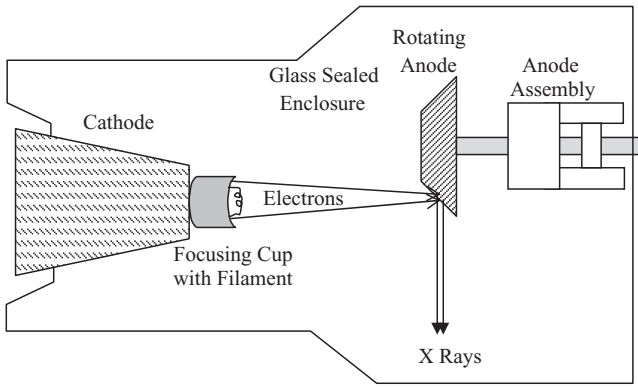


Figure 4.3 A simplified schematic diagram of an X-ray generation tube with rotating anode.

Figure 4.4 A picture of rotating anode X-ray tube.

(SNR) of the imaging system. Higher intensity levels provide more reliable counts at the detector.

An X-ray tube has a cathode block with a focusing cup and a hot filament that determines the focal spot size. Often a dual filament assembly is used to provide two focal spot sizes. X-ray CT systems typically utilize spot sizes ranging from 0.5 to 2 mm. The higher the spot size, the larger the power requirement. For example, a spot size of 1.8 mm may require a 2000 W power load, versus a spot size of 0.8 mm for which a power load of 500 W may be sufficient. Figure 4.3 shows a simplified schematic diagram of an X-ray generation tube; an actual tube is shown in Figure 4.4. As shown in Figure 4.1, the accelerated electrons bombard the target atoms available at the stationary or rotating anode where X rays are generated in a scattered direction through an exit window of a sealed glass enclosure. The rotating anode with an annular target close to the edge (Fig. 4.4) is very efficient in X-ray production and quite commonly used in X-ray tubes. Because the focal spot can become

very hot, the anode is made out of material that can withstand high temperatures (in the range of 2000–3000°C). High-performance digital radiography and CT systems utilize X-ray tubes of less than 1 mm focal spot size and a power rating up to 100 kW.

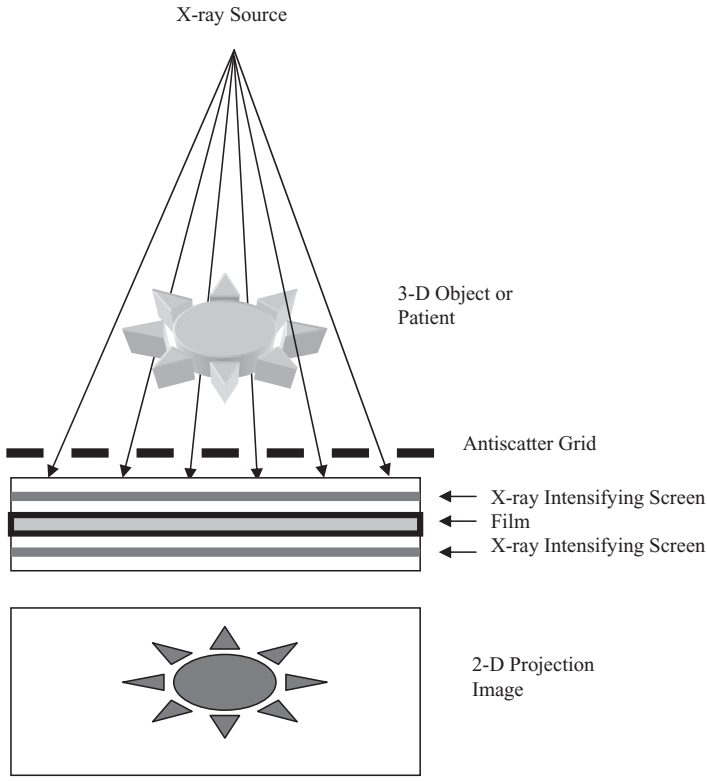
4.3. X-RAY 2-D PROJECTION IMAGING

Two-dimensional projection radiography is the oldest medical imaging modality and is still one of the most widely used imaging methods in diagnostic radiology (1, 2). Conventional film radiography uses an X-ray tube to focus a beam on the imaging area of a patient's body to record an image on a film. The image recorded on the film is a 2-D projection of the three-dimensional (3-D) anatomical structure of the human body. The image is thus obtained through transmission of X rays through the body. The film is developed to show the image that represents the localized sum of attenuation coefficients of material (such as air, blood, tissue, or bone) present in the body along the X-ray path. X-ray screens are used to improve the sensitivity of X-ray detection on the film and therefore reduce the required X-ray exposure to the patient (1, 2).

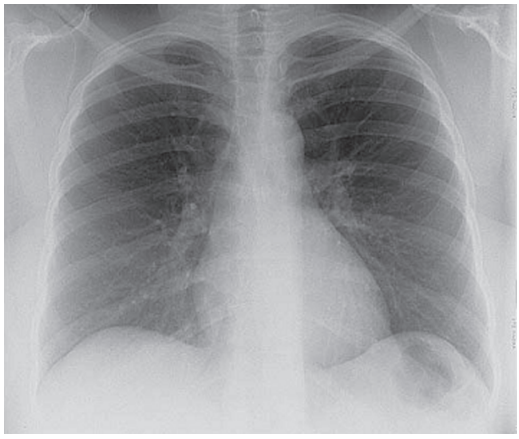
Scattering can create a major problem in projection radiography. The scattered photons can create artifacts and artificial structures in the image that can lead to an incorrect interpretation or at least create a difficult situation for diagnosis. Projection imaging assumes that all photons travel in a straight line so that the sum of the attenuation coefficients along the path can be recorded correctly on the respective location in the image. If there are additional scattered photons that are reaching the same location, the exposure recorded at the location would be inaccurate and may cause a spurious structure in the image that is actually not present in the patient's body. Also, the scattered photon leads to an incorrect exposure and therefore may cause loss of diagnostic information. For example, in X-ray mammography, scattered radiation can mislead a radiologist in the identification of microcalcifications. In quantitative imaging and analysis applications, such as digital subtraction angiography (DSA), scattered radiation can introduce major errors. In film-screen radiography, the scattered radiation also reduces the overall contrast in the image (2, 3, 8).

Antiscatter grids and collimators are used to reduce the number of scattered radiation reaching at the detector or film (see Fig. 4.2). The grids provide tunnels for passing the radiation with no attenuation of the photons that are traveling parallel to the axis. The primary photons are thus unaffected as they are aligned with the axis. The bars are made of high-density material such as lead or tantalum, causing photoelectric absorption of the scattered X-ray photons. Scattered photons enter the tunnel at an angle and hit the lead walls for absorption (2).

X-ray intensifying screens, also called X-ray screens are used to improve the detection of X-ray photons to produce a better quality image on X-ray films. These screens are used along with an antiscatter grid as shown in Figure 4.5. A complete X-ray intensifying screen contains several layers including an outer plastic protective layer that is transparent to X rays (about 15–20 μm thick), a phosphor layer (100–500 μm thick), a light reflecting layer (20 μm thick), and a plastic base layer



(a)



(b)

Figure 4.5 (a) A schematic diagram of a 2-D X-ray film-screen radiographic system. A 2-D projection image of the 3-D object is shown at the bottom. (b) X-ray radiographic image of a normal male chest.

(20 μm thick). The phosphor layer absorbs X-ray photons and produces light that is reflected toward the film for better exposure. The phosphor layer is formed from specific materials with high X-ray-to-light photon conversion efficiency such as terbium-doped gadolinium oxysulfide (Gd_2O_2S:Tb) or terbium-doped lanthanum oxybromide (LaOBr:Tb). The X-ray image intensifying screen improves SNR and image quality on X-ray film.

In case of digital radiography, the combination of intensifying screen and film is replaced by a phosphor layered screen coupled with a charge-coupled device (CCD)-based panel. A solid-state detector system in digital radiography and mammography may use a structured thallium-doped cesium iodide (CsI) scintillation material to convert X-ray photons into light photon, which are then converted into electrical signal by CCDs through a fiber optics coupling interface. Other scintillation material (discussed in Chapter 3) or phosphor screens along with thin CCD-based panels have also been investigated for specific digital radiographic applications.

Film-based radiography provides a very high-resolution image depending on the granular structure of the film emulsion. It may suffer from some artifacts such as uneven contrast due to inconsistencies in film processing and chemical effects in the film-developing process. Underexposed and overexposed films are usually of significant concern because they can contribute to incorrect interpretation or conjecture by the physician. On the other hand, brightness levels are much better controlled with digital detector systems instead of film-based systems. As described above, digital radiographic systems use scintillation crystals optically coupled with CCDs in which electrical output signal sensitivity can be controlled much more efficiently than in a film-based system. The digital detector system also provides excellent linearity and gain control, which directly affects the SNR of the acquired data. For this reason, a digital detection system provides a superior dynamic range compared with the film-based systems. However, the resolution of a digital image is limited by the detector size and data collection method (2).

4.4. X-RAY MAMMOGRAPHY

X-ray film-screen mammography is a specialized radiographic imaging method used for breast imaging for diagnosis of breast diseases. Detection of breast cancer in the early stages, an important clinical health care issue for women, imposes much more challenging requirements for imaging than conventional radiography. Breast tissue is quite vascular and soft with low X-ray attenuation coefficients as compared to other anatomical structures. Detection of architectural distortions, small lesions, and particularly microcalcifications require high spatial resolution on the order of 50–100 microns. The mammographic images are required to have high sensitivity and specificity for early detection of breast cancer, but at the same time, mammographic imaging techniques must minimize scattering radiation to provide accurate imaging of breast tissue with a relatively low radiation dose. Recent advanced X-ray film-screen imaging methods use specialized X-ray tubes, breast compression devices, antiscatter grids, and optimized detector systems to optimize diagnostic capabilities for the early detection of breast cancer (2, 3).

X-ray tubes for mammography operate on a low operating voltage, typically less than 30 kV. X-ray tubes with X-ray energies between 17 and 25 keV are preferred for breast imaging. Typically, molybdenum or rhodium is used as a target material in X-ray mammography tubes to provide the characteristic X-ray radiation in the desired range. The K-, L-, and M-shell binding energies for molybdenum are 20, 2.8, and 0.5 keV, respectively. The shell-binding energy levels of molybdenum yield the characteristic X-ray radiation with energies around 17 keV. Rhodium can provide the characteristic X-ray radiation with energies around 20 keV as its K-, L-, and M-shell binding energies, respectively, are around 23, 3.4, and 0.6 keV. With a specific design of the cathode and anode, the X-ray tube used in mammography provides a small focal spot on the order of 0.1 mm required for high-resolution magnification imaging. Inherent filtration methods are used to make the X-ray tube monochromatic with the X-ray radiation beam tightly distributed around a single energy level. Antiscatter grids and compression devices are used to minimize the scattered radiation along with computer-controlled processing to compensate for breast thickness, breast density, and grid effects. Compression of the breast is critical in obtaining good mammographic images. An appropriate compression of the breast along with oscillations of the moving antiscatter grid provides a significant reduction in X-ray scatter and radiation dose. Magnification of the image is obtained by adjusting the distance between the X-ray source, compressed breast, and the film, as shown in Figure 4.6. However, the magnification factor is critically limited by the blurring effect caused by the size of the cathode focal spot and its distance from the breast tissue. In recent film-screen mammography scanners, a double screen with double film emulsions is used for breast imaging. However, digital X-ray mammographic

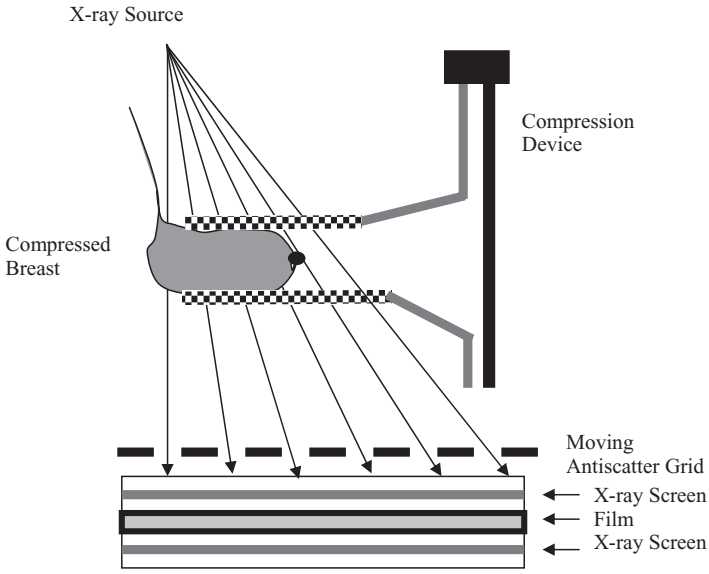


Figure 4.6 A film-screen X-ray mammography imaging system.

Figure 4.7 A digital X-ray mammography scanner.

scanners are also replacing film-screen with scintillation crystal- and semiconductor technology-based digital detector systems (2).

The low-dose X-ray mammographic imaging systems use a tungsten–rhenium target with a power load of about 8 kW to produce low-energy X-ray photons for breast imaging. Recently, digital mammographic systems have been developed with high sensitivity to low-dose mammograms and portability. They utilize a solid-state semiconductor technology-based detection system instead of vacuum tubes-based photomultiplier technology. The detector system in digital mammographic systems use a phosphor screen or scintillation material thallium-doped CsI layer optically coupled with CCDs. Some portable systems such as SenoScan® digital mammography systems can complete a scan in a few seconds with resolution as small as $0.027\,mm^2$ pixel size. A digital X-ray mammographic scanner is shown in Figure 4.7, and a mammographic image of a female breast is shown in Figure 4.8.

4.5. X-RAY CT

X-ray conventional radiography creates a 2-D image of a 3-D object projected on the detector plane. While 2-D projection radiography may be adequate for many diagnostic applications, it does not provide 3-D qualitative and quantitative information about the anatomical structures and associated pathology necessary for diagnostics and treatment of a number of diseases or abnormalities. For example, a tumor

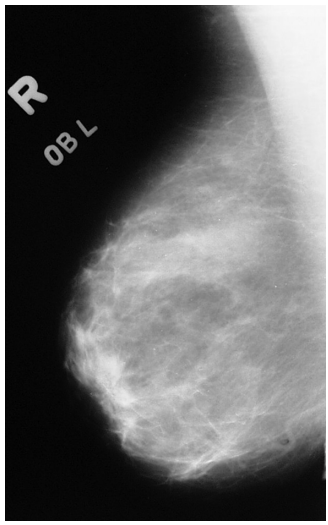


Figure 4.8 An X-ray film-screen mammographic image of a female breast.

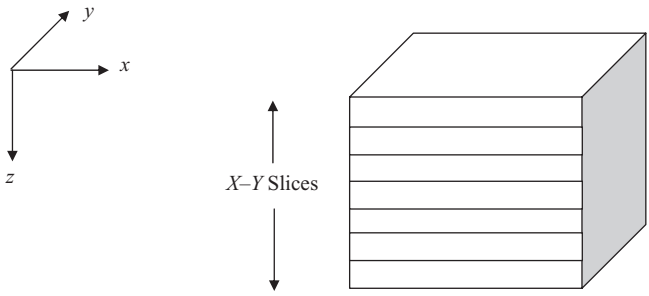


Figure 4.9 3-D object representation as a stack of 2-D *x*–*y* slices.

size and its 3-D shape are important features for diagnostic and therapeutic purposes. Also, a fracture in a bone may require 3-D information for better treatment. Other examples may include 3-D imaging of the heart and brain.

Combining 2-D projection radiography with 3-D scanning geometry and advances in image processing algorithms, 3-D CT provides a very useful and sophisticated imaging tool in diagnostic radiology and therapeutic intervention protocols. The basic principle of X-ray CT is the same as that of X-ray digital radiography. X rays are transmitted through the body and collected by an array of detectors to measure the total attenuation along the X-ray path.

Let us assume a 3-D object to be a stack of 2-D slices as shown in Figure 4.9. Let us consider a monochromatic X-ray radiation source aligned with a detector such that the source–detector pair translates to cover the entire 2-D slice for scanning (Fig 4.10). The output intensity of a radiation beam parallel to the *x*-direction for a specific *y*-coordinate location, $I_{\text{out}}(y; x, z)$ would be given by

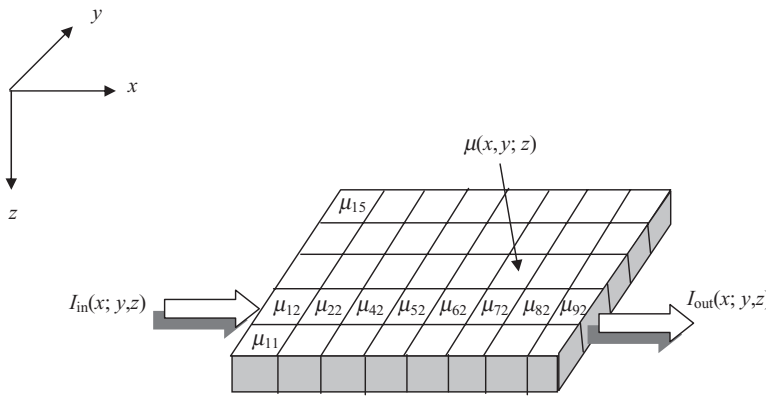


Figure 4.10 Source–detector pair-based translation method to scan a selected 2-D slice of a 3-D object to give a projection along the *y*-direction.

$$I_{\text{out}}(y; x, z) = I_{\text{in}}(y; x, z)e^{-\int \mu(x,y;z)dx}. \tag{4.1}$$

One simple way of performing a 3-D scan is to use a pair of point X-ray source and point detector aligned along the ray parallel to the *x*-axis (2, 8–10). For each position along the *y*-axis, the source–detector pair will provide one measurement for the *y*-direction projection of the selected slice. Now, if a projection along *x*-direction is to be obtained, the source and detector are aligned along the *y*-direction and then translated along the *x*-axis. To obtain a projection of the 2-D slice at a different angle, we need to translate the source–detector pair along the direction parallel to the viewing angle. Thus, the angular scan is obtained through the rotation of the source–detector assembly around the object. For a specific viewing angle, the 2-D scan is performed through a linear motion involving translation of the source and detector pair. Figure 4.11 shows a translate–rotate parallel-beam geometry for obtaining a set of projection around a selected 2-D slice of the 3-D object. Such a principle was used in first generation CT scanners. From 1-D projections obtained at different angles around the selected 2-D slice, a 2-D image is reconstructed on a computer using a reconstruction algorithm such as the filtered backprojection method (image reconstruction algorithms are presented in Chapter 8). Thus, a 3-D object can be scanned slice-by-slice along the *z*-axis as shown in the 3-D representation of stacked slices in Figure 4.9. Correspondingly, a 3-D reconstructed image can be displayed by stacking the reconstructed 2-D slice images. The 3-D reconstructed image is usually interpolated between the slices to obtain a smooth 3-D representation.

The scanner geometry for X-ray CT had several stages of development through four generations. As described above, the first-generation CT scanner utilized an X-ray source–detector pair that was translated in parallel-beam geometry to acquire projections related to a viewing angle. The source–detector pair geometry was then rotated to obtain additional views as shown in Figure 4.12. The second-generation X-ray CT scanners used a fan-beam geometry with a divergent X-ray source and a linear array of detectors. The scanning technique was still based on translation to

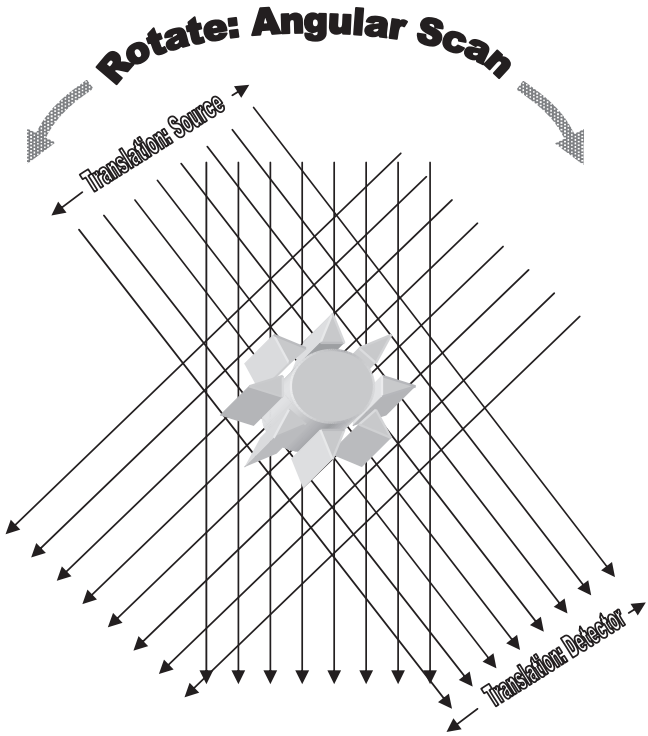


Figure 4.11 The translate–rotate parallel-beam geometry of first-generation CT scanners.

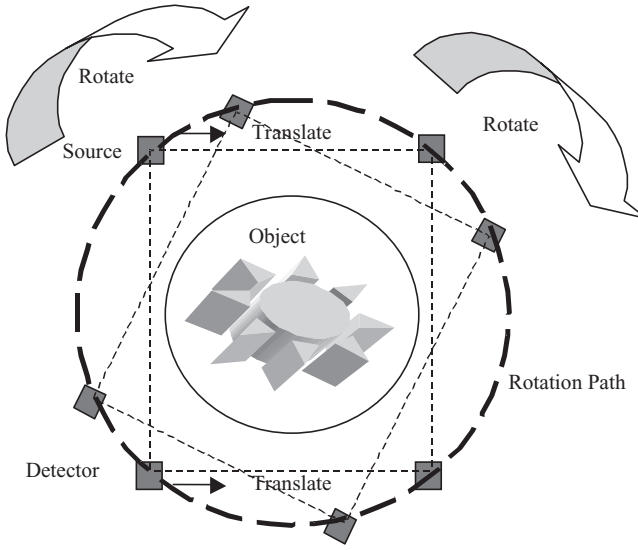


Figure 4.12 The first-generation X-ray CT scanner geometry.

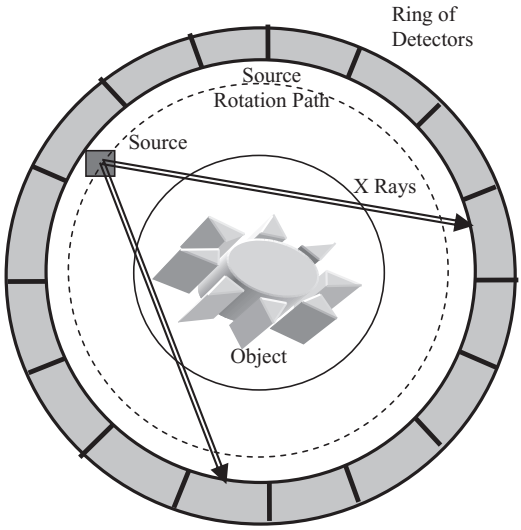


Figure 4.13 The fourth-generation X-ray CT scanner geometry.

cover the object and rotation to obtain additional views. The third-generation X-ray CT scanner used a fan-beam geometry with a divergent X-ray source and an arc of detectors. The divergence of the X-ray fan beam was designed to cover the object without any translation. Thus, the entire projection was obtained from a single position of the X-ray source. Additional views were obtained by simultaneous rotation of the X-ray source and detector assembly. This scanning geometry was called "rotate-only." The fourth-generation X-ray CT scanners use a detector ring around the object as shown in Figure 4.13. The X-ray source provides a divergent fan beam of radiation to cover the object for a single ring of detectors. Scanners with multiple rings of detectors utilize a cone beam to cover multiple slices (8–12). Recent fast spiral CT scanners use a spiral movement instead of parallel movement for selection of axial slices (see the next section). Fourth-generation scanners normally use a ring of 720 or more detectors that are equally spaced around the circumference of a circle. The detectors used in X-ray CT scanners are ionization chambers filled with xenon gas or scintillators with photomultiplier tubes. Recent hybrid X-ray CT scanners such as CT-PET or CT-SPECT also utilize scintillation detectors coupled with photomultiplier tubes or flat panel scintillation detectors coupled with solid-state CCDs.

A modern X-ray CT scanner is shown in Figure 4.14. Figure 4.15 shows a CT reconstructed image of a selected 2-D slice of the 3-D cardiac cavity of a cadaver. For comparison purposes, the pathological section of the selected slice is shown in Figure 4.16.

4.6. SPIRAL X-RAY CT

Certain applications such as brain imaging requires high-resolution imaging that is produced by collecting data for consecutive axial slices (with parallel sections) by

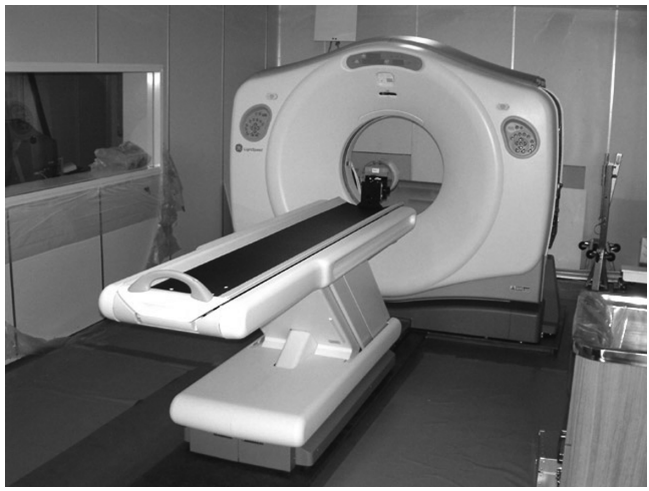


Figure 4.14 An X-ray CT scanner.

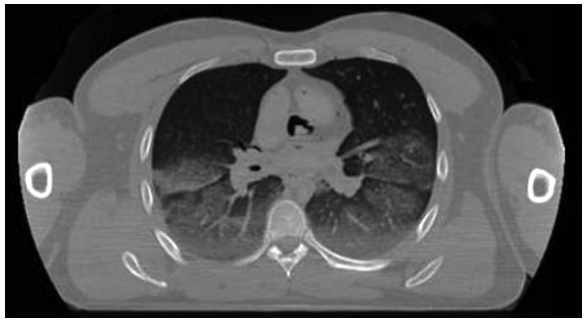


Figure 4.15 X-ray CT image of a selected slice of cardiac cavity of a cadaver.

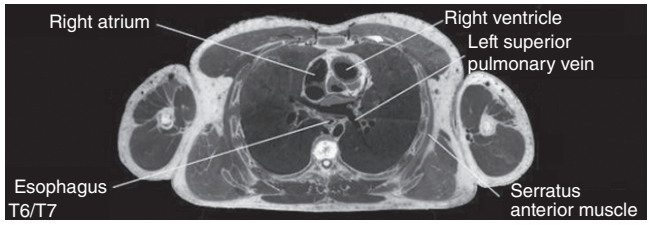


Figure 4.16 The pathological image of the selected slice shown with the X-ray CT image in Figure 4.15.

an X-ray CT scanner. This usually requires longer scan time. A faster scan at lower radiation dose can be obtained for other body imaging applications by spiral (helical) CT while slightly sacrificing image resolution and sharpness. Nevertheless, a faster spiral CT scan provides images with higher temporal resolution to allow investigation of the bolus of the contrast agent administered in the body. In full-resolution

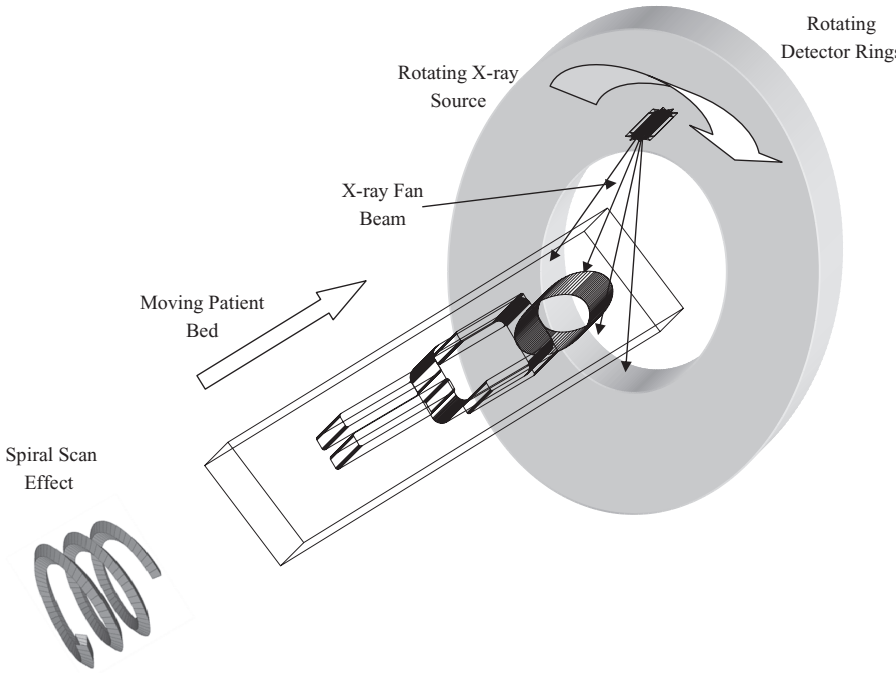


Figure 4.17 A schematic diagram of a spiral CT scanner with spiral effect caused by the movement of patient bed.

CT imaging the patient bed is kept stationary while the source–detector ring gantry is translated to select axial slices. In spiral CT, the patient bed is moved at a constant speed into the gantry space during imaging while the gantry is rotated within the circular opening. The forward movement of the patient bed enables the sampling points to provide the data along a spiral or helical path as shown in Figure 4.17. Due to the motion of patient bed, total scan time for collecting data for multiple sections is significantly reduced. For a single detector ring-based configuration, an X-ray source with a fan-shaped beam is used with width ranging from one to several millimeters. For multislice volume imaging, spiral CT scanners use multiple detector rings with an X-ray source that emits a cone-shaped beam to cover multiple sections at the same time. The spiral CT scans collect continuous data during the entire scan with the rotating gantry. Therefore, a spiral CT scanner requires an X-ray source with better cooling capabilities and detectors with better efficiency. The slip-ring configuration of a spiral CT scanner allows rotation of both X-ray source and detector ring for faster gantry rotation. The slip-ring gantry utilizes a set of rings with electrical components that can rotate and slide seamlessly to enable continuous scanning. During the scanning, slice thickness (t in mm) is determined by the pitch, which is defined by the movement of bed (d in mm) one complete rotation (360 degrees) of gantry as

$$p = \frac{d}{t}. \tag{4.2}$$

For example, a pitch of value "1" for a slice thickness of 2 mm would require the patient's bed to move 2 mm per second if the gantry is rotated 360 degrees per second. In this case, the total scan time of raw data collection will be 30 s to cover a 60 mm wide scanning volume with 2 mm thick contiguous slices. The raw data collected at sampling points distributed along the spiral (helix) is re-binned and interpolated (and even extrapolated as needed) to create projection data that is then used to reconstruct sectional images of desired slice thickness. By combining the detector outputs, images of different slice thickness can be easily reconstructed from the same data. Details about the projection data and reconstruction methods are discussed in Chapter 8.

Multislice spiral CT scanners are used for fast scanning (higher temporal resolution) in dynamic volume imaging in a number of clinical applications including trauma, cardiovascular, abdominal, skeletal, and pediatric imaging.

4.7. CONTRAST AGENT, SPATIAL RESOLUTION, AND SNR

To improve contrast of selected physiological structures in X-ray images, materials that increase the total attenuation coefficient of X rays are used orally or through an intravenous injection. For example, barium sulfate is administered orally to the patent to enhance contrast in upper gastrointestinal (GI) tract imaging, including the esophagus and stomach. Barium atom has a K-edge at 37.4 keV, which causes much higher attenuation of X rays. Thus, when barium sulfate is consumed through the GI tract, respective areas in the image are seen with higher contrast because of increased attenuation. Iodine also has a K-edge at 33.2 keV, causing greater attenuation of X rays. Iodine-based contrast agents are used through intravenous injection in angiography, urography, and intra-arterial DSA to improve visibility of arteries and blood vessels. In X-ray angiography, a series of images with and without contrast agent are compared to visualize the arteries and blood vessels. In DSA, digital images of the tissue before and after administering the contrast agent are taken and subtracted to enhance to contrast between the blood vessels and tissue. Iodine-based contrast agents can also be used in brain imaging for detection of tumors and metastases.

Spatial resolution in X-ray imaging modalities depends on the effective size of the focal spot of X-ray tube, and distance between the X-ray source and the patient due to magnification of the point spread function as discussed in Chapter 2. In addition, the thickness of the intensifying screen and the speed of X-ray film directly affect the spatial resolution in X-ray film-screen imaging. The size and shape of the collimators and detectors also impact considerably on the spatial resolution of X-ray images.

The signal-to-noise ratio of X-ray imaging modalities depends on the X-ray exposure, source and detector instrumentation, thickness and heterogeneity of the tissue, scattering, and contrast agents. The SNR of X-ray imaging is proportional to the square root of the product of the exposure time and X-ray tube current. X rays with higher monochromatic energy provide higher SNR. As X rays are attenuated

in the body, they lose energy as they move forward in thicker tissue. Since the attenuation coefficients are energy-dependent (higher energy photons attenuate less), the X-ray beam does not provide same attenuation coefficients for the same type of tissue at the entrance and exit levels of a thick medium. This is called beam hardening, which causes errors and lower SNR in reconstructed images in CT. Specific estimation methods can be used to compensate for the beam-hardening effect based on the thickness of the medium (2, 9). In addition, nonuniform responses of the intensifying screen, phosphor, and scintillation detectors causes lower SNR in X-ray images. In X-ray CT, spatial resolution and SNR of reconstructed images are directly correlated with the number of projections (angular sampling) and number of detectors (projection sampling). For higher spatial resolution, data must be acquired with higher sampling rate in the projection space. However, in practice, the higher sampling rate is compromised by the limitations on scan time, method of data collection, and detector size and efficiency.

4.8. EXERCISES

4.1. Describe the major differences between the anatomical and functional imaging modalities. Identify at least two imaging modalities in each category.

4.2. What is the principle of X-ray computed tomography (CT)?

4.3. How are X rays generated? What is the range of diagnostic X rays? Justify the selection of this range.

4.4. What types of detectors are commonly used in X-ray CT?

4.5. How is the mammography imaging different from conventional X-ray chest radiography? What are the challenges in mammography imaging?

4.6. What is the detector system used in digital mammography systems?

4.7. What contrast agents can be used with X-ray imaging? Describe their significance.

4.8. What are the advantages of fourth-generation CT scanners over the second generation systems?

4.9. What is the difference between a fourth-generation X-ray CT and spiral CT imaging systems?

4.10. If the pitch of a spiral scan is changed from 1 to 2, what will be the effect on the radiation dose?

4.11. X rays of 25 keV are received with an X-ray intensifying screen that produces light photons at 425 nm. If the conversion efficiency of intensifying screen is 20%, calculate how many light photons will be generated by an X-ray photon.

4.12. What is an X-ray contrast agent and why is it used in imaging?

4.13. What consideration should be taken to improve the signal-to-noise ratio (SNR) of X-ray CT images? Describe if there is any harmful effect associated with them.

4.14. Display an axial brain image of a CT scan with brain hemorrhage in MATLAB. In another window, display axial MR brain images including proton density, T_1-, and T_2-weighted images of a stroke patient. Compare and comment on the contrast of the primary hemorrhage and soft tissue regions in the CT and MR images.

4.9. REFERENCES

1. H. Barrett and W. Swindell, *Radiological Imaging: The Theory of Image Formation, Detection and Processing*, Volumes 1–2, Academic Press, New York, 1981.
2. J.T. Bushberg, J.A. Seibert, E.M. Leidholdt, and J.M. Boone, *The Essentials of Medical Imaging*, Williams & Wilkins, Philadelphia, 1994.
3. Z.H. Cho, J.P. Jones, and M. Singh, *Fundamentals of Medical Imaging*, John Wiley & Sons, New York, 1993.
4. A.P. Dhawan, "A review on biomedical image processing and future trends," *Comput. Methods Programs Biomed.*, Vol. 31, No. 3–4, pp. 141–183, 1990.
5. K.K. Shung, M.B. Smith, and B. Tsui, *Principles of Medical Imaging*, Academic Press, San Diego, CA, 1992.
6. G.N. Hounsfield, "A method and apparatus for examination of a body by radiation such as X or gamma radiation," Patent 1283915, The Patent Office, London, 1972.
7. G.N. Hounsfield, "Computerized transverse axial scanning tomography: Part-1, description of the system," *Br.J. Radiol.*, Vol. 46, pp. 1016–1022, 1973.
8. A.M. Cormack, "Representation of a function by its line integrals with some radiological applications," *J. Appl. Phys.*, Vol. 34, pp. 2722–2727, 1963.
9. E. Seeram, *Computed Tomography: Physical Principles, Clinical Applications, and Quality Control*, Saunders, Philadelphia, 2001.
10. E.K. Fisherman and R.B. Jeffrey (Eds), *Spiral CT: Principles, Techniques and Applications*, Lippincott-Raven, Philadelphia, 1998.
11. T. Fuchs, M. Kachelriess, and W.A. Kalender, "Technical advances in multi-slice spiral CT," *Eur. J. Radiol.*, Vol. 36, pp. 69–73, 2000.
12. W.A. Kalender and M. Prokop, "3D CT angiography," *Crit. Rev. Diagn. Imaging*, Vol. 42, pp. 1–28, 2001.