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College of Engineering & Technology

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Basic principles of CT

POLYCHROMATIC X-RAY BEAMS

All x-ray beam sources for CT and conventional radiography produce x-ray energy that is polychromatic. That is, the x-ray beam comprises photons with varying energies ranging from weak x-ray photons to others that are relatively strong. Low-energy x-ray photons are more readily attenuated by the patient. The detectors cannot differentiate and adjust for differences in attenuation that are caused by low-energy x-ray photons. To the detectors, any x-ray photon that reaches the detector is treated identically, whether it began with high or low energy (Fig. 1-6).

This phenomenon can produce artifacts (objects) seen on the image, which degrade the image quality. Artifacts that result from preferential absorption of the low energy photons, which leaves higher-intensity photons to strike the detector array, are called beam-hardening artifacts. This effect is most obvious when the x-ray beam must first penetrate a dense structure, such as the base of the skull. Beam-hardening artifacts appear as dark streaks or vague areas of decreased density, sometimes called cupping artifacts (Fig. 1-7).

Filtering the x-ray beam with a substance, such as Teflon or aluminum, helps to reduce the range of x-ray energies that reach the patient by eliminating the photons with weaker energies. It makes the x-ray beam more homogeneous. Creating a beam intensity that is more uniform improves the CT image by reducing artifacts. Additionally, filtering the soft (low-energy) photons reduces the radiation dose to the patient.

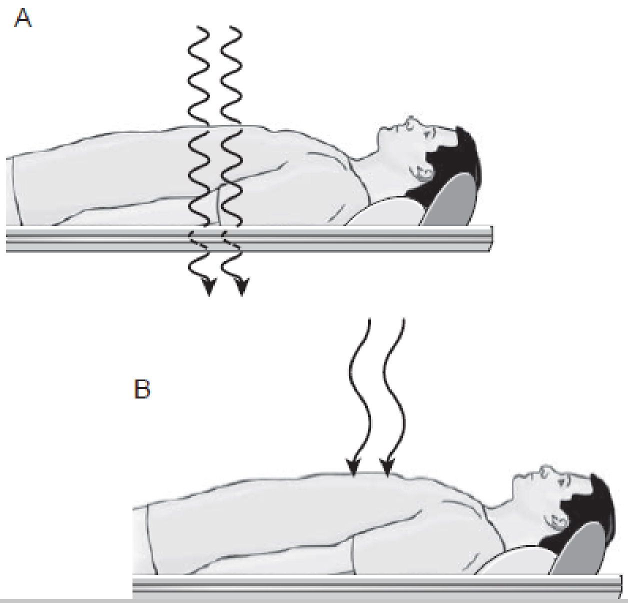


FIGURE 1-6 In (A), the photon has a higher energy and is able to penetrate the object. In (B), the photon is much weaker, and even though the object is identical, it is attenuated. The system is unable to adjust for difference in photon strength; instead, it displays the object in (B) as if it were composed of tissue of a higher density.

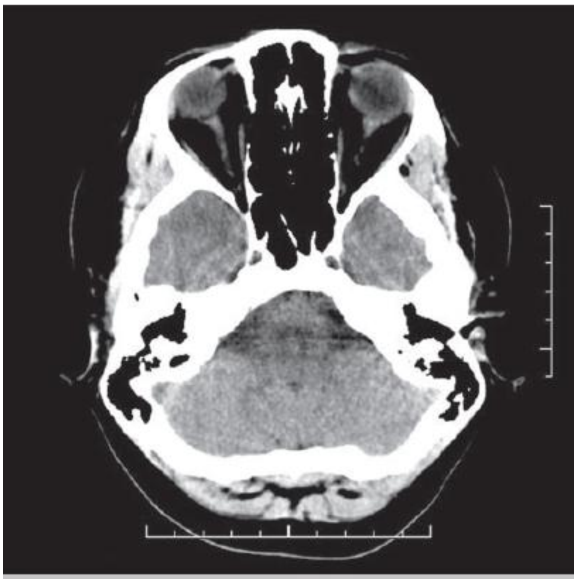


FIGURE 1-7 Streak artifacts arising from the dense petrous bones are a result of beam-hardening.

VOLUME AVERAGING

All CT examinations are performed by obtaining data for a series of slices through a designated area of interest. The nature of the anatomy and the pathology suspected determines how the examination is performed. Scanners allow the technologist to select slice thickness, and these scanners vary in the thickness choices available. In general, the smaller the object being scanned, the thinner the CT slice required. Thicker CT slices increase the likelihood of missing very small objects. For example, if 10-mm slices are created, and the area of pathologic tissue measures just 2mm, normal tissue represents 8mm and is averaged in with the pathologic tissue, potentially making the pathologic tissue less apparent on the image. This process is referred to as volume averaging, or partial volume effect. Therefore, if an area scanned produces images that are suspicious for a mass, but not definitive, creating thinner slices of the same area may be useful.

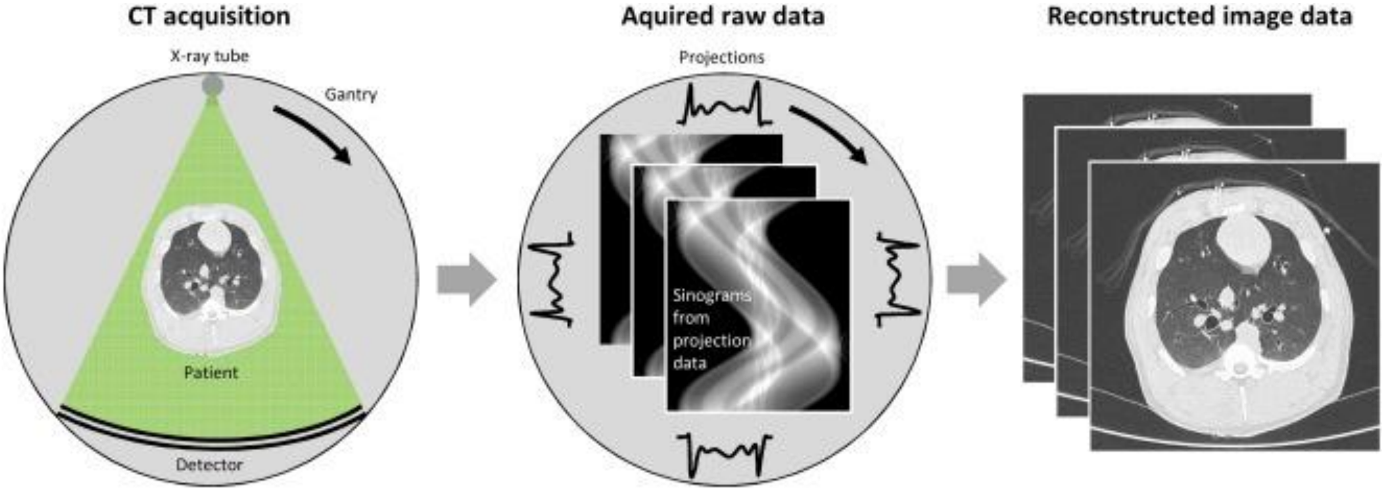
Modern scanners acquire data very quickly and have the capability of creating slices thinner than 1mm. However, thinner slices result in a higher radiation dose to the patient. In addition, if the area to be scanned is large, a huge number of slices are produced. Scanning procedures are designed to provide the image quality necessary for diagnosis at an acceptable radiation dose. Generally, if the structures being investigated are very small (coronary arteries, for example) and the region to be scanned is not extensive (the heart versus the entire abdomen), then slice thickness can be quite thin. Conversely, scan protocols that span a longer anatomic region (such as the abdomen and pelvis) typically use a slice thickness of 5 to 7 mm. In addition, spiral-scanning techniques have allowed options for using data sets to retrospectively adjust the slice thickness when circumstances dictate.

The *Z* axis, which is defined by the slice thickness, has a significant effect on the degree of volume averaging that is present on the image. In addition, the *X* and *Y* dimensions of the pixel also affect the likelihood of volume averaging. The larger the *X* and *Y* dimensions (i.e., the larger the pixel), the more chance that the pixel will contain tissues of different densities. Because the Hounsfield unit of a single pixel is the average of all data measurements within that pixel, this type of averaging can lead to inaccuracies in the image. For example, imagine a pixel that contains equal parts calcium (measuring 600 HU) and lung tissue (measuring −600 HU). The resulting density of the specific pixel is the average of the two tissues, or 0 HU. In this case, the image pixel does not accurately reflect either the calcium or the lung tissue. Using a small pixel size reduces the likelihood of volume averaging by limiting the amount of data to be averaged. Pixel size is determined by the matrix size and the field of view selected for display.

RAW DATA VERSUS IMAGE DATA

All of the thousands of bits of data acquired by the system with each scan are called raw data. The terms scan data and raw data are used interchangeably to refer to computer data waiting to be processed to create an image. Raw data have not yet been sectioned to create pixels; hence, Hounsfield unit values have not yet been assigned. The process of using the raw data to create an image is called image reconstruction. Once raw data have been processed so that each pixel is assigned a Hounsfield unit value, an image can be created; the data included in the image are now referred to as image data.

The reconstruction that is automatically produced during scanning is often called prospective reconstruction. The same raw data may be used later to generate new images. This process is referred to as retrospective reconstruction.



SCAN MODES DEFINED

Step-and-Shoot Scanning

The scanning systems of the 1980s operated exclusively in a “step-and-shoot” mode. In this method 1) the x-ray tube rotated 360° around the patient to acquire data for a single slice, 2) the motion of the x-ray tube was halted while the patient was advanced on the CT table to the location appropriate to collect data for the next slice, and 3) steps one and two were repeated until the desired area was covered. The step-and-shoot method was necessary because the rotation of the x-ray tube entwined the system cables, limiting rotation to 360°. Consequently, gantry motion had to be stopped before the next slice could be taken, this time with the x-ray tube moving in the opposite direction so that the cables would unwind. Although the terms are imprecise, this method is commonly referred to as axial scanning, conventional scanning, or serial scanning.

Helical (Spiral) Scanning

Many technical developments of the 1990s allowed for the development of a continuous acquisition scanning mode most often called spiral or helical scanning. Key among the advances was the development of a system that eliminated the cables and thereby enabled continuous rotation of the gantry. This, in combination with other improvements, allowed for uninterrupted data acquisition that traces a helical path around the patient.

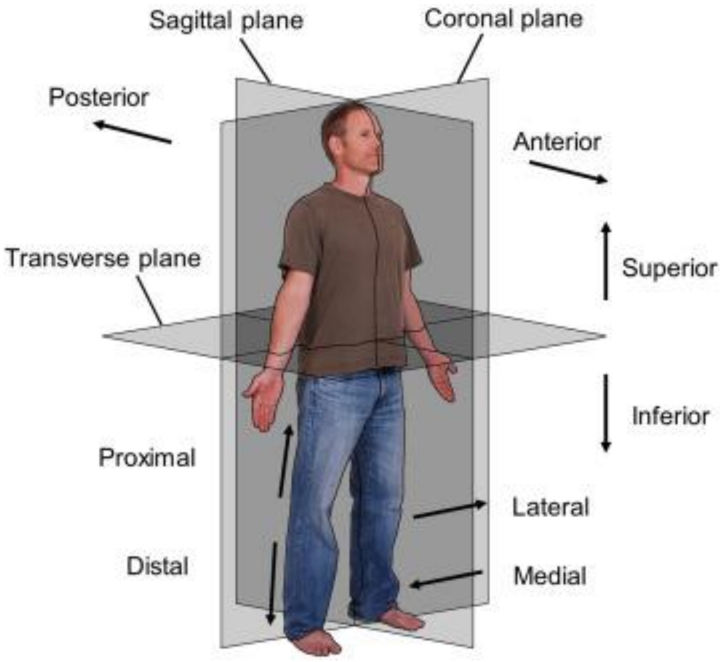
Multidetector Row CT Scanning

The first helical scanners emitted x-rays that were detected by a single row of detectors, yielding one slice per gantry rotation. This technology was expanded on in 1992 when scanners were introduced that contained two rows of detectors, capturing data for two slices per gantry rotation. Further improvements equipped scanners with multiple rows of detectors, allowing data for many slices to be acquired with each gantry rotation. Each of these scanning modes will be explored further in this semester.

IMAGING PLANES

Understanding the intricacies of CT scanning requires familiarity with imaging planes. A brief review of the directional terms used in medicine may also make a discussion of body planes easier to understand. All directional terms are based on the body being viewed in the anatomic position. This position is characterized by an individual standing erect, with the palms of the hands facing forward (Fig. 1-8). This position is used internationally and guarantees uniformity in descriptions of direction. The terms anterior and ventral refer to movement forward (toward the face). Posterior and dorsal are equivalent terms used to describe movement toward the back surface of the body. Inferior refers to movement toward the feet (down) and is synonymous with caudal (toward the tail or, in humans, the feet). Superior defines movement toward the head (up) and is used interchangeably with the term cranial or cephalic. Lateral refers to movement toward the sides of the body. Inversely, medial refers to movement toward the midline of the body.

Figure 1-8 the anatomical position.



The terms distal and proximal are most often used in referring to extremities (limbs). Distal (away from) refers to movement toward the ends. The distal end of the forearm is the end to which the hand is attached. Proximal (close to), which is the opposite of distal, may be defined as situated near the point of attachment. For example, the proximal end of the arm is the end at which it attaches to the shoulder.

To help visualize the imaginary body planes, it is helpful to think of large sheets of glass cutting through the body in various ways. All sheets of glass that are parallel to the floor are called horizontal, or transverse, planes. Those that stand perpendicular to the floor are called vertical, or longitudinal, planes (Fig. 1-9).

A sheet of glass that divides the body into anterior and posterior sections is the coronal plane. The sagittal plane divides the body into right and left sections. The sagittal plane that is located directly in the center, making left and right sections of equal size, is appropriately referred to as the median, or midsagittal, plane. A parasagittal plane is located to either the left or the right of the midline. Axial planes are cross-sectional planes that divide the body into upper and lower sections. Oblique planes are sheets of glass that are slanted and lie at an angle to one of the three standard planes (Fig. 1-10).

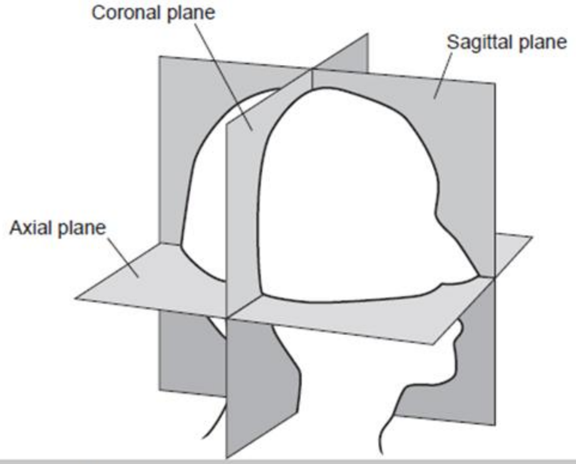


FIGURE 1-9 Imaging planes.

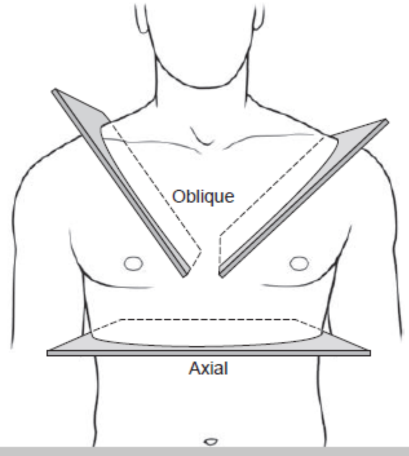


FIGURE 1-10 Oblique planes lie at an angle to one of the standard planes.

Changing the image plane shows the same structures in a new perspective. The loaf of bread analogy can help to explain this change. For example, if a coin is baked within the bread and lies standing on edge in the loaf, a sharp knife cutting through the bread lengthwise will show the coin as a flat, rectangular density. However, if the bread is restacked and cut in an axial plan, the coin appears circular (Fig. 1-11).

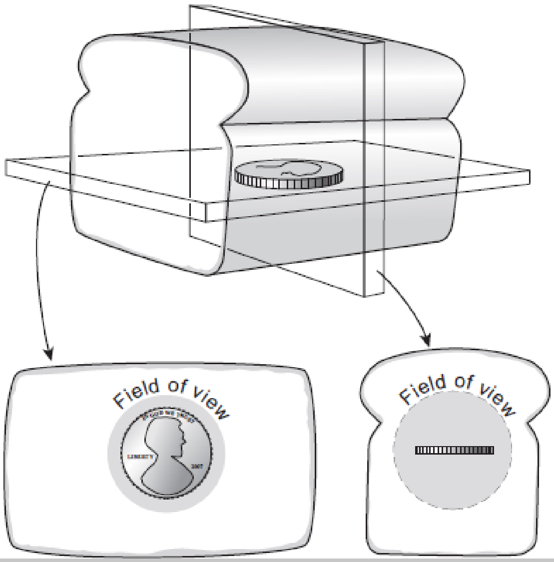


FIGURE 1-11 The imaging plane will affect the way that an area of anatomy is represented on the cross-sectional slice.

The image plan can be adjusted by positioning the patient, gantry, or both to permit scanning in the desired plane or by reformatting the image data. Scanning in the desired planes produces better images than reformatting existing data, although advances in CT technology have reduced the quality difference.

Changing the image plane in CT provides additional information in a fashion similar to the coin within the bread. Changing the image plane from axial to coronal is indicated for two distinct reasons. The primary reason is when the anatomy of interest lies vertically rather than horizontally. The ethmoid sinuses are an example of this principle. Because the ethmoid turbinates lie predominately in the vertical plane, images taken in an axial plane show only sections of the anatomy, with no view of the entire ethmoid complex (Fig. 1-12A). In Figure 1-

12B, the images are taken in the coronal plane, which is more suitable for displaying the ethmoid sinus structures and more readily shows an obstruction.

In the case of the sinuses, it is relatively easy to change the patient's position so that images can be acquired coronally. Obviously, this practice is not possible with all areas of the anatomy that may benefit from coronal imaging, for example, the pelvis. Because fat planes in the pelvis often run obliquely or parallel to the transverse plane, in some cases, images obtained in the coronal plane may be superior to those obtained in the axial plane. However, scanning in the coronal plane is not common because of the difficulty of positioning the pelvis. In this case, reformatting image data from an axial into a coronal plane may prove useful.

The second indication for scanning in a different plane is to reduce artifacts created by surrounding structures. For this reason, the coronal plane is preferred for scanning the pituitary gland. In the axial plane, the number of streak artifacts and the partial volume effect are greater than in the coronal plane.

Most scans are performed in the axial plane, but many head protocols require coronal scans.

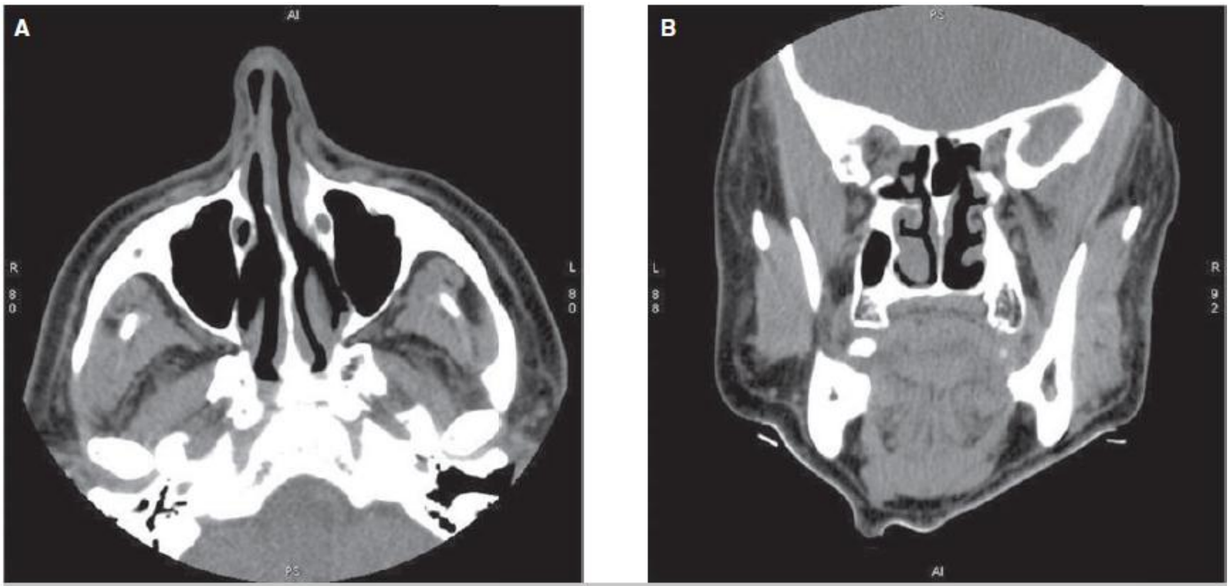
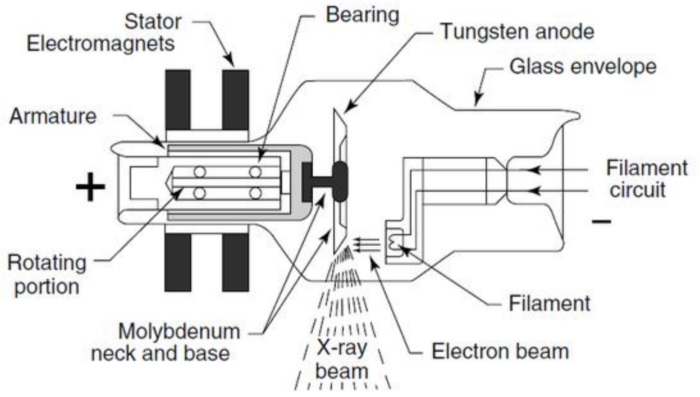


FIGURE 1-12 **A.** A sinus slice taken in the axial plane. **B.** A sinus slice taken in the coronal plane.

OVERVIEW OF CT SYSTEM OPERATION

Scanners vary widely in their mechanical makeup, and the ideal configuration and composition of detectors and tube are hotly debated topics within the industry. X-ray photons are created when fast-moving electrons slam into a metal target. The kinetic energy (the energy of motion) of the electrons is transformed into electromagnetic energy.

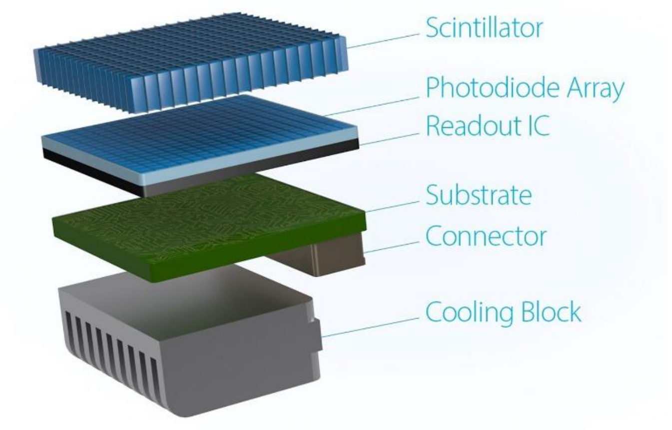
In a CT system, the components that produce x-ray beams are housed in the gantry. The x-ray tube contains filaments that provide the electrons that create x-ray photons. This is accomplished by heating the filament until electrons start to boil off, hovering around the filament in what is known as a space cloud. The generator produces high voltage (or kV) and transmits it to the x-ray tube. This high voltage propels the electrons from the x-ray tube filament to the anode. The area of the anode where the electrons strike and the x-ray beam is produced is the focal spot. The quantity of electrons propelled is controlled by the tube current and is measured in thousandths of an ampere, milliamperes (mA). The electrons then strike the rotating anode target and disarrange the electrons in the target material. The result is the production of heat and x-ray photons. The vast majority (generally more than 99%) of the kinetic energy of the projectile electrons is converted to thermal energy. To spread the heat over a larger area, the target rotates. Increasing the voltage increases the energy with which the electrons strike the target and, hence, increases the intensity of the x-ray beam. The intensity of the x-ray beam is controlled by the kVp setting.

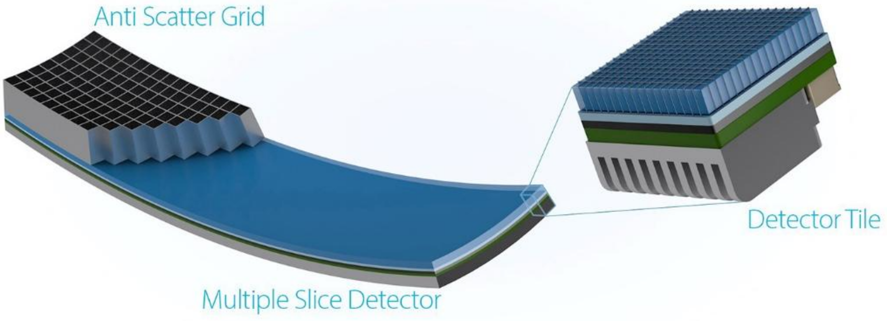


The ability of the tube to withstand the resultant heat is called its heat capacity, whereas its ability to rid itself of the heat is its heat dissipation. The length and frequency of scans are determined in part by the tube’s heat capacity and dissipation rate.

The x-ray photons that pass through the patient strike the detector. If the detector is made from a solid-state scintillator material, the energy of the x-ray photons detected is converted to light. Other elements in the detector, usually a photodiode, convert the light levels into an electric current. On older CT systems, detectors are sometimes of the xenon gas variety. In this case, the striking photon ionizes the xenon gas. These ions are accelerated by the high voltage on the detector plates.

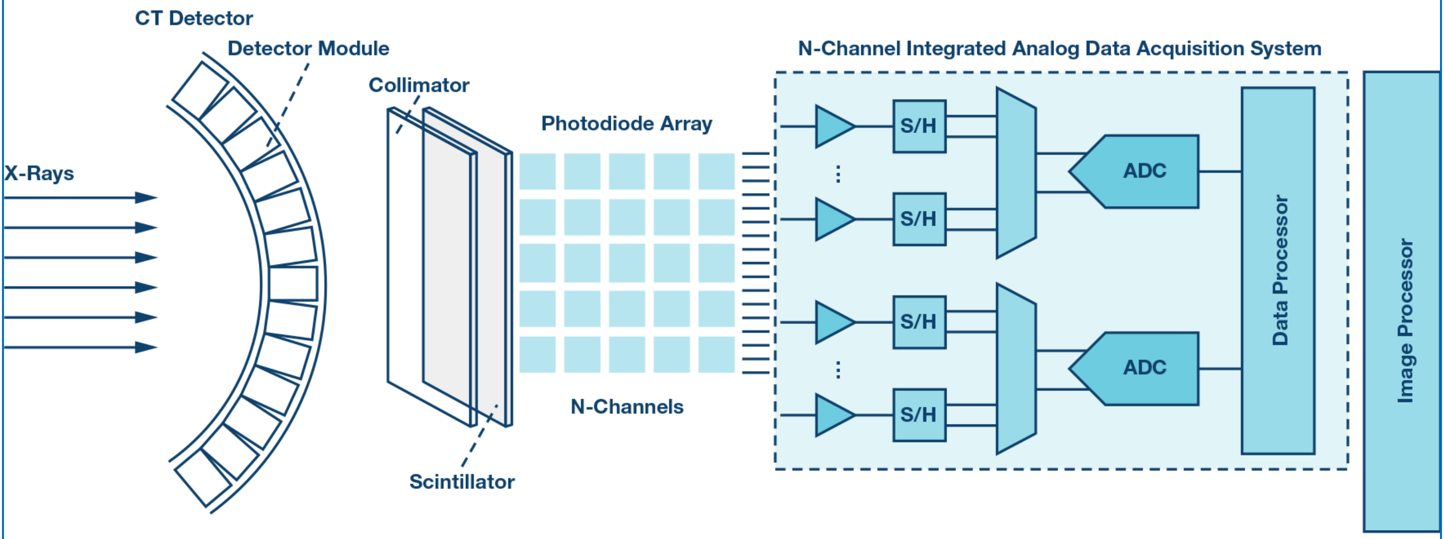
Regardless of the detector material, each detector cell is sampled and converted to a digital format by the data acquisition system (DAS). Each complete sample is called a view. The digital data from the DAS are then transmitted to the central processing unit (CPU). The CPU is often referred to as the brain of the CT scanner.





The reconstruction processor takes the individual views and reconstructs the densities within the slice. To create an image, information from the DAS must be translated into a matrix. To do so, the system assigns each pixel in the matrix one value, or density number. This density number, in Hounsfield units, is the average of all attenuation measurements for that pixel. These digitized data are then sent to a display processor that converts them into shades that can be displayed on a computer monitor.

Although there is wide variation in the design of scanners, they share some characteristics. The CT process can be broken down into three general segments: data acquisition, image reconstruction, and image display. In the first segment, the x-ray photons are created and directed through the patient, where either they are absorbed or they penetrate the patient to strike the CT system's detectors. The goal of this phase is to acquire the information. In the second segment, the data are sorted so that each pixel has one associated Hounsfield value. The goal of this phase is to use the information collected in the previous segment and prepare it for display. In the final phase of creating the CT image, the processed data are converted into shades of gray for viewing. Therefore, one can generalize the phases involved in creating a CT image as 1) obtaining data, 2) using data, and 3) displaying data.



Key concepts:

- The degree to which an x-ray beam is reduced by an object is referred to as *attenuation*.
- To differentiate adjacent objects on a CT image, there must be a density difference between the two objects.
- Hounsfield units quantify the degree that a structure attenuates an x-ray beam.
- The process in CT by which different tissue attenuation values are averaged to produce one less accurate pixel reading is called *volume averaging*.
- X-rays are produced when a substance is bombarded by fast-moving electrons.
- The CT process can be broken down into three main segments:
  Data Acquisition → Get Data
  Image Reconstruction → Use Data
  Image Display → Display Data