

COMPUTED TOMOGRAPHY

Computed tomography (CT) is in its fourth decade of clinical use and has proved invaluable as a diagnostic tool for many clinical applications, from cancer diagnosis to trauma to osteoporosis screening. CT was the first imaging modality that made it possible to probe the inner depths of the body, slice by slice. Since 1972, when the first head CT scanner was introduced, CT has matured greatly and gained technological sophistication. Concomitant changes have occurred in the quality of CT images. The first CT scanner, an EMI Mark 1, produced images with 80×80 pixel resolution (3-mm pixels), and each pair of slices required approximately 4.5 minutes of scan time and 1.5 minutes of reconstruction time. Because of the long acquisition times required for the early scanners and the constraints of cardiac and respiratory motion, it was originally thought that CT would be practical only for head scans.

CT is one of the many technologies that was made possible by the invention of the computer. The clinical potential of CT became obvious during its early clinical use, and the excitement forever solidified the role of computers in medical imaging. Recent advances in acquisition geometry, detector technology, multiple detector arrays, and x-ray tube design have led to scan times now measured in fractions of a second. Modern computers deliver computational power that allows reconstruction of the image data essentially in real time.

The invention of the CT scanner earned Godfrey Hounsfield of Britain and Allan Cormack of the United States the Nobel Prize for Medicine in 1979. CT scanner technology today is used not only in medicine but in many other industrial applications, such as nondestructive testing and soil core analysis.

13.1 BASIC PRINCIPLES

The mathematical principles of CT were first developed by Radon in 1917. Radon's treatise proved that an image of an unknown object could be produced if one had an infinite number of projections through the object. Although the mathematical details are beyond the scope of this text, we can understand the basic idea behind tomographic imaging with an example taken from radiography.

With plain film imaging, the three-dimensional (3D) anatomy of the patient is reduced to a two-dimensional (2D) projection image. The density at a given point on an image represents the x-ray attenuation properties within the patient along a line between the x-ray focal spot and the point on the detector corresponding to the

point on the image. Consequently, with a conventional radiograph of the patient's anatomy, information with respect to the dimension parallel to the x-ray beam is lost. This limitation can be overcome, at least for obvious structures, by acquiring both a posteroanterior (PA) projection and a lateral projection of the patient. For example, the PA chest image yields information concerning height and width, integrated along the depth of the patient, and the lateral projection provides information about the height and depth of the patient, integrated over the width dimension (Fig. 13-1). For objects that can be identified in both images, such as a pulmonary nodule on PA and lateral chest radiographs, the two films provide valuable location information. For more complex or subtle pathology, however, the two projections are not sufficient. Imagine that instead of just two projections, a series of 360 radiographs were acquired at 1-degree angular intervals around the patient's

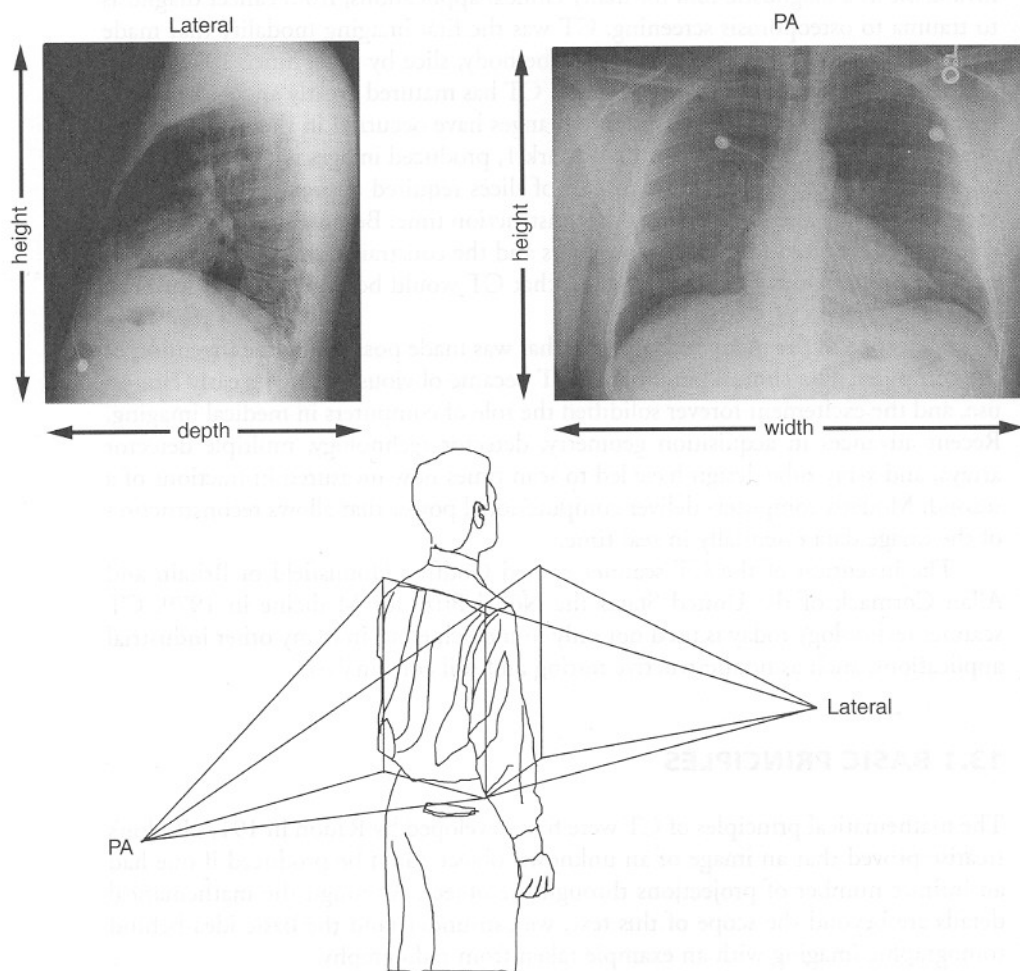


FIGURE 13-1. Posteroanterior and lateral chest radiographs give three-dimensional information concerning the location of an abnormality. In computed tomography, the two views shown here are extended to almost 1,000 views, and with appropriate computer processing true three-dimensional information concerning the patient's anatomy can be displayed.

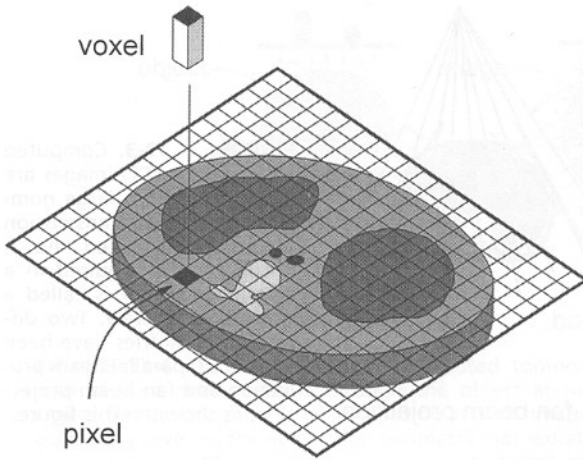


FIGURE 13-2. A pixel (picture element) is the basic two-dimensional element of a digital image. Computed tomographic (CT) images are typically square arrays containing 512×512 pixels, each pixel representing 4,096 possible shades of gray (12 bits). Each pixel in the CT image corresponds to a voxel (volume element) in the patient. The voxel has two dimensions equal to the pixel in the plane of the image, and the third dimension represents the slice thickness of the CT scan.

thoracic cavity. Such a set of images provides essentially the same data as a thoracic CT scan. However, the 360 radiographic images display the anatomic information in a way that would be impossible for a human to visualize: cross-sectional images. If these 360 images were stored into a computer, the computer could in principle reformat the data and generate a complete thoracic CT examination.

The tomographic image is a picture of a slab of the patient's anatomy. The 2D CT image corresponds to a 3D section of the patient, so that even with CT, three dimensions are compressed into two. However, unlike the case with plain film imaging, the CT slice-thickness is very thin (1 to 10 mm) and is approximately uniform. The 2D array of pixels (short for *picture elements*) in the CT image corresponds to an equal number of 3D voxels (*volume elements*) in the patient. Voxels have the same in-plane dimensions as pixels, but they also include the slice thickness dimension. Each pixel on the CT image displays the average x-ray attenuation properties of the tissue in the corresponding voxel (Fig. 13-2).

Tomographic Acquisition

A single transmission measurement through the patient made by a single detector at a given moment in time is called a *ray*. A series of rays that pass through the patient at the same orientation is called a *projection* or *view*. There are two projection geometries that have been used in CT imaging (Fig. 13-3). The more basic type is *parallel beam geometry*, in which all of the rays in a projection are parallel to each other. In *fan beam geometry*, the rays at a given projection angle diverge and have the appearance of a fan. All modern CT scanners incorporate fan beam geometry in the acquisition and reconstruction process. The purpose of the CT scanner hardware is to acquire a large number of transmission measurements through the patient at different positions. The acquisition of a single axial CT image may involve approximately 800 rays taken at 1,000 different projection angles, for a total of approximately 800,000 transmission measurements. Before the axial acquisition of the next slice, the table that the patient is lying on is moved slightly in the cranial-caudal direction (the "z-axis" of the scanner), which positions a different slice of tissue in the path of the x-ray beam for the acquisition of the next image.

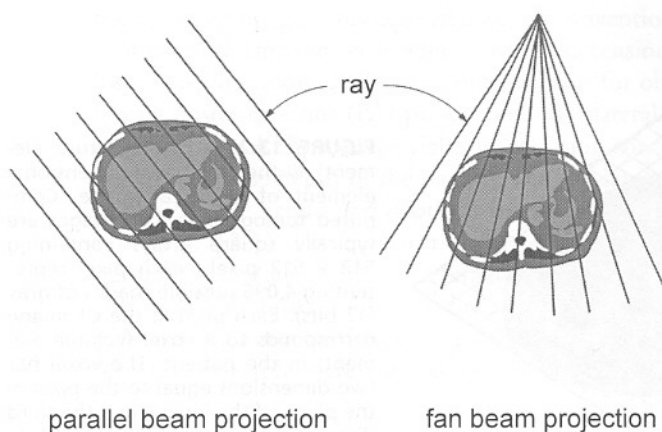


FIGURE 13-3. Computed tomographic (CT) images are produced from a large number of x-ray transmission measurements called rays. A group of rays acquired in a certain geometry is called a *projection* or *view*. Two different geometries have been used in CT, parallel beam projection and fan beam projection, as shown in this figure.

Tomographic Reconstruction

Each ray that is acquired in CT is a transmission measurement through the patient along a line, where the detector measures an x-ray intensity, I_t . The unattenuated intensity of the x-ray beam is also measured during the scan by a reference detector, and this detects an x-ray intensity I_o . The relationship between I_t and I_o is given by the following equation:

$$I_t = I_o e^{-\mu t}$$

where t is the thickness of the patient along the ray and μ is the average linear attenuation coefficient along the ray. Notice that I_t and I_o are machine-dependent values, but the product μt is an important parameter relating to the anatomy of the patient along a given ray. When the equation is rearranged, the measured values I_t and I_o can be used to calculate the parameter of interest:

$$\ln(I_o/I_t) = \mu t$$

where $\ln$ is the natural logarithm (to base e , $e = 2.78 \dots$), t ultimately cancels out, and the value μ for each ray is used in the CT reconstruction algorithm. This computation, which is a preprocessing step performed before image reconstruction, reduces the dependency of the CT image on the machine-dependent parameters, resulting in an image that depends primarily on the patient's anatomic characteristics. This is very much a desirable aspect of imaging in general, and the high clinical utility of CT results, in part, from this feature. By comparison, if a screen-film radiograph is underexposed (I_o is too low) it appears too white, and if it is overexposed (I_o too high) it appears too dark. The density of CT images is independent of I_o , although the noise in the image is affected.

After preprocessing of the raw data, a *CT reconstruction algorithm* is used to produce the CT images. There are numerous reconstruction strategies; however, *filtered backprojection* reconstruction is most widely used in clinical CT scanners. The backprojection method builds up the CT image in the computer by essentially reversing the acquisition steps (Fig. 13-4). During acquisition, attenuation information along a known path of the narrow x-ray beam is integrated by a detector. During backprojection reconstruction, the μ value for each ray is smeared along this same path in the image of the patient. As the data from a large number of rays are

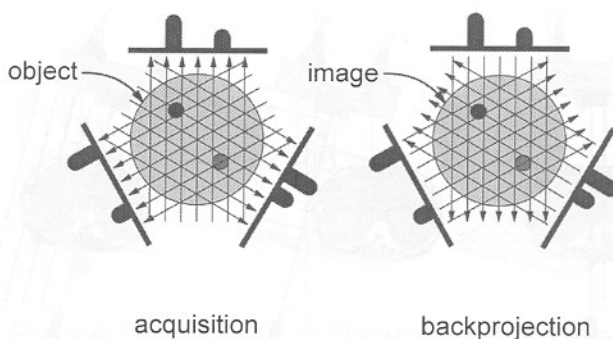


FIGURE 13-4. Data acquisition in computed tomography (CT) involves making transmission measurements through the object at numerous angles around the object (**left**). The process of computing the CT image from the acquisition data essentially reverses the acquisition geometry mathematically (**right**). Each transmission measurement is backprojected onto a digital matrix. After backprojection, areas of high attenuation are positively reinforced through the backprojection process whereas other areas are not, and thus the image is built up from the large collection of rays passing through it.

backprojected onto the image matrix, areas of high attenuation tend to reinforce each other, and areas of low attenuation also reinforce, building up the image in the computer.

13.2 GEOMETRY AND HISTORICAL DEVELOPMENT

Part of understanding an imaging modality involves appreciation of the maturation of that modality. In addition to historical interest, the discussion of the evolution of CT scanners also allows the presentation and reinforcement of several key concepts in CT imaging.

First Generation: Rotate/Translate, Pencil Beam

CT scanners represent a marriage of diverse technologies, including computer hardware, motor control systems, x-ray detectors, sophisticated reconstruction algorithms, and x-ray tube/generator systems. The first generation of CT scanners employed a rotate/translate, pencil beam system (Fig. 13-5). Only two x-ray detectors were used, and they measured the transmission of x-rays through the patient for two different slices. The acquisition of the numerous projections and the multiple rays per projection required that the single detector for each CT slice be physically moved throughout all the necessary positions. This system used parallel ray geometry. Starting at a particular angle, the x-ray tube and detector system translated linearly across the field of view (FOV), acquiring 160 parallel rays across a 24-cm FOV. When the x-ray tube/detector system completed its translation, the whole system was rotated slightly, and then another translation was used to acquire the 160 rays in the next projection. This procedure was repeated until 180 projections were acquired at 1-degree intervals. A total of $180 \times 160 = 28,800$ rays were measured.

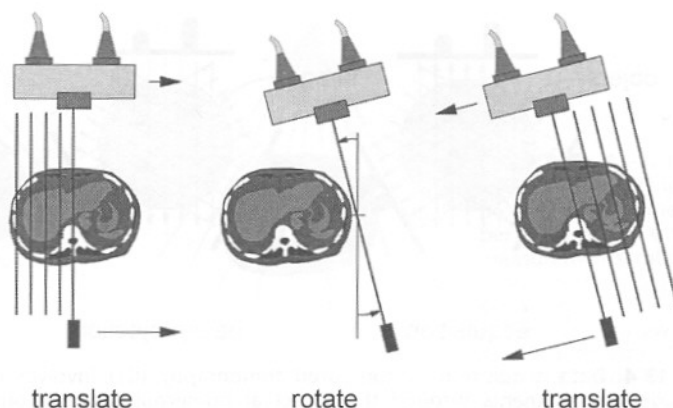


FIGURE 13-5. First-generation (rotate/translate) computed tomography (CT). The x-ray tube and a single detector (per CT slice) translate across the field of view, producing a series of parallel rays. The system then rotates slightly and translates back across the field of view, producing ray measurements at a different angle. This process is repeated at 1-degree intervals over 180 degrees, resulting in the complete CT data set.

As the system translated and measured rays from the thickest part of the head to the area adjacent to the head, a huge change in x-ray flux occurred. The early detector systems could not accommodate this large change in signal, and consequently the patient's head was pressed into a flexible membrane surrounded by a water bath. The water bath acted to bolus the x-rays so that the intensity of the x-ray beam outside the patient's head was similar in intensity to that inside the head. The NaI detector also had a significant amount of "afterglow," meaning that the signal from a measurement taken at one period of time decayed slowly and carried over into the next measurement if the measurements were made temporally too close together.

One advantage of the first-generation CT scanner was that it employed pencil beam geometry—only two detectors measured the transmission of x-rays through the patient. The pencil beam allowed very efficient scatter reduction, because scatter that was deflected away from the pencil ray was not measured by a detector. With regard to scatter rejection, the pencil beam geometry used in first-generation CT scanners was the best.

Second Generation: Rotate/Translate, Narrow Fan Beam

The next incremental improvement to the CT scanner was the incorporation of a linear array of 30 detectors. This increased the utilization of the x-ray beam by 30 times, compared with the single detector used per slice in first-generation systems. A relatively narrow fan angle of 10 degrees was used. In principle, a reduction in scan time of about 30-fold could be expected. However, this reduction time was not realized, because more data ($600 \text{ rays} \times 540 \text{ views} = 324,000 \text{ data points}$) were acquired to improve image quality. The shortest scan time with a second-generation scanner was 18 seconds per slice, 15 times faster than with the first-generation system.

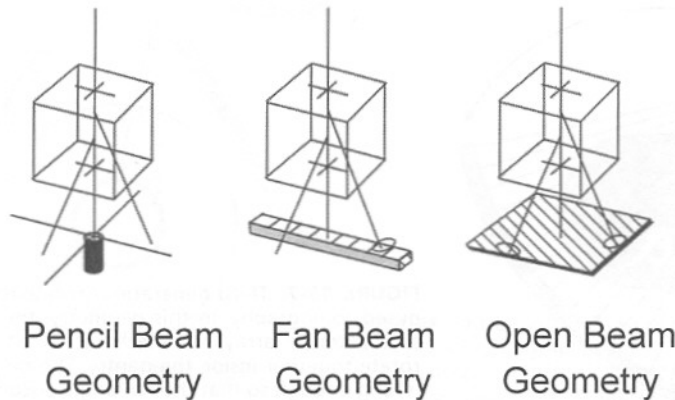


FIGURE 13-6. Pencil beam geometry makes inefficient use of the x-ray source, but it provides excellent x-ray scatter rejection. X-rays that are scattered away from the primary pencil beam do not strike the detector and are not measured. Fan beam geometry makes use of a linear x-ray detector and a divergent fan beam of x-rays. X-rays that are scattered in the same plane as the detector can be detected, but x-rays that are scattered out of plane miss the linear detector array and are not detected. Scattered radiation accounts for approximately 5% of the signal in typical fan beam scanners. Open beam geometry, which is used in projection radiography, results in the highest detection of scatter. Depending on the dimensions and the x-ray energy used, open beam geometries can lead to four detected scatter events for every detected primary photon ($s/p=4$).

Incorporating an array of detectors, instead of just two, required the use of a narrow fan beam of radiation. Although a narrow fan beam provides excellent scatter rejection compared with plain film imaging, it does allow more scattered radiation to be detected than was the case with the pencil beam used in first-generation CT. The difference between pencil beam, fan beam, and open beam geometry in terms of scatter detection is illustrated in Fig. 13-6.

Third Generation: Rotate/Rotate, Wide Fan Beam

The translational motion of first- and second-generation CT scanners was a fundamental impediment to fast scanning. At the end of each translation, the motion of the x-ray tube/detector system had to be stopped, the whole system rotated, and the translational motion restarted. The success of CT as a clinical modality in its infancy gave manufacturers reason to explore more efficient, but more costly, approaches to the scanning geometry.

The number of detectors used in third-generation scanners was increased substantially (to more than 800 detectors), and the angle of the fan beam was increased so that the detector array formed an arc wide enough to allow the x-ray beam to interrogate the entire patient (Fig. 13-7). Because detectors and the associated electronics are expensive, this led to more expensive CT scanners. However, spanning the dimensions of the patient with an entire row of detectors eliminated the need for translational motion. The multiple detectors in the detector array capture the same number of ray measurements in one instant as was required by a complete

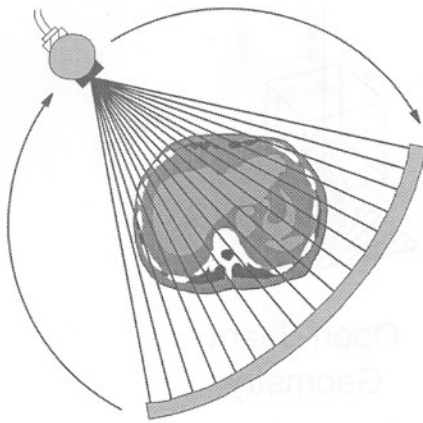


FIGURE 13-7. Third-generation (rotate/rotate) computed tomography. In this geometry, the x-ray tube and detector array are mechanically attached and rotate together inside the gantry. The detector array is long enough so that the fan angle encompasses the entire width of the patient.

translation in the earlier scanner systems. The mechanically joined x-ray tube and detector array rotate together around the patient without translation. The motion of third-generation CT is “rotate/rotate,” referring to the rotation of the x-ray tube and the rotation of the detector array. By elimination of the translational motion, the scan time is reduced substantially. The early third-generation scanners could deliver scan times shorter than 5 seconds. Newer systems have scan times of one-half second.

The evolution from first- to second- and second- to third-generation scanners involved radical improvement with each step. Developments of the fourth- and fifth-generation scanners led not only to some improvements but also to some compromises in clinical CT images, compared with third-generation scanners. Indeed, rotate/rotate scanners are still as viable today as they were when they were introduced in 1975. The features of third- and fourth-generation CT should be compared by the reader, because each offers some benefits but also some trade-offs.

Fourth Generation: Rotate/Stationary

Third-generation scanners suffered from the significant problem of ring artifacts, and in the late 1970s fourth-generation scanners were designed specifically to address these artifacts. It is never possible to have a large number of detectors in perfect balance with each other, and this was especially true 25 years ago. Each detector and its associated electronics has a certain amount of drift, causing the signal levels from each detector to shift over time. The rotate/rotate geometry of third-generation scanners leads to a situation in which each detector is responsible for the data corresponding to a ring in the image (Fig. 13-8). Detectors toward the center of the detector array provide data in the reconstructed image in a ring that is small in diameter, and more peripheral detectors contribute to larger diameter rings.

Third-generation CT uses a fan geometry in which the vertex of the fan is the x-ray focal spot and the rays fan out from the x-ray source to each detector on the detector array. The detectors toward the center of the array make the transmission measurement I_t , while the reference detector that measures I_0 is positioned near the edge of the detector array. If g_1 is the gain of the reference detector, and g_2 is the

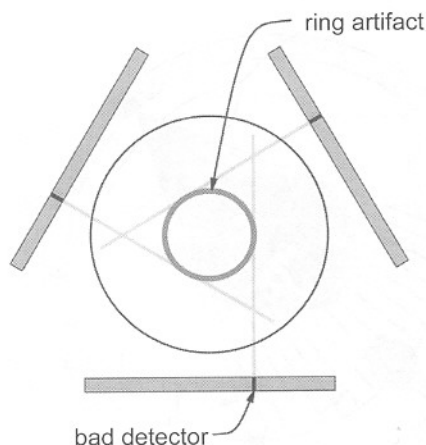


FIGURE 13-8. With third-generation geometry in computed tomography, each individual detector gives rise to an annulus (ring) of image information. When a detector becomes miscalibrated, the tainted data can lead to ring artifacts in the reconstructed image.

gain of the other detector, then the transmission measurement is given by the following equation:

$$\ln(g_1 I_0 / g_2 I_t) = \mu t$$

The equation is true only if the gain terms cancel each other out, and that happens when $g_1 = g_2$. If there is electronic drift in one or both of the detectors, then the gain changes between detectors, so that $g_1 \neq g_2$. So, for third-generation scanners, even a slight imbalance between detectors affects the μt values that are back-projected to produce the CT image, causing the ring artifacts.

Fourth-generation CT scanners were designed to overcome the problem of ring artifacts. With fourth-generation scanners, the detectors are removed from the rotating gantry and are placed in a stationary 360-degree ring around the patient (Fig. 13-9), requiring many more detectors. Modern fourth-generation CT systems use about 4,800 individual detectors. Because the x-ray tube rotates and the detectors are stationary, fourth-generation CT is said to use a rotate/stationary geometry. During acquisition with a fourth-generation scanner, the divergent x-ray beam

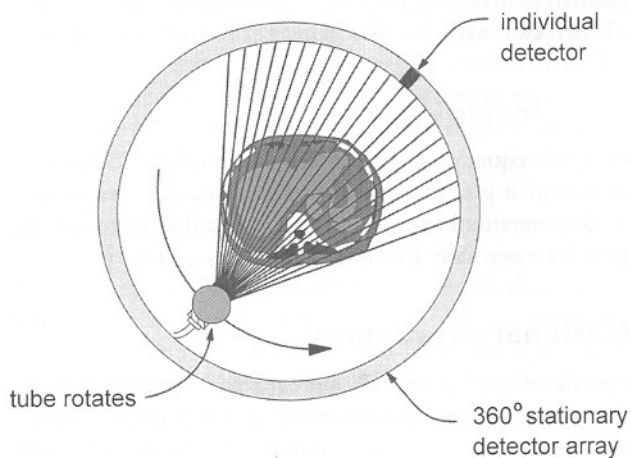


FIGURE 13-9. Fourth-generation (rotate/stationary) computed tomography (CT). The x-ray tube rotates within a complete circular array of detectors, which are stationary. This design requires about six times more individual detectors than a third-generation CT scanner does. At any point during the scan, a divergent fan of x-rays is detected by a group of x-ray detectors.

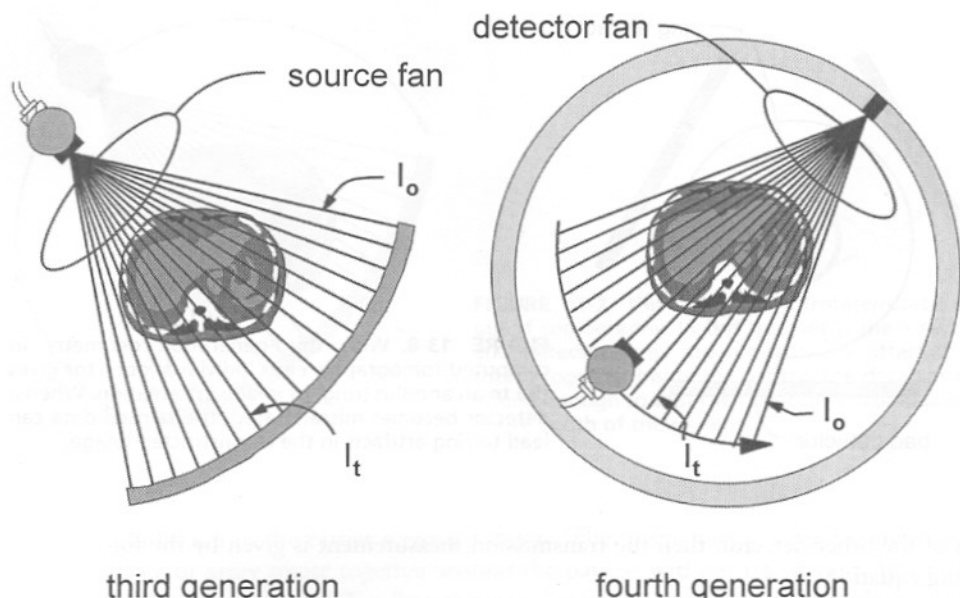


FIGURE 13-10. The fan beam geometry in third-generation computed tomography uses the x-ray tube as the apex of the fan (source fan). Fourth-generation scanners normalize the data acquired during the scan so that the apex of the fan is an individual detector (detector fan). With third-generation scanners, the detectors near the edge of the detector array measure the reference x-ray beam. With fourth-generation scanners, the reference beam is measured by the same detector used for the transmission measurement.

emerging from the x-ray tube forms a fan-shaped x-ray beam. However, the data are processed for fan beam reconstruction with each detector as the vertex of a fan, the rays acquired by each detector being fanned out to different positions of the x-ray source. In the vernacular of CT, third-generation design uses a *source fan*, whereas fourth-generation uses a *detector fan*. The third-generation fan data are acquired by the detector array simultaneously, in one instant of time. The fourth-generation fan beam data are acquired by a single detector over the period of time that is required for the x-ray tube to rotate through the arc angle of the fan. This difference is illustrated in Fig. 13-10. With fourth-generation geometry, each detector acts as its own reference detector. For each detector with its own gain, g , the transmission measurement is calculated as follows:

$$\ln(g I_o / g I_t) = \mu t$$

Note that the single g term in this equation is guaranteed to cancel out. Therefore, ring artifacts are eliminated in fourth-generation scanners. It should be mentioned, however, that with modern detectors and more sophisticated calibration software, third-generation CT scanners are essentially free of ring artifacts as well.

Fifth Generation: Stationary/Stationary

A novel CT scanner has been developed specifically for cardiac tomographic imaging. This "cine-CT" scanner does not use a conventional x-ray tube; instead, a large arc of tungsten encircles the patient and lies directly opposite to the detector ring.

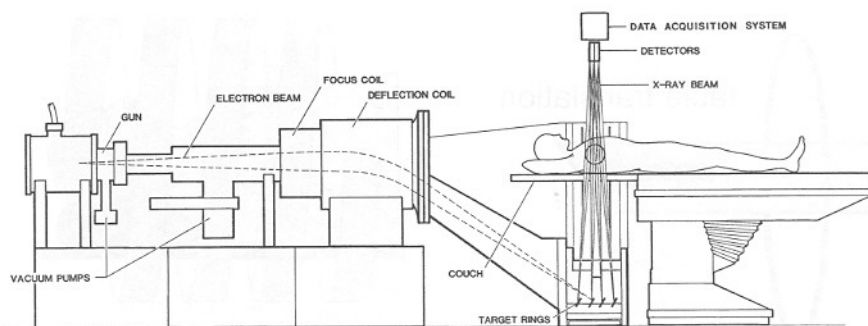


FIGURE 13-11. A schematic diagram of the fifth-generation, Imatron (South San Francisco, CA) cine-CT scanner is shown.

X-rays are produced from the focal track as a high-energy electron beam strikes the tungsten. There are no moving parts to this scanner gantry. The electron beam is produced in a cone-like structure (a vacuum enclosure) behind the gantry and is electronically steered around the patient so that it strikes the annular tungsten target (Fig. 13-11). Cine-CT systems, also called *electron beam scanners*, are marketed primarily to cardiologists. They are capable of 50-msec scan times and can produce fast-frame-rate CT movies of the beating heart.

Sixth Generation: Helical

Third-generation and fourth-generation CT geometries solved the mechanical inertia limitations involved in acquisition of the individual projection data by eliminating the translation motion used in first- and second-generation scanners. However, the gantry had to be stopped after each slice was acquired, because the detectors (in third-generation scanners) and the x-ray tube (in third- and fourth-generation machines) had to be connected by wires to the stationary scanner electronics. The ribbon cable used to connect the third-generation detectors with the electronics had to be carefully rolled out from a cable spool as the gantry rotated, and then as the gantry stopped and began to rotate in the opposite direction the ribbon cable had to be retracted. In the early 1990s, the design of third- and fourth-generation scanners evolved to incorporate *slip ring* technology. A slip ring is a circular contact with sliding brushes that allows the gantry to rotate continually, untethered by wires. The use of slip-ring technology eliminated the inertial limitations at the end of each slice acquisition, and the rotating gantry was free to rotate continuously throughout the entire patient examination. This design made it possible to achieve greater rotational velocities than with systems not using a slip ring, allowing shorter scan times.

Helical CT (also inaccurately called *spiral CT*) scanners acquire data while the table is moving; as a result, the x-ray source moves in a helical pattern around the patient being scanned. Helical CT scanners use either third- or fourth-generation slip-ring designs. By avoiding the time required to translate the patient table, the total scan time required to image the patient can be much shorter (e.g., 30 seconds for the entire abdomen). Consequently, helical scanning allows the use of less contrast agent and increases patient throughput. In some instances the entire scan can be performed within a single breath-hold of the patient, avoiding inconsistent levels of inspiration.

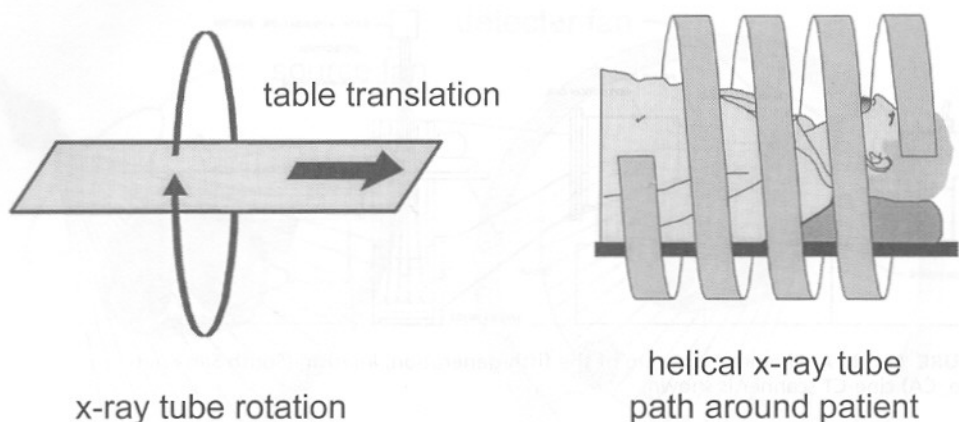


FIGURE 13-12. With helical computed tomographic scanners, the x-ray tube rotates around the patient while the patient and the table are translated through the gantry. The net effect of these two motions results in the x-ray tube traveling in a helical path around the patient.

The advent of helical scanning has introduced many different considerations for data acquisition. In order to produce reconstructions of *planar* sections of the patient, the raw data from the helical data set are interpolated to approximate the acquisition of planar reconstruction data (Fig. 13-12). The speed of the table motion relative to the rotation of the CT gantry is a very important consideration, and the *pitch* is the parameter that describes this relationship (discussed later).

Seventh Generation: Multiple Detector Array

X-ray tubes designed for CT have impressive heat storage and cooling capabilities, although the instantaneous production of x-rays (i.e., x-rays per milliamper-second [mAs]) is constrained by the physics governing x-ray production. An approach to overcoming x-ray tube output limitations is to make better use of the x-rays that are produced by the x-ray tube. When multiple detector arrays are used (Fig. 13-13), the collimator spacing is wider and therefore more of the x-rays that are produced by the x-ray tube are used in producing image data. With conventional, single detector array scanners, opening up the collimator increases the slice thickness, which is good for improving the utilization of the x-ray beam but reduces spatial resolution in the slice thickness dimension. With the introduction of multiple detector arrays, the slice thickness is determined by the detector size and not by the collimator. This represents a major shift in CT technology.

For example, a multiple detector array CT scanner may operate with four contiguous, 5-mm detector arrays and 20-mm collimator spacing. For the same technique (kilovoltage [kV] and mAs), the number of x-rays being detected is four times that of a single detector array with 5-mm collimation. Furthermore, the data set from the 4×5 mm multiple detector array can be used to produce true 5-mm slices, or data from adjacent arrays can be added to produce true 10-, 15-, or even 20-mm slices, all from the same acquisition. The flexibility of CT acquisition protocols and increased efficiency resulting from multiple detector array CT scanners allows better patient imaging; however, the number of parameters involved in the CT acqui-

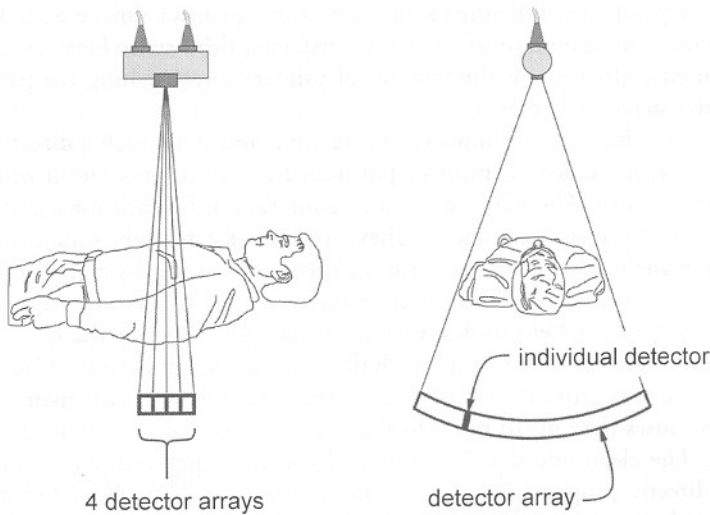


FIGURE 13-13. Multiple detector array computed tomographic (CT) scanners use several, closely spaced, complete detector arrays. With no table translation (nonhelical acquisition), each detector array acquires a separate axial CT image. With helical acquisition on a multiple detector array system, table speed and detector pitch can be increased, increasing the coverage for a given period of time.

sition protocol is increased as well. Also with multiple detector arrays, the notion of helical pitch needs to be redefined. This topic is discussed later in the chapter.

13.3 DETECTORS AND DETECTOR ARRAYS

Xenon Detectors

Xenon detectors use high-pressure (about 25 atm) nonradioactive xenon gas, in long thin cells between two metal plates (Fig. 13-14). Although a gaseous detector does not have the same detection efficiency as a solid one, the detector can be made very thick (e.g., 6 cm) to compensate in part for its relatively low density. The metal

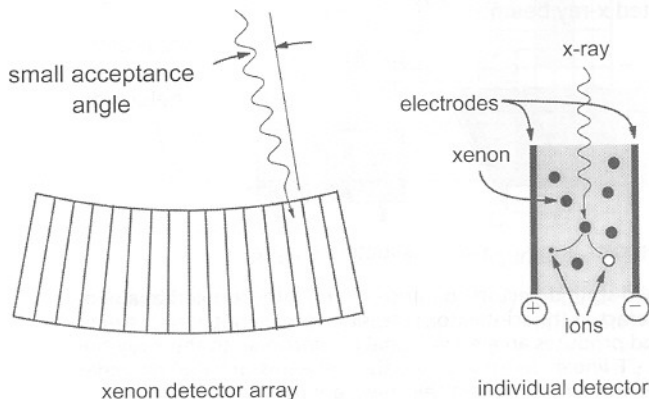


FIGURE 13-14. Xenon detector arrays are a series of highly directional xenon-filled ionization chambers. As x-rays ionize xenon atoms, the charged ions are collected as electric current at the electrodes. The current is proportional to the x-ray fluence.

septa that separate the individual xenon detectors can also be made quite thin, and this improves the geometric efficiency by reducing dead space between detectors. The geometric efficiency is the fraction of primary x-rays exiting the patient that strike active detector elements.

The long, thin ionization plates of a xenon detector are highly directional. For this reason, xenon detectors must be positioned in a fixed orientation with respect to the x-ray source. Therefore, xenon detectors cannot be used for fourth-generation scanners, because those detectors have to record x-rays as the source moves over a very wide angle. Xenon detectors can be used only for third-generation systems.

Xenon detectors for CT are ionization detectors—a gaseous volume is surrounded by two metal electrodes, with a voltage applied across the two electrodes (see Chapter 20). As x-rays interact with the xenon atoms and cause ionization (positive atoms and negative electrons), the electric field (volts per centimeter) between the plates causes the ions to move to the electrodes, where the electronic charge is collected. The electronic signal is amplified and then digitized, and its numerical value is directly proportional to the x-ray intensity striking the detector. Xenon detector technology has been surpassed by solid-state detectors, and its use is now relegated to inexpensive CT scanners.

Solid-State Detectors

A solid-state CT detector is composed of a scintillator coupled tightly to a photodiode. The scintillator emits visible light when it is struck by x-rays, just as in an x-ray intensifying screen. The light emitted by the scintillator reaches the photodiode, typically a photodiode, which is an electronic device that converts light intensity into an electrical signal proportional to the light intensity (Fig. 13-15). This scintillator-photodiode design of solid-state CT detectors is very similar in concept to many digital radiographic x-ray detector systems; however, the performance requirements of CT are slightly different. The detector size in CT is measured in millimeters (typically 1.0×15 mm or 1.0×1.5 mm for multiple detector array scanners), whereas detector elements in digital radiography systems are typi-

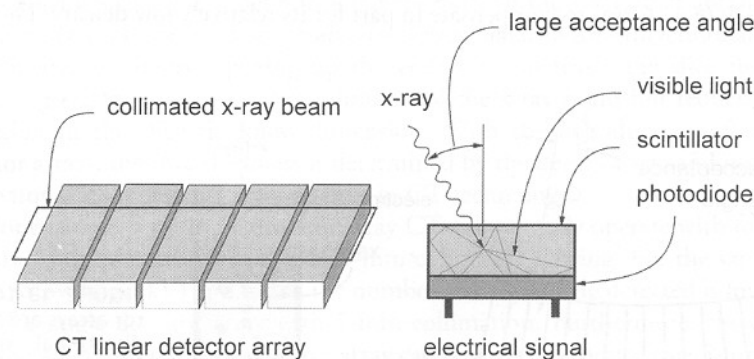


FIGURE 13-15. Solid-state detectors comprise a scintillator coupled to a photodiode. X-rays interact in the scintillator, releasing visible light, which strikes the photodiode and produces an electric signal proportional to the x-ray fluence. For a single CT linear detector array, the detectors are slightly wider than the widest collimated x-ray beam thickness, approximately 12 mm.

cally 0.10 to 0.20 mm on each side. CT requires a very high-fidelity, low-noise signal, typically digitized to 20 or more bits.

The scintillator used in solid-state CT detectors varies among manufacturers, with CdWO_4 , yttrium and gadolinium ceramics, and other materials being used. Because the density and effective atomic number of scintillators are substantially higher than those of pressurized xenon gas, solid-state detectors typically have better x-ray absorption efficiency. However, to reduce crosstalk between adjacent detector elements, a small gap between detector elements is necessary, and this reduces the geometric efficiency somewhat.

The top surface of a solid-state CT detector is essentially flat and therefore is capable of x-ray detection over a wide range of angles, unlike the xenon detector. Solid-state detectors are required for fourth-generation CT scanners, and they are used in most high-tier third-generation scanners as well.

Multiple Detector Arrays

Multiple detector arrays are a set of several linear detector arrays, tightly abutted (Fig. 13-16). The multiple detector array is an assembly of multiple solid-state

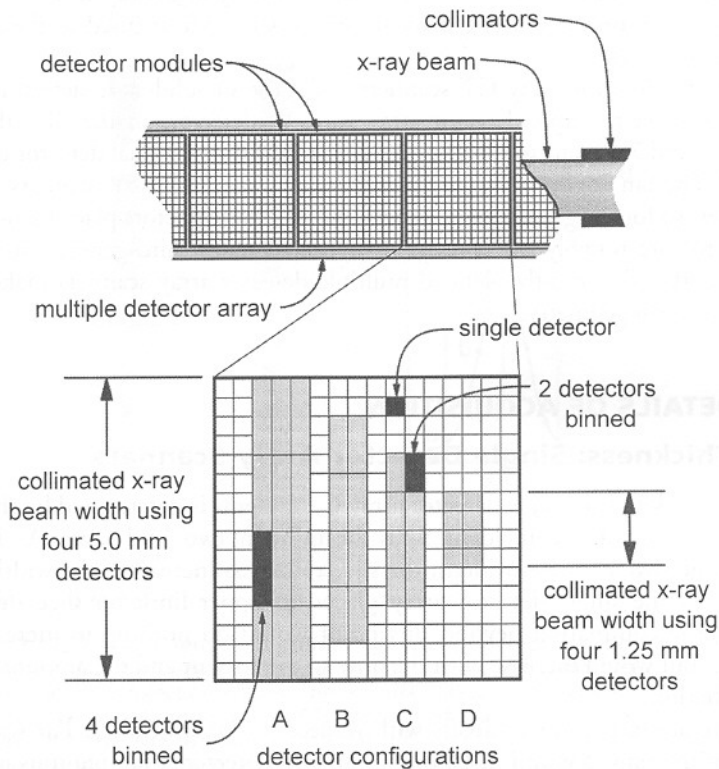


FIGURE 13-16. A multiple detector array composed of multiple detector modules is illustrated in the top panel. In the bottom panel, four different detector configurations are shown, (A–D) each allowing the detection of four simultaneous slices but at different slice widths determined by how many detectors are binned together.

detector array modules. With a traditional single detector array CT system, the detectors are quite wide (e.g., 15 mm) and the adjustable collimator determines slice thickness, typically between 1 and 13 mm. With these systems, the spacing between the collimator blades is adjusted by small motors under computer control. With multiple detector arrays, slice width is determined by the detectors, not by the collimator (although a collimator does limit the beam to the total slice thickness). To allow the slice width to be adjustable, the detector width must be adjustable. It is not feasible, however, to physically change the width of the detector arrays *per se*. Therefore, with multislice systems, the slice width is determined by grouping one or more detector units together. For one manufacturer, the individual detector elements are 1.25 mm wide, and there are 16 contiguous detectors across the module. The detector dimensions are referenced to the scanner's *isocenter*, the point at the center of gantry rotation. The electronics are available for four detector array channels, and one, two, three or four detectors on the detector module can be combined to achieve slices of 4×1.25 mm, 4×2.50 mm, 4×3.75 mm, or 4×5.00 mm. To combine the signal from several detectors, the detectors are essentially wired together using computer-controlled switches. Other manufacturers use the same general approach but with different detector spacings. For example, one manufacturer uses 1-mm detectors everywhere except in the center, where four 0.5-mm-wide detectors are used. Other manufacturers use a gradually increasing spacing, with detector widths of 1.0, 1.5, 2.5, and 5.0 mm going away from the center. Increasing the number of active detector arrays beyond four (used in the example discussed) is a certainty.

Multiple detector array CT scanners make use of solid-state detectors. For a third-generation multiple detector array with 16 detectors in the slice thickness dimension and 750 detectors along each array, 12,000 individual detector elements are used. The fan angle commonly used in third-generation CT scanners is about 60 degrees, so fourth-generation scanners (which have detectors placed around 360 degrees) require roughly six times as many detectors as third-generation systems. Consequently, all currently planned multiple detector array scanners make use of third-generation geometry.

13.4 DETAILS OF ACQUISITION

Slice Thickness: Single Detector Array Scanners

The slice thickness in single detector array CT systems is determined by the physical collimation of the incident x-ray beam with two lead jaws. As the gap between the two lead jaws widens, the slice thickness increases. The width of the detectors in the single detector array places an upper limit on slice thickness. Opening the collimation beyond this point would do nothing to increase slice thickness, but would increase both the dose to the patient and the amount of scattered radiation.

There are important tradeoffs with respect to slice thickness. For scans performed at the same kV and mAs, the number of detected x-ray photons increases *linearly* with slice thickness. For example, going from a 1-mm to a 3-mm slice thickness *triples* the number of detected x-ray photons, and the signal-to-noise ratio (SNR) increases by 73%, since $\sqrt{3} = 1.73$. Increasing the slice thickness from

5 to 10 mm with the same x-ray technique (kV and mAs) *doubles* the number of detected x-ray photons, and the SNR increases by 41%: $\sqrt{2} = 1.41$. Larger slice thicknesses yield better contrast resolution (higher SNR) with the same x-ray techniques, but the spatial resolution in the slice thickness dimension is reduced. Thin slices improve spatial resolution in the thickness dimension and reduce partial volume averaging. For thin slices, the mAs of the study protocol usually is increased to partially compensate for the loss of x-ray photons resulting from the collimation.

It is common to think of a CT image as literally having the geometry of a slab of tissue, but this is not actually the case. The contrast of a small (e.g., 0.5 mm), highly attenuating ball bearing is greater if the bearing is in the center of the CT slice, and the contrast decreases as the bearing moves toward the edges of the slice (Fig. 13-17). This effect describes the *slice sensitivity profile*. For single detector array scanners, the shape of the slice sensitivity profile is a consequence of the finite width of the x-ray focal spot, the penumbra of the collimator, the fact that the image is computed from a number of projection angles encircling the patient, and other minor factors. Furthermore, helical scans have a slightly broader slice sensitivity profile due to translation of the patient during the scan. The *nominal slice thickness* is that which is set on the scanner control panel. Conceptually, the nominal slice is thought of as having a rectangular slice sensitivity profile.

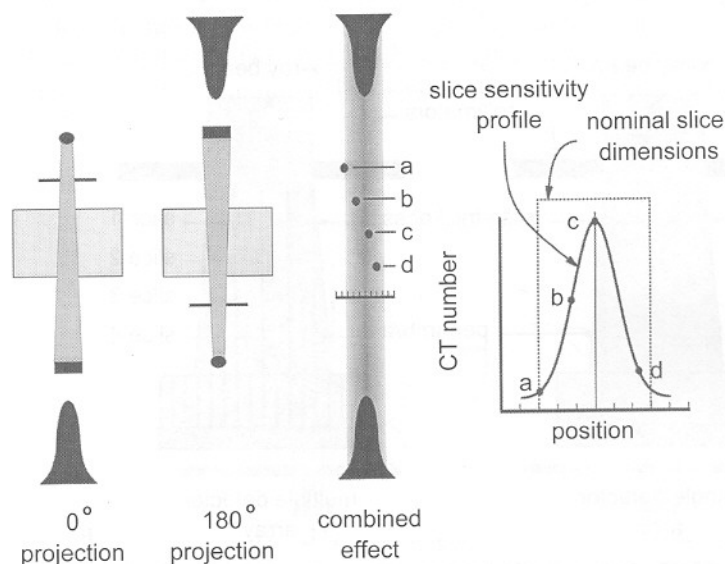


FIGURE 13-17. The slice sensitivity profile is a result of detector width; focal spot size and distribution; collimator penumbra, and the combined influence of all the projection angles. The nominal slice dimensions convey that the computed tomographic (CT) image is of a slab of tissue of finite thickness; however, the same object at different locations in the slice volume (e.g., positions A, B, C, and D) will produce different CT numbers in the image. Consequently, when a small object is placed in the center of the CT slice, it produces greater contrast from background (greater difference in CT number) than when the same object is positioned near the edge of the slice volume.

Slice Thickness: Multiple Detector Array Scanners

The slice thickness of multiple detector array CT scanners is determined not by the collimation, but rather by the width of the detectors in the slice thickness dimension. The width of the detectors is changed by *binning* different numbers of individual detector elements together—that is, the electronic signals generated by adjacent detector elements are electronically summed. Multiple detector arrays can be used both in conventional axial scanning and in helical scanning protocols. In axial scanning (i.e., with no table movement) where, for example, four detector arrays are used, the width of the two center detector arrays almost completely dictates the thickness of the slices. For the two slices at the edges of the scan (detector arrays 1 and 4 of the four active detector arrays), the inner side of the slice is determined by the edge of the detector, but the outer edge is determined either by the collimator penumbra or the outer edge of the detector, depending on collimator adjustment (Fig. 13-18). With a multiple detector array scanner in helical mode, each detector array contributes to every reconstructed image, and therefore the slice sensitivity profile for each detector array needs to be similar to reduce artifacts. To accommodate this condition, it is typical to adjust the collimation so that the focal spot–collimator blade penumbra falls outside the edge detectors. This causes the radiation dose to be a bit higher (especially for small slice widths) in multislice scanners, but it reduces artifacts by equalizing the slice sensitivity profiles between the detector arrays.

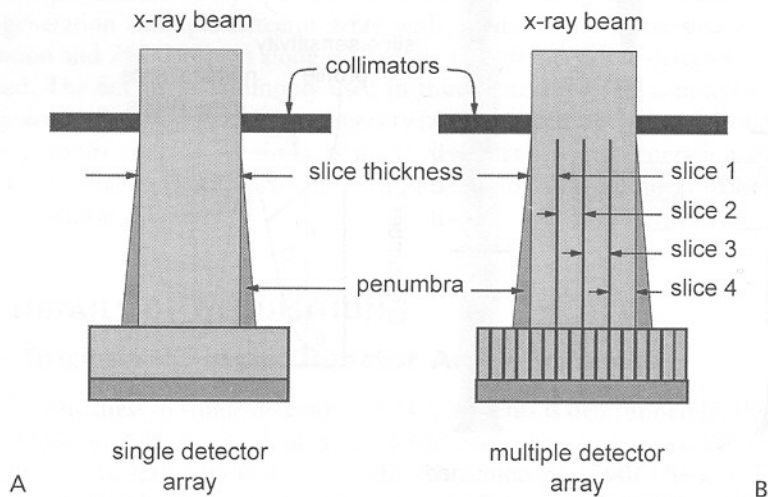


FIGURE 13-18. A: In a single detector array computed tomography (CT) scanner, the width of the CT slice is determined by the collimator. With single detector array scanners, the collimator width is always smaller than the maximum detector width. **B:** Multiple detector array CT scanners acquire several slices simultaneously (four slices shown here), and the x-ray beam collimation determines the outer two edges of the acquired slices. Slice thickness is changed in the single detector array by mechanically changing the collimator position. Slice thickness is determined in the multiple detector array scanner by binning different numbers of detector subunits together and by physically moving the collimator to the outer edges of all four slices.

Detector Pitch and Collimator Pitch

Pitch is a parameter that comes to play when helical scan protocols are used. In a helical CT scanner with one detector array, the pitch is determined by the collimator (collimator pitch) and is defined as:

$$\text{Collimator pitch} = \frac{\text{table movement (mm) per 360-degree rotation of gantry}}{\text{collimator width (mm) at isocenter}}$$

It is customary in CT to measure the collimator and detector widths at the isocenter of the system. The collimator pitch (Fig. 13-19) represents the traditional notion of *pitch*, before the introduction of multiple detector array CT scanners. Pitch is an important component of the scan protocol, and it fundamentally influences radiation dose to the patient, image quality, and scan time. For single detector array scanners, a pitch of 1.0 implies that the number of CT views acquired, when averaged over the long axis of the patient, is comparable to the number acquired with contiguous axial CT. A pitch of less than 1.0 involves overscanning, which may result in some slight improvement in image quality and a higher radiation dose to the patient. CT manufacturers have spent a great deal of developmental effort in optimizing scan protocols for pitches greater than 1.0, and pitches up to 1.5 are commonly used. Pitches with values greater than 1.0 imply some degree of partial scanning along the long axis of the patient. The benefit is faster scan time, less patient motion, and, in some circumstances, use of a smaller volume of contrast agent. Although CT acquisitions around 360 degrees are typical for images of the highest fidelity, the minimum requirement to produce an adequate CT image is a scan of 180 degrees plus the fan angle. With fan angles commonly at about 60 degrees, this means that, at a minimum, $(180 + 60)/360$, or 0.66, of the full circle is required.

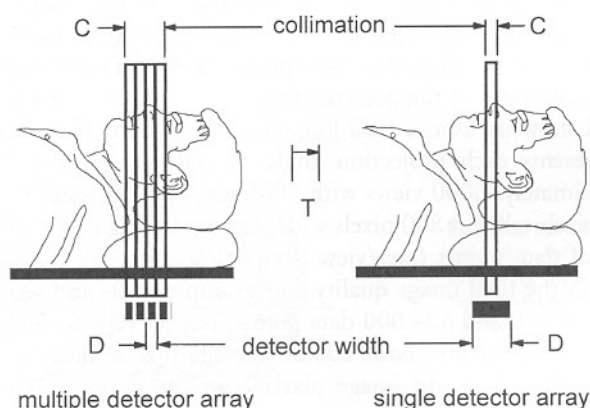


FIGURE 13-19. With a single detector array computed tomographic (CT) scanner, the collimation width is always narrower than the maximum single detector width. A multiple detector array scanner uses collimation that is always wider than a single detector array width. Letting T represent the table translation distance during a 360-degree rotation of the gantry, C would be the collimation width and D would be the detector width. Collimator pitch is defined as T/C , and detector width is defined by T/D . For a multiple detector array CT scanner with four detector arrays, a collimator pitch of 1.0 is equal to a detector pitch of 4.0.

This implies that the upper limit on pitch should be about 1/0.66, or 1.5, because 66% of the data in a 1.5-pitch scan remains contiguous.

Scanners that have multiple detector arrays require a different definition of pitch. The collimator pitch defined previously is still valid, and collimator pitches range between 0.75 and 1.5, as with single detector array scanners. The *detector pitch* (see Fig. 13-19) is also a useful concept for multiple detector array scanners, and it is defined as:

$$\text{Detector pitch} = \frac{\text{table movement (mm) per 360-degree rotation of gantry}}{\text{detector width (mm)}}$$

The need to define detector pitch and collimator pitch arises because beam utilization between single and multiple detector array scanners is different. For a multiple detector array scanner with N detector arrays, the collimator pitch is as follows:

$$\text{Collimator pitch} = \frac{\text{Detector pitch}}{N}$$

For scanners with four detector arrays, detector pitches running from 3 to 6 are used. A detector pitch of 3 for a four-detector array scanner is equivalent to a collimator pitch of 0.75 (3/4), and a detector pitch of 6 corresponds to a collimator pitch of 1.5 (6/4).

13.5 TOMOGRAPHIC RECONSTRUCTION

Rays and Views: The Sinogram

The data acquired for one CT slice can be displayed before reconstruction. This type of display is called a *sinogram* (Fig. 13-20). Sinograms are not used for clinical purposes, but the concepts that they embody are interesting and relevant to understanding tomographic principles. The horizontal axis of the sinogram corresponds to the different rays in each projection. For a third-generation scanner, for example, the horizontal axis of the sinogram corresponds to the data acquired at one instant in time along the length of the detector array. A bad detector in a third-generation scanner would show up as a vertical line on the sinogram. The vertical axis in the sinogram represents each projection angle. A state-of-the-art CT scanner may acquire approximately 1,000 views with 800 rays per view, resulting in a sinogram that is 1,000 pixels tall and 800 pixels wide, corresponding to 800,000 data points. The number of data points (rays/view $\times$ views) acquired by the CT scanner has some impact on the final image quality. For example, first- and second-generation scanners used 28,800 and 324,000 data points, respectively. A modern 512×512 circular CT image contains about 205,000 image pixels, and therefore the ratio between raw data points and image pixels is in the range of about 3.2 to 3.9 (800,000/205,000 = 3.9).

The number of rays used to reconstruct a CT image has a profound influence on the radial component of spatial resolution, and the number of views affects the circumferential component of the resolution (Fig. 13-21). CT images of a simulated object reconstructed with differing numbers of rays (Fig. 13-22) show that reducing the ray sampling results in low-resolution, blurred images. CT images of the simulated object reconstructed with differing numbers of views

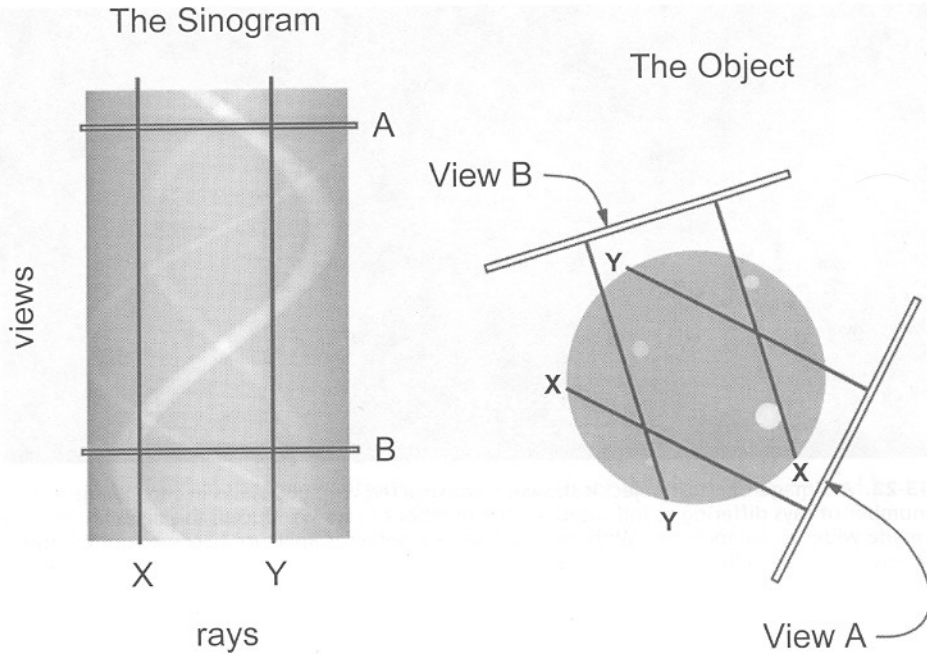


FIGURE 13-20. The sinogram is an image of the raw data acquired by a computed tomographic (CT) scanner before reconstruction. Here, the rays are plotted horizontally and the views are shown on the vertical axis. The sinogram illustrated corresponds to the circular object shown with four circular components. The two views marked A and B on the sinogram correspond to the two views illustrated with the object. During the 360-degree CT acquisition of a particular object, the position of the ray corresponding to that object varies sinusoidally as a function of the view angle. Objects closer to the edge of the field of view produce a sinusoid of high amplitude, and objects closer to the center of rotation have reduced sinusoidal amplitude. An object located exactly at the center of rotation of the gantry produces a shadow that appears vertical on the sinogram.

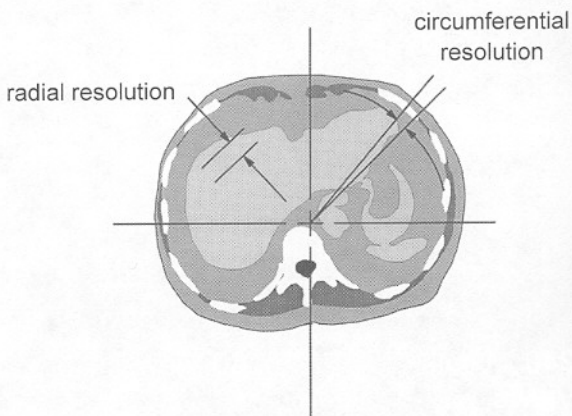


FIGURE 13-21. Unlike screen-film radiography, where the resolution of the image is largely independent of position, the spatial resolution in computed tomography, because of the reconstruction process, has both a radial and circumferential component. The radial component is governed primarily by the spacing and width of the ray data. The circumferential resolution is governed primarily by the number of views used to reconstruct the image.

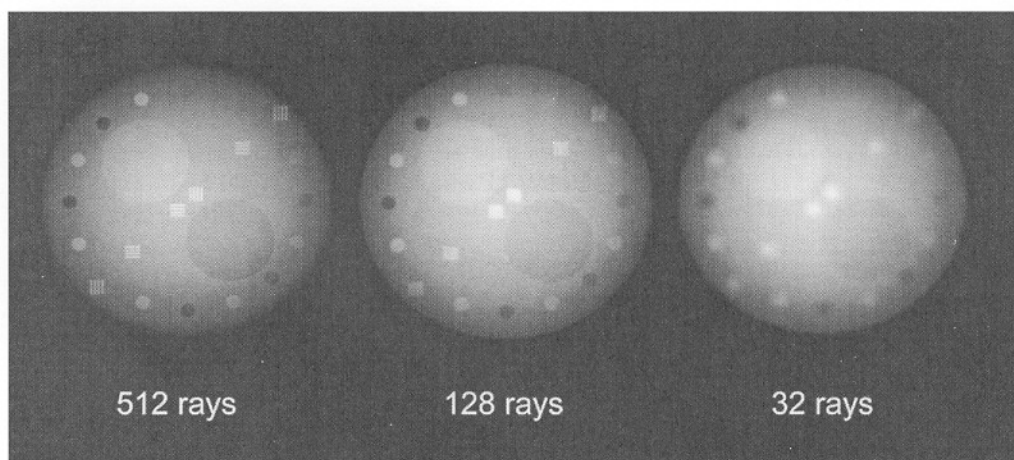


FIGURE 13-22. An image of a test object is shown, reconstructed with 960 views in each image, but with the number of rays differing as indicated. As the number of rays is reduced, the detector aperture was made wider to compensate. With only 32 rays, the detector aperture becomes quite large and the image is substantially blurred as a result.

(Fig. 13-23) show the effect of using too few angular views (*view aliasing*). The sharp edges (high spatial frequencies) of the bar patterns produce radiating artifacts that become more apparent near the periphery of the image. The view aliasing is mild with 240 views and becomes pronounced with 60 views. The presence of high-frequency objects (sharp edges) in the image exacerbates the impact of view aliasing.

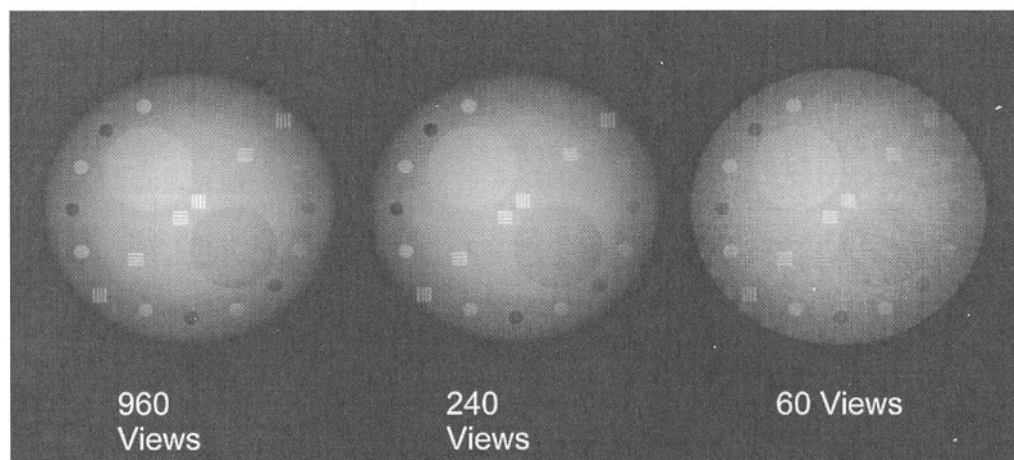


FIGURE 13-23. Each of the three images was reconstructed using 512 rays but differing numbers of views. In going from 960 to 240 views, a slight degree of view aliasing is seen (**center image**). When only 60 views are used (**right image**), the view aliasing is significant. The presence of the aliasing is exacerbated by objects with sharp edges, such as the line pair phantom in the image.

Preprocessing the Data

The raw data acquired by a CT scanner is preprocessed before reconstruction. Numerous preprocessing steps are performed. Electronic detector systems produce a digitized data set that is easily processed by the computer. Calibration data determined from air scans (performed by the technologist or service engineer periodically) provide correction data that are used to adjust the electronic gain of each detector in the array. Variation in geometric efficiencies caused by imperfect detector alignments is also corrected. For example, in fourth-generation scanners (see Fig. 13-9), the x-ray source rotates in an arc around each of the detectors in the array. As the angle and the distance between the detector and the x-ray tube change over this arc, the geometric efficiency of x-ray detection changes. (Similarly, the intensity of sunlight shining on one's palm changes as the wrist is rotated.) These geometric efficiencies are measured during calibration scans and stored in the computer; they are corrected in the preprocessing steps.

After the digital data are calibrated, the logarithm of the signal is computed by the following equation:

$$\ln(I_0/I_t) = \mu t$$

where I_0 is the reference data, I_t is the data corresponding to each ray in each view, t is the patient's overall thickness (in centimeters) for that ray, and μ is the linear attenuation coefficient (in cm^{-1}). Note that the data used to calculate the images is linear with respect to the linear attenuation coefficient, μ . The linear attenuation coefficient is determined by the composition and density of the tissue inside each voxel in the patient. Whereas μ is the total attenuation coefficient for each ray (Fig. 13-24), it can be broken down into its components in each small path length Δt :

$$\mu t = \mu_1 \Delta t + \mu_2 \Delta t + \mu_3 \Delta t + \dots + \mu_n \Delta t = \Delta t [\mu_1 + \mu_2 + \mu_3 + \dots + \mu_n]$$

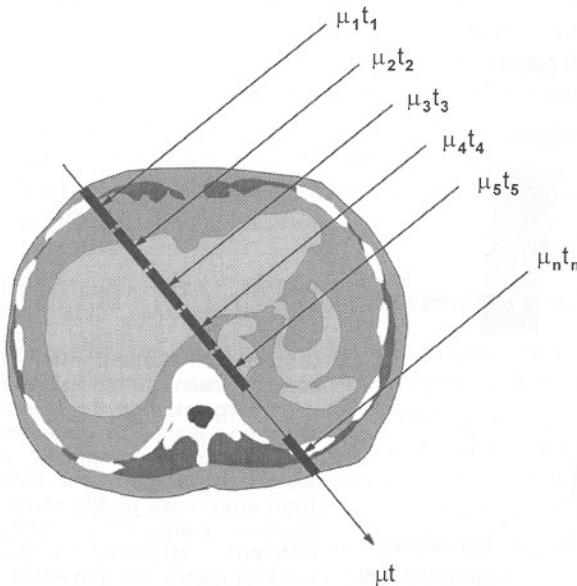


FIGURE 13-24. The measured value of a single ray is equivalent to the attenuation path (t) times the linear attenuation coefficient (μ). The single measurement of μt can be broken up into a series of measurements $\mu_1 t_1$, $\mu_2 t_2$, ..., $\mu_n t_n$, as shown.

The constant Δt factors out, resulting in the following:

$$\mu = \mu_1 + \mu_2 + \mu_3 + \dots + \mu_n$$

Therefore, the reconstructed value in each pixel is the linear attenuation coefficient for the corresponding voxel.

Interpolation (Helical)

Helical CT scanning produces a data set in which the x-ray source has traveled in a helical trajectory around the patient. Present-day CT reconstruction algorithms assume that the x-ray source has negotiated a circular, not a helical, path around the patient. To compensate for these differences in the acquisition geometry, before the actual CT reconstruction the helical data set is interpolated into a series of planar image data sets, as illustrated in Fig. 13-25. Although this interpolation represents an additional step in the computation, it also enables an important feature. With conventional axial scanning, the standard is to acquire contiguous images, which abut one another along the cranial-caudal axis of the patient. With helical scanning, however, CT images can be reconstructed at any position along the length of the scan to within $(\frac{1}{2})$ (pitch) (slice thickness) of each edge of the scanned volume. Helical scanning allows the production of additional overlapping images with *no additional dose* to the patient. Figure 13-26 illustrates the added clinical importance of using interleaved images. The sensitivity of the CT image to objects not centered in the voxel is reduced (as quantified by the slice sensitivity profile), and therefore subtle lesions, which lay between two contiguous images, may be missed. With helical CT scanning, interleaved reconstruction allows the placement of additional images along the patient, so that the clinical examination is almost uniformly sensitive to subtle abnormalities. Interleaved reconstruction adds no additional radiation dose to the patient, but additional time is required to reconstruct the images. Although an increase in the

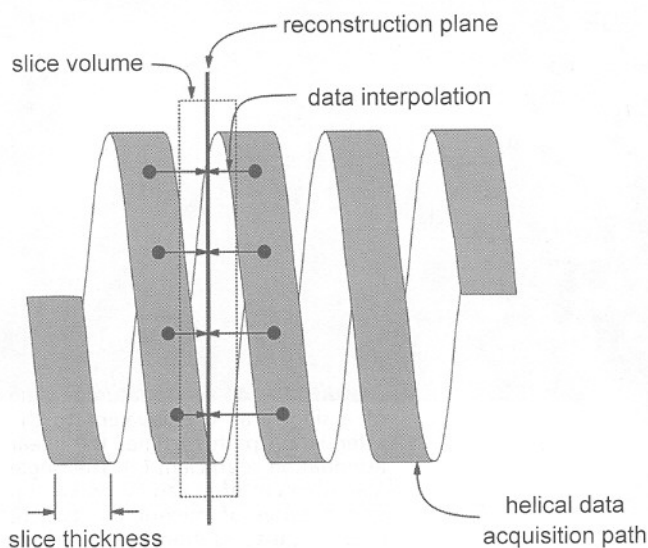


FIGURE 13-25. During helical acquisition, the data are acquired in a helical path around the patient. Before reconstruction, the helical data are interpolated to the reconstruction plane of interest. Interpolation is essentially a weighted average of the data from either side of the reconstruction plane, with slightly different weighting factors used for each projection angle.

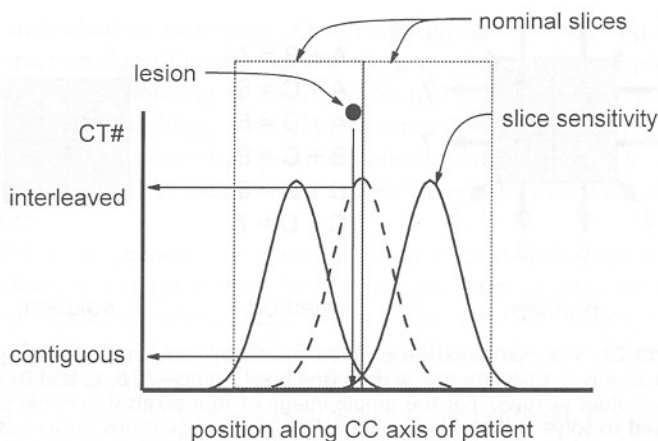


FIGURE 13-26. This figure illustrates the value of interleaved reconstruction. The nominal slice for contiguous computed tomographic (CT) images is illustrated conceptually as two adjacent rectangles; however, the sensitivity of each CT image is actually given by the slice sensitivity profile (solid lines). A lesion that is positioned approximately between the two CT images (black circle) produces low contrast (i.e., a small difference in CT number between the lesion and the background) because it corresponds to low slice sensitivity. With the use of interleaved reconstruction (dashed line), the lesion intersects the slice sensitivity profile at a higher position, producing higher contrast.

image count would increase the interpretation time for traditional side-by-side image presentation, this concern will ameliorate as more CT studies are read by radiologists at computer workstations.

It is important not to confuse the ability to reconstruct CT images at short intervals along the helical data set with the axial resolution itself. The slice thickness (governed by collimation with single detector array scanners and by the detector width in multislice scanners) dictates the actual spatial resolution along the long axis of the patient. For example, images with 5-mm slice thickness can be reconstructed every 1 mm, but this does not mean that 1-mm spatial resolution is achieved. It simply means that the images are sampled at 1-mm intervals. To put the example in technical terms, the *sampling pitch* is 1 mm but the *sampling aperture* is 5 mm. In practice, the use of interleaved reconstruction much beyond a 2:1 interleave yields diminishing returns, except for multiplanar reconstruction (MPR) or 3D rendering applications.

Simple Backprojection Reconstruction

Once the image raw data have been preprocessed, the final step is to use the planar projection data sets (i.e., the preprocessed sinogram) to reconstruct the individual tomographic images. *Iterative reconstruction techniques* are discussed in Chapter 22 and will not be discussed here because they are not used in clinical x-ray CT. As a basic introduction to the reconstruction process, consider Fig. 13-27. Assume that a very simple 2×2 "image" is known only by the projection values. Using algebra ("N equations in M unknowns"), one can solve for the image values in the simple case of a 4-pixel image. A modern CT image contains approx-

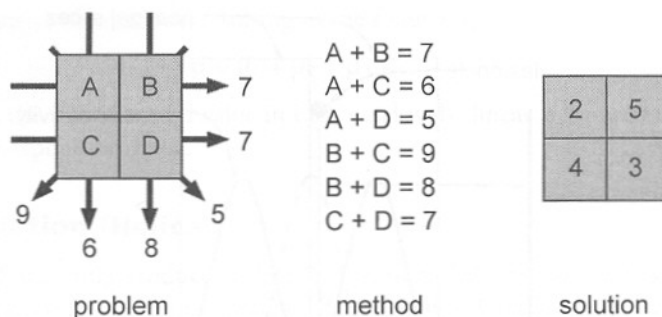


FIGURE 13-27. The mathematical problem posed by computed tomographic (CT) reconstruction is to calculate image data (the pixel values—A, B, C, and D) from the projection values (arrows). For the simple image of four pixels shown here, algebra can be used to solve for the pixel values. With the six equations shown, using substitution of equations, the solution can be determined as illustrated. For the larger images of clinical CT, algebraic solutions become unfeasible, and filtered backprojection methods are used.

imately 205,000 pixels (the circle within a 512×512 matrix) or “unknowns,” and each of the 800,000 projections represent an individual equation. Solving this kind of a problem is beyond simple algebra, and backprojection is the method of choice.

Simple backprojection is a mathematical process, based on trigonometry, which is designed to emulate the acquisition process in reverse. Each ray in each view rep-

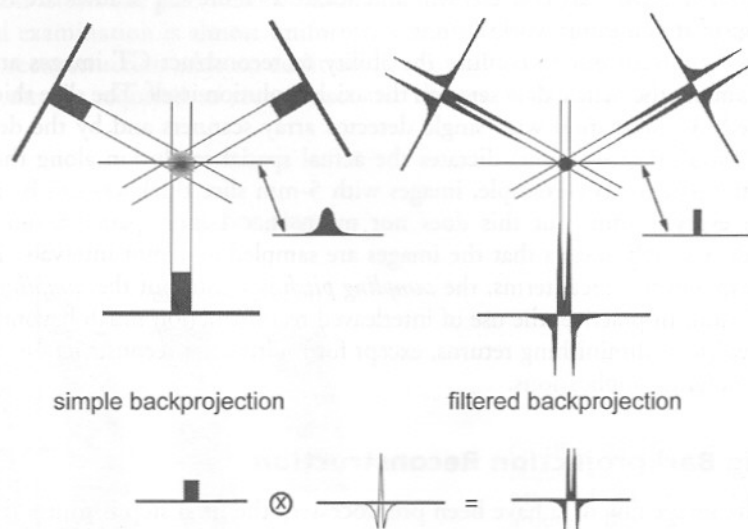


FIGURE 13-28. Simple backprojection is shown on the left; only three views are illustrated, but many views are actually used in computed tomography. A profile through the circular object, derived from simple backprojection, shows a characteristic $1/r$ blurring. With filtered backprojection, the raw projection data are convolved with a convolution kernel and the resulting projection data are used in the backprojection process. When this approach is used, the profile through the circular object demonstrates the crisp edges of the cylinder, which accurately reflects the object being scanned.

resents an individual measurement of μ . In addition to the value of μ for each ray, the reconstruction algorithm also "knows" the acquisition angle and position in the detector array corresponding to each ray. Simple backprojection starts with an empty image matrix (an image with all pixels set to zero), and the μ value from each ray in all views is smeared or backprojected onto the image matrix. In other words, the value of μ is added to each pixel in a line through the image corresponding to the ray's path.

Simple backprojection comes very close to reconstructing the CT image as desired. However, a characteristic $1/r$ blurring (see later discussion) is a byproduct of simple backprojection. Imagine that a thin wire is imaged by a CT scanner perpendicular to the image plane; this should ideally result in a small point on the image. Rays not running through the wire will contribute little to the image ($\mu = 0$). The backprojected rays, which do run through the wire, will converge at the position of the wire in the image plane, but these projections run from one edge of the reconstruction circle to the other. These projections (i.e., lines) will therefore "radiate" geometrically in all directions away from a point input. If the image gray scale is measured as a function of distance away from the center of the wire, it will gradually diminish with a $1/r$ dependency, where r is the distance away from the point. This phenomenon results in a blurred image of the actual object (Fig. 13-28) when simple backprojection is used. A filtering step is therefore added to correct this blurring, in a process known as *filtered backprojection*.

Filtered Backprojection Reconstruction

In filtered backprojection, the raw view data are mathematically filtered before being backprojected onto the image matrix. The filtering step mathematically reverses the image blurring, restoring the image to an accurate representation of the object that was scanned. The mathematical filtering step involves *convolving* the projection data with a convolution *kernel*. Many convolution kernels exist, and different kernels are used for varying clinical applications such as soft tissue imaging or bone imaging. A typical convolution kernel is shown in Fig. 13-28. The kernel refers to the shape of the filter function in the spatial domain, whereas it is common to perform (and to think of) the filtering step in the frequency domain. Much of the nomenclature concerning filtered backprojection involves an understanding of the frequency domain (which was discussed in Chapter 10). The Fourier transform (FT) is used to convert a function expressed in the spatial domain (millimeters) into the spatial frequency domain (cycles per millimeter, sometimes expressed as mm^{-1}); the inverse Fourier transform (FT^{-1}) is used to convert back. Convolution is an integral calculus operation and is represented by the symbol $\otimes$. Let $p(x)$ represent projection data (in the spatial domain) at a given angle ($p(x)$ is just one horizontal line from a sinogram; see Fig. 13-20), and let $k(x)$ represent the spatial domain kernel as shown in Fig. 13-28. The filtered data in the spatial domain, $p'(x)$, is computed as follows:

$$p'(x) = p(x) \otimes k(x)$$

The difference between filtered backprojection and simple backprojection is the mathematical filtering operation (convolution) shown in this equation above. In filtered backprojection, $p'(x)$ is backprojected onto the image matrix, whereas in

simple backprojection, $p(x)$ is backprojected. The equation can also be performed, quite exactly, in the frequency domain:

$$p'(x) = FT^{-1}\{FT[p(x)] \times K(f)\}$$

where $K(f) = FT[k(x)]$, the kernel in the frequency domain. This equation states that the convolution operation can be performed by Fourier transforming the projection data, *multiplying* (not convolving) this by the frequency domain kernel ($K(f)$), and then applying the inverse Fourier transform on that product to get the filtered data, ready to be backprojected.

Various convolution filters can be used to emphasize different characteristics in the CT image. Several filters, shown in the frequency domain, are illustrated in Fig. 13-29, along with the reconstructed CT images they produced. The Lak filter, named for Dr. Lakshminarayanan, increases the amplitude linearly as a function of frequency and is also called a *ramp filter*. The $1/r$ blurring function in the spatial domain becomes a $1/f$ blurring function in the frequency domain; therefore, multiplying by the Lak filter, where $L(f) = f$, exactly compensates the unwanted $1/f$ blurring, because $1/f \times f = 1$, at all f . This filter works well when there is no noise

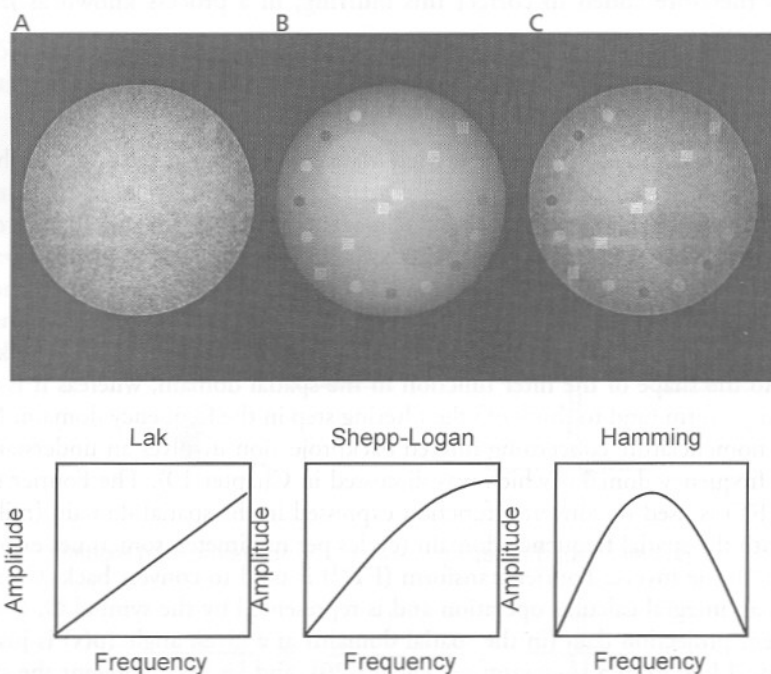


FIGURE 13-29. Three computed tomography (CT) reconstruction filters are illustrated. The plot of the reconstruction filter is shown below, with the image it produced above. **(A)** The Lak is the ideal reconstruction filter in the absence of noise. However, with the quantum noise characteristic in x-ray imaging including CT, the high-frequency components of the noise are amplified and the reconstructed image is consequently very noisy. **(B)** The Shepp-Logan filter is similar to the Lak filter, but roll-off occurs at the higher frequencies. This filter produces an image that is a good compromise between noise and resolution. **(C)** The Hamming filter has extreme high frequency roll-off.

in the data, but in x-ray images there is always x-ray quantum noise, which tends to be more noticeable in the higher frequencies. The Lak filter, as shown in Fig. 13-29A, produces a very noisy CT image. The Shepp-Logan filter in Fig. 13-29B is similar to the Lak filter but incorporates some *roll-off* at higher frequencies, and this reduction in amplification at the higher frequencies has a profound influence in terms of reducing high-frequency noise in the final CT image. The Hamming filter (see Fig. 13-29C) has an even more pronounced high-frequency roll-off, with better high-frequency noise suppression.

Bone Kernels and Soft Tissue Kernels

The reconstruction filters derived by Lakshminarayanan, Shepp and Logan, and Hamming provide the mathematical basis for CT reconstruction filters. In clinical CT scanners, the filters have more straightforward names, and terms such as “bone filter” and “soft tissue filter” are common among CT manufacturers. The term *kernel* is also used. Bone kernels have less high-frequency roll-off and hence accentuate higher frequencies in the image at the expense of increased noise. CT images of bones typically have very high contrast (high signal), so the SNR is inherently quite good. Therefore, these images can afford a slight decrease in SNR ratio in return for sharper detail in the bone regions of the image.

For clinical applications in which high spatial resolution is less important than high contrast resolution—for example, in scanning for metastatic disease in the liver—soft tissue reconstruction filters are used. These kernels have more roll-off at higher frequencies and therefore produce images with reduced noise but lower spatial resolution. The resolution of the images is characterized by the modulation transfer function (MTF); this is discussed in detail in Chapter 10. Figure 13-30 demonstrates two MTFs for state-of-the-art CT scanners. The high-resolution MTF corresponds to use of the bone filter at small FOV, and the standard resolution corresponds to images produced with the soft tissue filter at larger FOV.

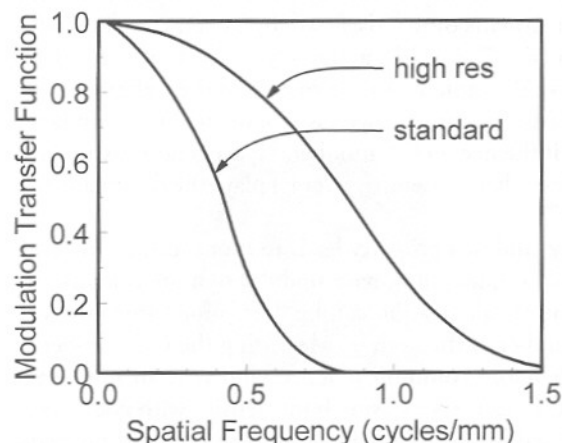


FIGURE 13-30. The modulation transfer function (MTF) for standard and high-resolution clinical computed tomographic scanners are shown. The standard MTF is for a large field of view and a soft tissue reconstruction filter. The high-resolution MTF corresponds to a small field of view and a bone reconstruction filter.

The units for the x-axis in Fig. 13-30 are in cycles per millimeter, and the cutoff frequency is approximately 1.0 cycles/mm. This cutoff frequency is similar to that of fluoroscopy, and it is five to seven times lower than in general projection radiography. CT manufacturers have adapted the unit *cycle/cm*—for example, 1.2 cycles/mm = 12 cycles/cm.

CT Numbers or Hounsfield Units

After CT reconstruction, each pixel in the image is represented by a high-precision floating point number that is useful for computation but less useful for display. Most computer display hardware makes use of integer images. Consequently, after CT reconstruction, but before storing and displaying, CT images are normalized and truncated to integer values. The number $CT(x,y)$ in each pixel, (x,y) , of the image is converted using the following expression:

$$CT(x,y) = 1,000 \frac{\mu(x,y) - \mu_{\text{water}}}{\mu_{\text{water}}}$$

where $\mu(x,y)$ is the floating point number of the (x,y) pixel before conversion, μ_{water} is the attenuation coefficient of water, and $CT(x,y)$ is the CT number (or Hounsfield unit) that ends up in the final clinical CT image. The value of μ_{water} is about 0.195 for the x-ray beam energies typically used in CT scanning. This normalization results in CT numbers ranging from about $-1,000$ to $+3,000$, where $-1,000$ corresponds to air, soft tissues range from -300 to -100 , water is 0, and dense bone and areas filled with contrast agent range up to $+3,000$.

What do CT numbers correspond to *physically* in the patient? CT images are produced with a highly filtered, high-kV x-ray beam, with an average energy of about 75 keV. At this energy in muscle tissue, about 91% of x-ray interactions are Compton scatter. For fat and bone, Compton scattering interactions are 94% and 74%, respectively. Therefore, CT numbers and hence CT images derive their contrast mainly from the physical properties of tissue that influence Compton scatter. Density (g/cm^3) is a very important discriminating property of tissue (especially in lung tissue, bone, and fat), and the linear attenuation coefficient, μ , tracks linearly with density. Other than physical density, the Compton scatter cross section depends on the electron density (ρ_e) in tissue: $\rho_e = NZ/A$, where N is Avogadro's number (6.023×10^{23} , a constant), Z is the atomic number, and A is the atomic mass of the tissue. The main constituents of soft tissue are hydrogen ($Z = 1$, $A = 1$), carbon ($Z = 6$, $A = 12$), nitrogen ($Z = 7$, $A = 14$), and oxygen ($Z = 8$, $A = 16$). Carbon, nitrogen, and oxygen all have the same Z/A ratio of 0.5, so their electron densities are the same. Because the Z/A ratio for hydrogen is 1.0, the relative abundance of hydrogen in a tissue has some influence on CT number. Hydrogenous tissue such as fat is well visualized on CT. Nevertheless, density (g/cm^3) plays the dominant role in forming contrast in medical CT.

CT numbers are *quantitative*, and this property leads to more accurate diagnosis in some clinical settings. For example, pulmonary nodules that are calcified are typically benign, and the amount of calcification can be determined from the CT image based on the mean CT number of the nodule. Measuring the CT number of a single pulmonary nodule is therefore common practice, and it is an important part of the diagnostic work-up. CT scanners measure bone density with good accuracy, and when phantoms are placed in the scan field along with the patient, quan-

titative CT techniques can be used to estimate bone density, which is useful in assessing fracture risk. CT is also quantitative in terms of linear dimensions, and therefore it can be used to accurately assess tumor volume or lesion diameter.

Computed Tomographic Fluoroscopy Reconstruction

In CT fluoroscopy, the scanner provides pseudo-real time tomographic images that are most commonly used for guidance during biopsies. CT fluoroscopy is similar to conventional CT, with a few added technologic nuances. Like conventional fluoroscopy, CT fluoroscopy provides an image sequence over the same region of tissue, and therefore the radiation dose from repetitive exposure to that tissue is a concern. To address this problem, CT scanners are capable of a CT fluoroscopic mode that uses a low milliamperage setting, typically 20 to 50 mA, whereas 150 to 400 mA is typical for conventional CT imaging.

To provide real-time CT images, special computer hardware is required. Each fluoroscopic CT frame in an imaging sequence is not unique from the preceding CT frame. For example, to achieve 6 CT images per second with a 1-second gantry rotation, the angle scanned during each frame time of $1/6$ second is 60 degrees ($360 \text{ degrees} / 6 = 60 \text{ degrees}$). For CT reconstruction using a full 360-degree projection set, each CT fluoroscopic image uses 17% ($1/6$) new information and 83% ($5/6$) old information (Fig. 13-31). The image acquisition time is still 1 second in this example, but the image is updated 6 times per second. This clever approach provides increased temporal resolution given the physical constraints of backprojection algorithms and the hardware constraints of current technology.

The full data set for each CT fluoroscopic image is not completely re-reconstructed during each frame interval. Using the example of 6 frames per second, after 1 second of operation, consider what occurs in the next $1/6$ second (167 msec): The scanner completes a 60-degree arc, and the projection data acquired during that 167

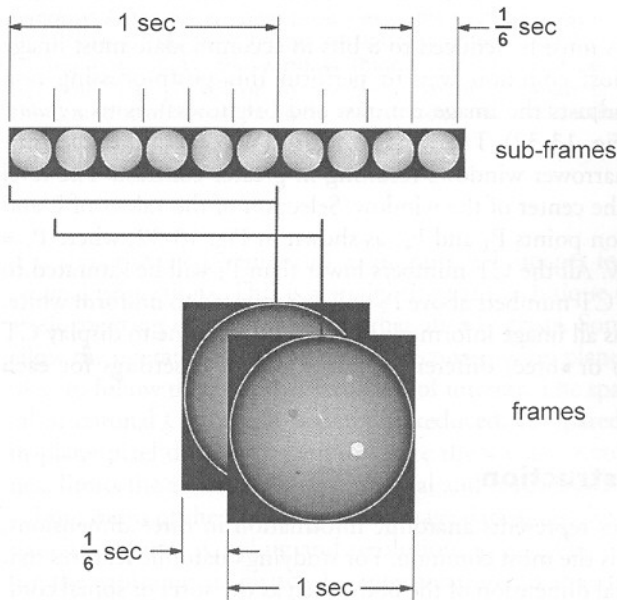


FIGURE 13-31. Data handling in computed tomographic fluoroscopy is illustrated, here using the example of acquisition at 6 frames per second and a 1-second gantry rotation time. During each $1/6$ second, the CT scanner acquires 60 degrees of projection data. Each of these subframes can be filtered and backprojected as shown. A full-frame image is produced by adding six subframes together. Six CT frames are produced each second; however, each CT image corresponds to a scan time of 1 second. For each subsequent CT frame, the newest subframe and the five previous subframes are added.

msec is preprocessed, mathematically filtered, and backprojected to create a CT subframe. This new CT subframe (an image) is added (pixel by pixel) to the five previous subframes, creating a complete CT frame ready for display. Therefore, every 167 msec, a new subframe is created and an old subframe is discarded, so that the most recent six subframes are summed to produce the displayed CT frame. Although 6 frames per second was used in this example, higher frame rates are clinically available.

13.6 DIGITAL IMAGE DISPLAY

Once the CT images from a patient's examination have been reconstructed, the image data must be conveyed to the physician for review and diagnosis. There are some basic postprocessing techniques that are applied to all CT images (i.e., windowing and leveling). For CT studies that have contiguous or interleaving images over a length of the patient, the volume data set lends itself to other forms of postprocessing.

Windowing and Leveling

CT images typically possess 12 bits of gray scale, for a total of 4,096 shades of gray (CT numbers range from -1,000 to +3,095). Electronic display devices such as computer monitors have the ability to display about 8 bits (256 shades of gray), and laser imagers used for "filming" CT studies have about 8 bits of display fidelity as well. The human eye has incredible dynamic range for adapting to *absolute* light intensities, from moonless nights to desert sunshine. Much of this dynamic range is a result of the eye's ability to modulate the diameter of the light-collecting aperture (the pupil). However, for resolving *relative* differences in gray scale at fixed luminance levels (fixed pupil diameter), as in viewing medical images, the human eye has limited ability (30 to 90 shades of gray), and 6 to 8 bits is considered sufficient for image display.

The 12-bit CT images must be reduced to 8 bits to accommodate most image display hardware. The most common way to perform this postprocessing task (which nondestructively adjusts the image contrast and brightness), is to *window and level* the CT image (Fig. 13-32). The window width (W) determines the contrast of the image, with narrower windows resulting in greater contrast. The level (L) is the CT number at the center of the window. Selection of the values of L and W determines the inflection points P_1 and P_2 , as shown in Fig. 13-32, where $P_1 = L - \frac{1}{2}W$ and $P_2 = L + \frac{1}{2}W$. All the CT numbers lower than P_1 will be saturated to uniform black, and all the CT numbers above P_2 will be saturated to uniform white, and in these saturated areas all image information is lost. It is routine to display CT images using several (two or three) different window and level settings for each image.

Multiplanar Reconstruction

A stack of axial CT images represents anatomic information in three dimensions; however, the axial display is the most common. For studying anatomic features that run along the cranial-caudal dimension of the body, such as the aorta or spinal cord,

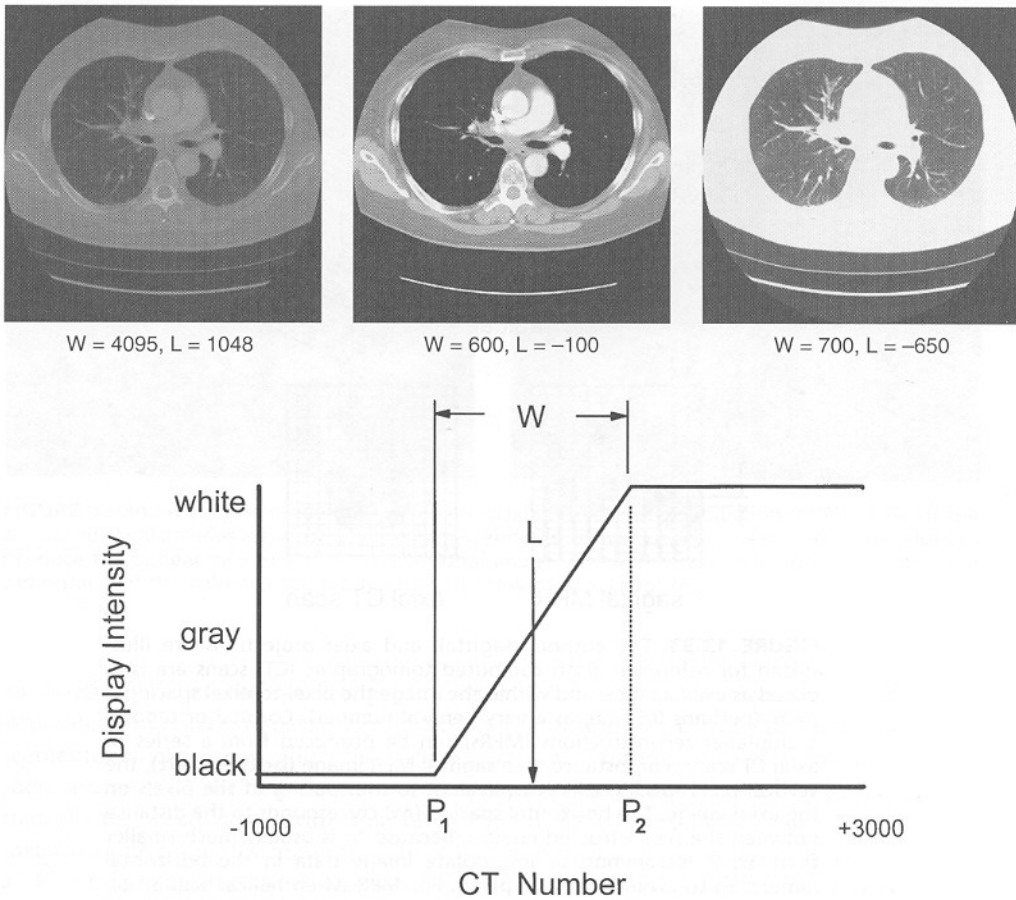


FIGURE 13-32. The concept of how the window and level values are used to manipulate the contrast of the CT image is illustrated. The *level* is the CT number corresponding to the center of the *window*. A narrow window produces a very high contrast image, corresponding to a large slope on the figure. CT numbers below the window (lower than P_1) will be displayed on the image as black; CT numbers above the window (higher than P_2) will be displayed as white. Only CT numbers between P_1 and P_2 will be displayed in a meaningful manner. The thoracic CT images (**top**) illustrate the dramatic effect of changing the window and level settings.

it is sometimes convenient to recompute CT images for display in the sagittal or coronal projections. This is a straightforward technique (Fig. 13-33), and is simply a reformatting of the image data that already exists. Some MPR software packages allow the operator to select a curved reconstruction plane for sagittal or coronal display, to follow the anatomic structure of interest. The spatial resolution of the sagittal or coronal CT images is typically reduced, compared with the axial views. The in-plane pixel dimensions approximate the x-y axis resolution, but the slice thickness limits the z-axis resolution. Sagittal and coronal MPR images combine the x or y dimensions of the CT image with image data along the z-axis, and therefore a mismatch in spatial sampling and resolution occurs. The MPR software compensates for the sampling mismatch by interpolation, so that the MPR images maintain

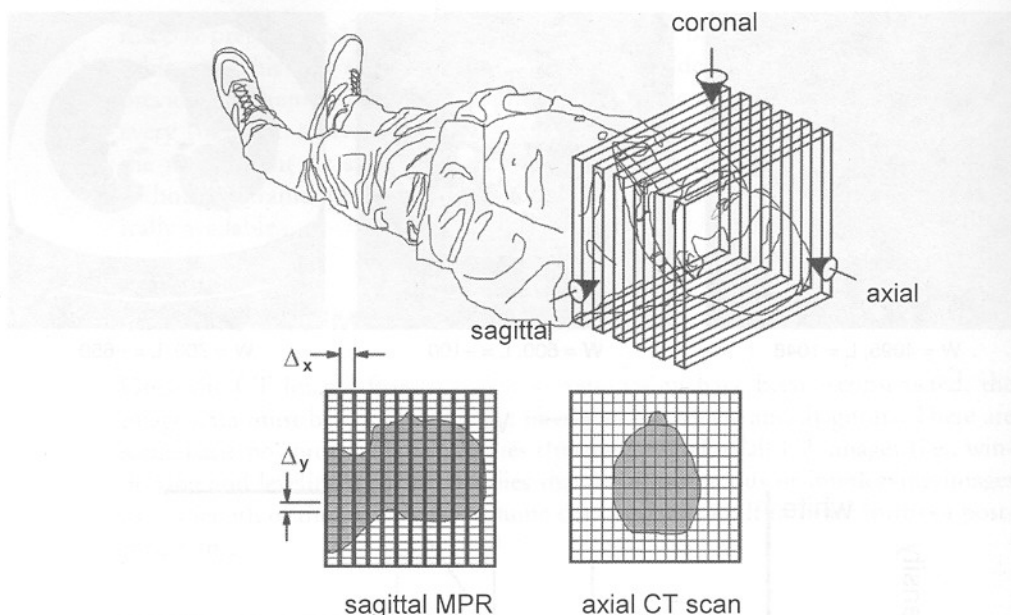


FIGURE 13-33. The coronal, sagittal, and axial projections are illustrated for reference. Both computed tomographic (CT) scans are produced as axial images, and within the image the pixel-to-pixel spacing is small (perhaps 0.5 mm, as a very general number). Coronal or sagittal multiplanar reconstructions (MPRs) can be produced from a series of axial CT scans. For instance, in a sagittal MPR image (**bottom left**), the vertical pixel spacing (Δy) is equivalent to the spacing of the pixels on the axial image. The horizontal spacing (Δx) corresponds to the distance between the reconstructed images. Because Δy is usually much smaller than Δx , it is common to interpolate image data in the horizontal dimension to produce square pixels. For MPR when helical acquisition was used, it is advantageous to reconstruct the images with a significant degree of interleave.

proper aspect ratio (pixel height equals pixel width). However, the system cannot improve the resolution in the z dimension, so MPR images appear a bit blurry along this axis. If it is known before the CT study that MPR images are to be computed, the use of a protocol with thinner CT slices (down to 1 mm) to improve z -axis resolution is warranted.

Three-Dimensional Image Display

Most radiologists are sufficiently facile at viewing series of 2D images to garner a 3D understanding of the patient's anatomy and pathology. In some cases, however, referring physicians prefer to see the CT image data presented in a pseudo-3D manner. Examples would include planning for craniofacial surgery, aortic stent grafting, or radiation treatment. There are two general classes of volume reconstruction techniques: volume rendering techniques and reprojection techniques (Fig. 13-34).

Volume rendering of CT images requires *segmentation*, the identification (by computer alone or with the aid of human intervention) of specific target structures on the 2D image series, before 3D rendering. Segmentation is easiest to perform

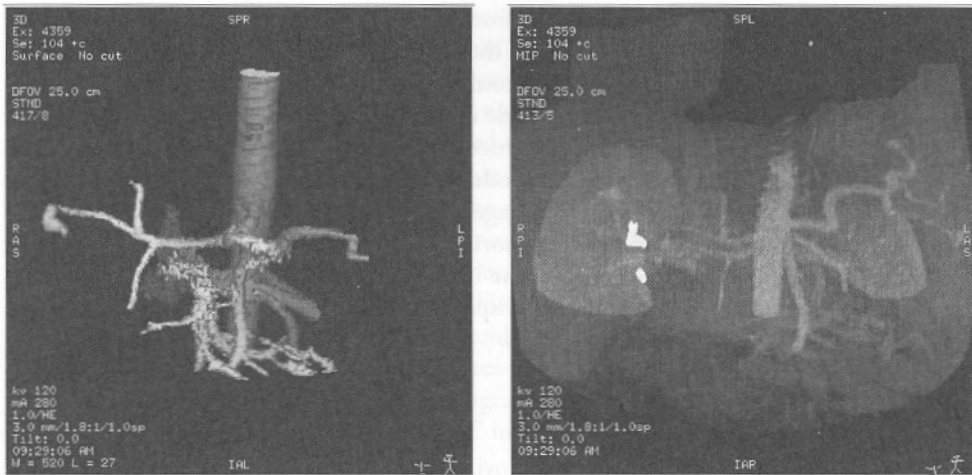


FIGURE 13-34. A volume-rendered image and maximum intensity projection (MIP) image of the aorta and renal arteries are illustrated. Volume-rendered images use surface shading techniques to produce the appearance of a three-dimensional image. The MIP display is similar to a projection radiograph of the volume data set. (Images courtesy of Rick Labonte.)

automatically when there is a large difference in CT number between the target structures and adjacent anatomy (e.g., between bone and soft tissue). The simplest segmentation algorithms allow the user to select a CT number, and the algorithm then assumes that the target structure occupies all pixels with CT numbers greater than the threshold value. Segmentation of the CT data set essentially converts the images to one-bit representations: a "0" if the target structure is not in the pixel and a "1" if it is. Once the target structures (e.g., the facial bones) have been accurately segmented, a computer program then calculates ("renders") the theoretical appearance of the structure from a specified angle. The software calculates a number of surfaces from the segmented data set, and this process is sometimes called *surface rendering*. Surfaces closest to the viewer obscure surfaces at greater depth. Depending on the sophistication of the software, single or multiple light sources may be factored into the computation to give the appearance of surface shading, color may be added to identify different target structures (i.e., green tumor, white bones, pink spinal cord), and so on. Volume rendering has reached spectacular technologic sophistication in the movie industry, with *Jurassic Park* being the classic example. Medical imaging software has and will continue to benefit from nonmedical rendering advancements, but it will never meet the same degree of success because of the SNR limitations in the data set. The dinosaurs in *Jurassic Park* were calculated from sophisticated mathematical models with perfect SNR, whereas rendering of cranial-facial bones requires a clinical CT data set with imperfect SNR.

Because of SNR limitations in CT imaging, in some studies segmentation, which is required for volume rendering, can be a time-consuming task requiring substantial human intervention, reducing the appeal of this approach. To address this problem, reprojection techniques that do not require segmentation have been developed for the display of volume data. From a specified viewing angle, images are computed from the CT volume data set which are similar geometrically to radiographic projection images. Reprojection techniques simply use ray-tracing soft-

ware through the volume data set, from the specified viewing angle. The software can display the normalized sum of the CT numbers from all the voxels through which each ray passed. More commonly, the maximum CT number that each ray encounters is displayed, and this mode is called *maximum intensity projection* (MIP). MIP displays are perhaps not as “three-dimensional” as volume-rendered images, but they can be generated quite reproducibly and completely automatically. To augment the 3D appearance of MIP images, often a series of such images are produced at different viewing angles, and a short movie sequence is generated. For example, a sequence rotating the cervical spine in real time could be displayed. The spatial-temporal cues from the real-time display of these movies can be quite revealing, especially for complicated structures.

Stack Mode Viewing

Many radiology departments are moving to a filmless environment, in which the actual diagnostic interpretation is performed by a radiologist who sits at a computer workstation and views *soft copy* images. One approach is to try to emulate the film alternator to the extent possible, with perhaps four or even six high-resolution monitors clustered together to display the entire CT data set simultaneously. Although this approach seeks to emulate an environment with which most radiologists are familiar, four- or six-monitor workstations are expensive and bulky, and this approach is not designed to take advantage of a key attribute of soft copy display—the computer.

An alternative approach is to simplify the computer hardware, to perhaps two high-resolution monitors, and allow the computer to do the work of the radiologist's neck. In stack mode, a single CT image is displayed and the radiologist selects the image in the study, which is displayed by moving the computer mouse. Often the scout view is presented next to the CT image, with the current image location highlighted. It is possible to simultaneously display two CT images at the same cut plane, such as precontrast and postcontrast images, images at different window and level settings, or images from the previous and the current CT examinations. Certainly many permutations of stack mode display exist, but the essential advantage of this approach is that it is interactive: The radiologist interacts with the computer in real time to visualize the image data as he or she interprets the case, following diagnostic clues from slice to slice. As one clinical example of the benefits of stack mode viewing, a common task in CT is to follow arteries from image to image. On an alternator-like, noninteractive display, the viewer must relocate the vessel spatially, shifting the gaze from image to image. In stack mode display, the viewer's eyes are fixed on the vessel while the various images are displayed under mouse control, allowing the viewer's gaze to follow the vessel as it changes position slightly from image to image.

13.7 RADIATION DOSE

Different x-ray modalities address radiation dose in different ways. For example, in chest radiography it is the entrance *exposure* (not the *dose*) that is the commonly quoted comparison figure. In mammography, the *average glandular dose* is the standard measure of dose. The distribution of radiation dose in CT is markedly differ-