Chapter 1

Instrumentation – CT

Andrew Maidment and Drew A. Torigian

1.1 Introduction to CT

Computed tomography (CT) uses x-rays to generate cross-sectional tomographic images and differs from conventional projection radiography where a snapshot image is taken with a fixed geometry between the x-ray source, the object, and the image receptor. In projection radiography, structures at different depths in the imaged object are superimposed in 2-D, whereas in CT, the 2-D superposition of structures is virtually eliminated. Advanced 3-D-rendered CT images can also be created as needed when contiguous tomographic slices are acquired by a helical acquisition.

CT provides excellent contrast resolution that is far superior to plain-film radiography. However, CT does have inferior spatial resolution relative to plain-film radiography, and does impart substantially more of a radiation dose to patients. Nonetheless, continual advances in CT technology over time have allowed for its practical application in diverse clinical settings, such as with CT colonography, CT angiography, and CT urography for detailed evaluation of colon, vasculature, and urothelial system, respectively.

The CT image is created, in very basic terms, by reconstruction from a large number of *projections*, each of which consists of many *rays* or measurements of x-ray transmission, through a patient. In effect, CT attempts to determine the internal anatomy of the patient by using x-ray projections created at multiple different angles. A typical medical CT scanner is shown in Figure 1.1. In this configuration, the patient lies on the table, and is moved into the opening in the gantry to the desired position. Unseen but inside the gantry are the x-ray tube and detectors, which rotate around the patient. The detectors measure the transmission projections at each of the various angles as the x-ray tube rotates around the patient. After the projections are acquired, a computer is used to reconstruct slices through the patient. The images are then sent to a computer workstation for clinical review.

1.2 Concepts of Attenuation and CT Number

The building blocks for a CT image are a series of transmission measurements of individual rays through the patient, utilizing a thin pencil-like x-ray beam and detectors, as shown in Figure 1.2. The detectors measure the intensity of the transmitted beam, I. The relative transmission is defined as the ratio $-\ln(I / I_0)$, where I_0 is the intensity of the beam entering the patient and is equal to the product of two properties of the patient, the x-ray attenuation coefficient, μ, and the thickness, t.

As shown in Figure 1.2, when x-rays pass through a patient, they are reduced in number or *attenuated*. The x-rays are attenuated via two primary mechanisms. The first of these is *Compton scattering* where the x-ray photon interacts with a free electron and is redirected with reduced energy out of the beam. Such photons are referred to as scattered radiation and are of concern when considering scanner design and image quality. The second attenuation mechanism is the *photoelectric effect*, in which the photon is completely absorbed and thus removed from the beam.

The Compton effect scales with electron density; this results in the Compton effect being nearly independent of the atomic number, Z, of the material. The photoelectric effect, on the other hand, is very strongly dependent on atomic number, where the probability of the photoelectric effect occurring increases proportional to Z^3. Thus, it is the photoelectric effect that is responsible for differentiation between different material or tissue types, when densities are equal. The probability of a Compton effect occurring is almost independent of x-ray energy over the diagnostic range, whereas the probability of a photoelectric effect occurring decreases proportional to the photon

Figure 1.1 A typical CT scanner consisting of a gantry housing the rotating x-ray source and detector arrays. The patient table is moved into the aperture in the gantry for imaging.

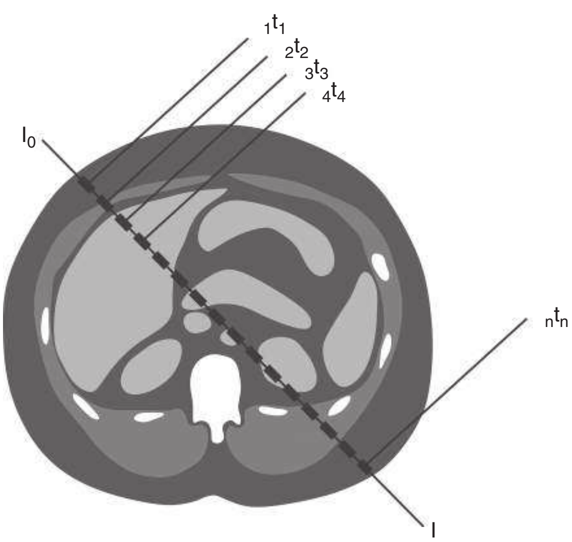

Figure 1.2 X-ray transmission measurements in CT are performed for each ray that passes through the body. The incident x-rays (I_0) are reduced in number after they have passed through the patient (I) due to attenuation by the various tissues of the body ($\mu_n t_n$).

energy cubed. Thus, the Compton effect dominates for high-energy CT scans and the photoelectric effect dominates at low energy.

The linear attenuation coefficient, μ, is difficult to work with in practice. A typical result, for example, might be 0.191 cm^{-1} in one region and 0.172 cm^{-1} in a region next to it with good contrast. A quantity called the CT number (in Hounsfield units (HUs)) is therefore used to make more manageable numbers and to reduce their energy dependence by normalizing them to water. The equation relating CT number (in HU) to μ is as follows:

$$\text{CT\_number} = 1000\frac{\mu_x - \mu_{H_2O}}{\mu_{H_2O}}$$

where μ_x is the linear attenuation coefficient of the voxel of interest, and μ_{H_2O} is the linear attenuation coefficient of water. A difference in CT number of 1 HU corresponds to a difference of 0.1 percent in contrast. The CT numbers for the attenuation coefficients mentioned earlier, 0.191 and 0.172 cm^{-1}, are 0 HU and –99 HU, respectively, roughly the difference between water and fat.

Normalization to water is useful since the body is principally composed of water and is in the middle of the range of CT numbers typical for a scan of a patient (see Table 1.1). It also reduces much of

Table 1.1 Approximate CT Number Ranges of Human Tissues

Material/Tissue	CT number ranges (in HU)
Bone/calcification/metal/ concentrated iodinated contrast material	> 150
Acute hemorrhage	50–90
Soft tissue	20–80
Water	0
Fat	−20 to −150
Lung	−400 to −1000
Air	−1000

the dependence on beam energy. However, atomic composition differences of materials can cause some variation in this rule. Thus, while these differences can usually be ignored over the energy range used in CT, some materials such as iodinated contrast agents are substantially better visualized at lower energies because of the energy dependence of their attenuation properties.

1.3 Principles of CT Image Acquisition, and Helical CT

In all modern scanners, many *rays* are acquired simultaneously, using a *fan beam* configuration as shown in Figure 1.3. A fan beam is a wide, wedge-shaped beam that is thin in the direction perpendicular to the slice (typically called the z direction). A *projection* is composed of the set of rays in the fan beam. An *axial slice* is reconstructed using a series of projections that encircle the patient or object of interest. Thus, the number of transmission measurements required to reconstruct a single slice is given by the number of rays in a single projection multiplied by the number of projections used in the slice.

All modern CT scanners use *slip ring* technology, that is, sliding electrical contacts, to power the x-ray acquisition system continuously, as opposed to older systems that were physically connected by wires, limiting image acquisition to single axial slices at a time. The advent of slip ring technology in the 1990s led to a new method of acquisition called helical (or spiral) CT. In helical acquisition mode, the table moves continuously through the gantry while the x-ray tube rotates around the table, and x-ray transmission data are collected. This continues until the total volume of interest in the patient has been scanned. No delays

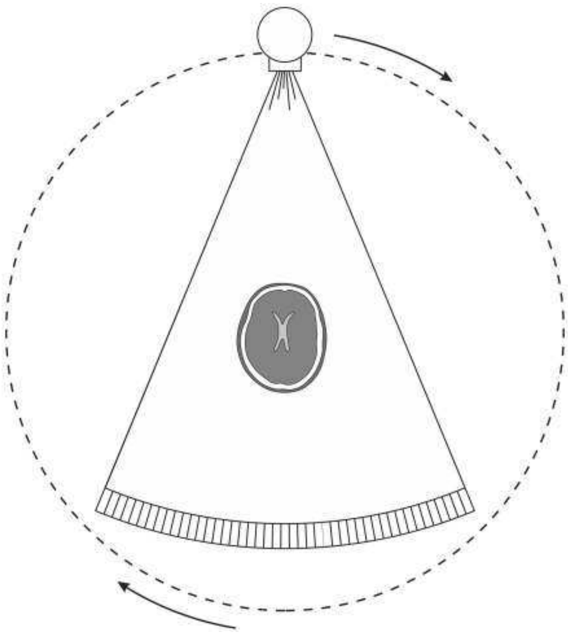

Figure 1.3 Fan beam image acquisition geometry in CT.

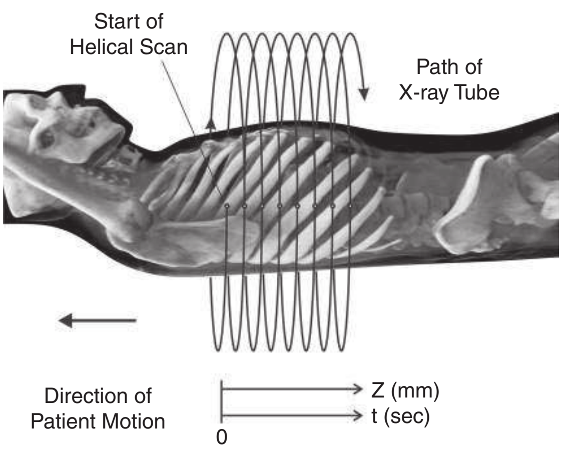

Figure 1.4 Illustration of the helical trajectory of the x-ray tube and detectors around a patient used in helical CT.

between "slices" are required. Tube heating capacity or the ability of a patient to breath-hold may impose some limits on the total acquisition time, however. This helical approach to data acquisition is volumetric, as the data are continuously collected over a volume of tissue, in contradistinction to the axial CT acquisition approach of obtaining single slices one at a time. Figure 1.4 demonstrates the helical trajectory that the x-ray beam makes around a patient in the volume acquisition mode.

Generally, the operator does not directly control the table motion when setting up a helical acquisition. A new parameter is introduced that the operator does adjust in this type of scanning that relates to the tightness of the helix. This parameter is called the pitch and is defined as the amount of table movement per 360° rotation of the x-ray tube divided by the collimator width. For example, if the collimators are set for a 5 mm slice and the pitch is 1.5, the resulting table movement will be 7.5 mm per rotation.

If the data from a 360° segment of a helical CT scan are reconstructed directly without modification, the resulting image will have major artifacts since the patient is undergoing movement by the table throughout the duration of the helical CT scan. It is thus necessary to compensate for this motion by first constructing quasi-planar datasets from the volumetric dataset. The image may then be reconstructed with the standard method that is used for slice-by-slice acquisitions. The construction of quasi-planar datasets is called z interpolation, since the axis along the gantry is the z axis and the tomographic planes are typically constructed perpendicular to this axis.

The most conceptually straightforward z interpolation method is the 360° linear interpolation (360° LI). For a desired slice position z, the projection, $P_z(\alpha)$, at a particular angle α is constructed from the two projections nearest z at the same angle α. One projection will be from a position slightly less than z, $P_j(\alpha)$, and the other will be from one slightly greater than z, $P_{j+1}(\alpha)$. $P_z(\alpha)$ is then constructed from these two projections according to the formula:

$$P_z(\alpha) = (1-w)P_j(\alpha) + wP_{j+1}(\alpha)$$

where w is a weighting factor that is linearly proportional to the relative distance between z and z_j, the z position where $P_j(\alpha)$ was acquired.

While 360° LI is capable of creating high-quality axial images, the resolution in the z direction is degraded. Other z interpolation methods are therefore used which take advantage of the observation that opposing projections essentially demonstrate the same features. In this way, it is possible to construct another (virtual) helix by shifting the original one by 180°. The original and 180° shifted helices can then be used to perform the z interpolation. This interpolation (180° LI) has better resolution in the z direction. Altogether, there are numerous alternative z interpolation methods that can also be employed.

1.4 Multislice CT and Dual-Energy CT

The benefits of helical acquisition are complemented by the recent development of multislice CT (MSCT) (or multidetector row CT (MDCT)) technology. While it is possible to perform multislice axial CT acquisitions, the primary use is in helical CT acquisition. The key component of MSCT is the detector array, which has several rows of detector elements that extend in the z direction, allowing for the acquisition of multiple simultaneous channels of data per gantry rotation (Figure 1.5). The width of a detector element in the z direction depends, in general, on the particular row within which it is located. Different vendors use different combinations of detector element widths of different rows to allow one to acquire varied numbers of slices and slice thicknesses as needed. At present, CT scanners with up to 320 channels of data acquisition are commercially available.

MSCT offers many advantages to single slice CT (SSCT). Scans can be completed more quickly, thus minimizing or eliminating many motion artifacts that occur due to bulk patient motion or from physiological organ motion due to respiration, cardiac motion, or bowel peristalsis, and allow for greater z axis coverage of bodily regions of interest during a CT examination. This gain in image acquisition speed and z axis coverage is particularly important for patients who are not able to hold breath for sufficiently long periods of time, and for patients who are potentially clinically unstable such as in the clinical setting of acute traumatic injury. The quicker scan times also make it possible to perform multiphase CT examinations more efficiently following the intravenous administration of contrast material such as with triple phase liver CT (where noncontrast, arterial phase postcontrast, and venous phase postcontrast images are obtained), 4-D cardiac CT, CT angiography (CTA), and CT urography (CTU). Furthermore, MSCT allows one to obtain scans with a submillimeter section thickness, providing high-resolution images that are isotropic, that is, composed of voxels that are cubic in shape. This minimizes the importance of patient positioning; obviates the need to obtain axial, coronal, and sagittal planes directly, as such images can be retrospectively reconstructed from previously acquired datasets; optimizes CT image postprocessing techniques (discussed in further sections) that can be useful for visualization and interpretation of the image data; and improves quantitative measurement of lesion volumes.

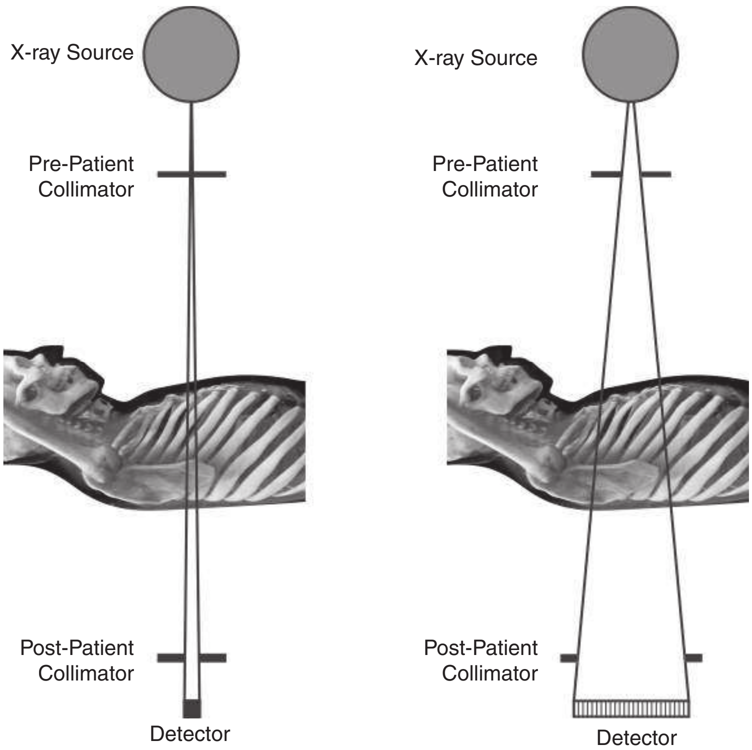

Figure 1.5 Single slice CT (SSCT) scanners (a) have a single row of detector elements, whereas multislice CT (MSCT) scanners (b) have multiple rows of detector elements that extend in the z direction. Note greater z axis coverage by x-ray beam in MSCT scanners compared to SSCT scanners. Also note two sets of collimators; the pre-patient collimator ensures that the x-ray beam striking the patient is as small as possible, while the post-patient collimator defines the width of the beam incident upon the detector, and eliminates scatter.

Another advance in CT technology is dual-energy CT (DECT), which is performed by simultaneously operating the two x-ray tubes of a dual-source CT scanner at different voltage levels (e.g., 80 kV and 140 kV) during patient imaging. Data obtained by DECT may be useful to provide functional information regarding the tissue composition of structures, based on the differential attenuation of the two separate x-ray beam spectra by different tissue components (e.g., calcium, iodine, soft tissue, water, and fat). Such information may potentially be useful for improved lesion characterization, improved quantification of contrast enhancement, and assessment of the chemical composition of renal calculi.

1.5 CT Hardware Components

A CT gantry is shown in Figure 1.1. The aperture of the gantry is the round space through which the patient passes to undergo CT imaging. The patient table, or couch, provides the physical support for the patient and must be strong enough to support overweight or obese patients, yet have sufficiently low attenuation properties so that it does not interfere with the imaging. The table must also be able to move in both vertical and horizontal directions. Movement in the vertical direction is designed to allow patients to access the table easily, yet ensure that the anatomical region of interest is centered in the gantry aperture for imaging. Movement in the horizontal (z axis) direction is necessary to image the desired region of the patient in either axial or helical modes.

The x-ray system (within the CT gantry) consists of an x-ray generator, x-ray tube, collimators, filters, a detector assembly, and associated electronics. The x-ray generator is responsible for supplying the correct power to the x-ray tube. Virtually all CT generators are of the high-frequency type, as high-frequency generators are more compact than other types of generators, allowing for placement inside of the rotating assembly instead of outside as was necessary with older CT systems. This was another key technical innovation in the 1990s that enabled the development of helical CT.

The production of x-rays occurs in the x-ray tube. The voltage potential provided from the x-ray generator, typically between 80 and 140 peak kilovoltage (kVp), is applied across the tube. The anode is made of a tungsten alloy that is chosen for its relatively high atomic number and high melting temperature, resulting in high x-ray output. The z-flying focal spot (zFFS) technology, available in some CT scanner models, allows for controlled motion of the x-ray tube focal spot in the z direction, leading to a doubling of the number of simultaneously acquired slices in MSCT with further improvement in longitudinal spatial resolution.

The x-rays produced in the x-ray tube are filtered before irradiating the patient. Filtration preferentially attenuates the lower energy x-rays present in the beam since the attenuation coefficient decreases with increasing energy. Filtration is necessary for several reasons. First, a harder, more penetrating beam is desired for CT to reduce patient radiation dose. Second, increased filtration results in an x-ray beam that more closely approximates a mono-energetic source, reducing certain artifacts. However, this must be balanced with the needs of tube loading and of obtaining sufficient x-ray fluence. Finally, additional filtration is provided for beam shaping. This last filter is often referred to as a "bow-tie" filter since it is shaped somewhat in the shape of a bow-tie, where the filter is thinner in the center and is increasingly thicker toward the periphery. This design is used to compensate for the typical ovoid cross-section of the patient, where total x-ray attenuation by the patient's body will be less at the periphery than at the center.

After filtration, the useful x-ray beam is shaped by collimators. These serve to define the size of the x-ray beam. In one dimension, they determine the size of the fan beam. In the other dimension, they determine the slice thickness of the beam. There are normally two sets of collimators, prepatient and postpatient (or predetector) (Figure 1.5). The prepatient collimators are physically attached to the x-ray tube and rotate with it, whereas the postpatient collimators are fixed in a position relative to the detectors, and serve to sharpen the profile further. These also absorb much of the scatter radiation produced in the patient/object undergoing imaging, preventing scatter from reaching the detectors and degrading image quality.

After passing through the postpatient collimators, the x-rays are incident upon the detectors. The detector is responsible for converting the x-rays into an electrical signal. It is important that CT detectors have the following characteristics: high efficiency, quick response time, stability, high reproducibility, and large dynamic range. High efficiency assures that most of the useful x-rays contribute to the CT image. This helps to minimize patient radiation dose and image noise. A quick response time is necessary so that afterglow, or residual signal after detection, is minimized, since afterglow can lead to the degradation of spatial resolution. Stability and high reproducibility of detector response are critical since the reconstruction process hinges on the comparison of x-ray intensity with and without the attenuating medium. Detector instability will lead to false readings and the need for more frequent calibrations. The large variations in x-ray intensity seen by the detector make a large dynamic range essential.

The data acquisition system is responsible for taking the analog signal from the detector, converting it to digital format, and sending it to the CT system computer. The computer is responsible for performing the image processing after receiving the data from the data acquisition system. As the data is received, it undergoes preprocessing which includes such tasks as normalization based on the detector calibration. This is then followed by image reconstruction.

1.6 CT Image Reconstruction

After the x-ray data have been collected, they are reconstructed to form an image of the patient or object that was scanned. The raw projection data would produce severe artifacts if reconstructed as is. Therefore, the projection data must first undergo some preprocessing operations before the reconstruction process begins. The first preprocessing step is to normalize the measured intensity to the assumed nonattenuated intensity. The specifics of how this is handled depend upon the scanner design. Preprocessing corrections must also be made since attenuation is not truly linearly related to path length as expected in the ideal case. Scattered radiation, nonlinear detector response, and beam hardening effects all contribute to this.

Filtered backprojection (FBP) is the most common reconstruction method currently in use. In FBP, the projections are filtered prior to undergoing backprojection. The applied filter is designed to remove the intrinsic blurring arising from the backprojection process. Filtering causes a negative shadow to occur on both sides of the backprojected objects. These negative backprojections, when added together for all projections encircling the scanned object, cancel out the false positive contributions caused by blurring.

The operator of the scanner generally has the option of choosing from several reconstruction filters (also called kernels). These filters differ slightly in the

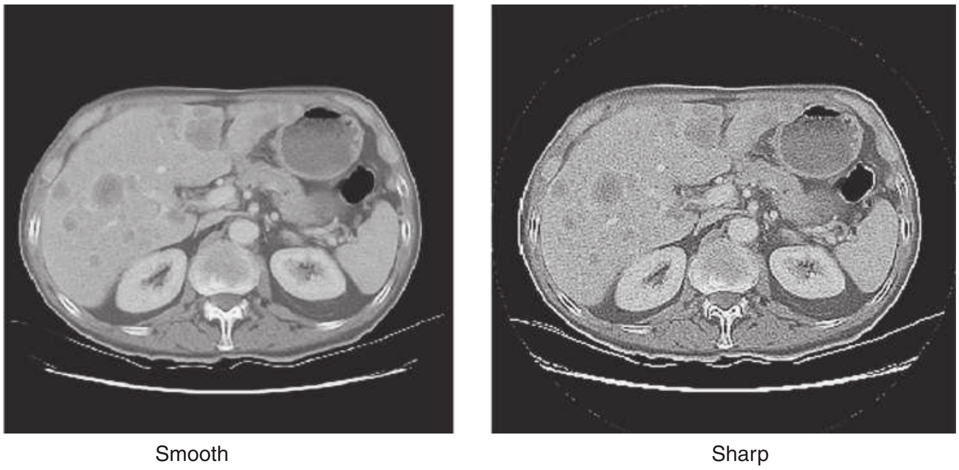

Figure 1.6 Effect of reconstruction filter (kernel) on CT image quality.

shape of the function used in the filtering process, which in turn influences image quality (Figure 1.6). For example, one filter might enhance spatial resolution at the expense of increasing image noise, whereas another might suppress noise at the cost of decreasing spatial resolution. The filter selected will depend on the information desired. For example, if one desires to detect low-contrast lesions in a patient, a "smoothing" or "soft tissue" filter might be selected since low-contrast lesion detectability is improved given the presence of less noise.

Iterative methods of image reconstruction are beginning to become more popular based on the availability of increasingly powerful computers. In iterative methods, an initial reconstruction of the patient is created. Simulated projections through this reconstruction are then compared to the actual projections. The differences between those projections are determined and used to make a revised estimate of the voxel values. This process is repeated until the differences become sufficiently small as to not further influence the reconstruction. Iterative methods have the advantage that the physics of image acquisition can be modeled, so as to allow for noise reduction or radiation dose reduction.

1.7 Advanced CT Image Post Processing and Display Techniques

In some cases, axial slices are not sufficient to display the imaging features of lesions of interest. In those instances, multiplanar reconstruction (MPR) can be used to improve the depiction of such features. It is possible to display sagittal, y-z plane, or coronal, x-z plane, images rather than traditional axial, x-y plane, images. Oblique planes of section may also be created as needed given the volumetric nature of the x-ray dataset. In fact, MPR images do not even have to be flat, but instead may be curved to follow a particular anatomical structure or lesion of interest. Furthermore, slice thickness may be adjusted accordingly as needed.

The data can also be displayed with volume rendering (VR) (Figure 1.7). This technique uses all of the data in the dataset rather than a subset of the dataset in a given plane or intensity range. Before VR can begin, the data are first preprocessed. Each voxel is assigned a brightness level or color and a transparency level based on its CT number. The image is then built by simulating rays passing through the dataset. The viewpoint is often external to the patient, although it can also be inside the patient. This display technique is useful for purposes of CT virtual endoscopy, where the viewpoint is located from inside the lumen of a vessel, bowel loop, or airway of interest.

Another display technique, which is commonly used in CT angiography applications, is called maximum intensity projection (MIP) (Figure 1.7). In this method, a ray passes through the volume data as it does in other volume-rendering methods. The corresponding pixel is assigned a value equal to the value of the maximum voxel along the ray. This technique has the advantage of preserving the attenuation information in the data, and produces high-contrast images.

Figure 1.7 Volume rendering (VR) (left), maximal intensity projection (MIP) (middle), and minimal intensity projection (MinIP) (right) display techniques are demonstrated through use of a CT angiography dataset.

One other display technique is called minimal intensity projection (MinIP) (Figure 1.7), where the corresponding pixel is assigned a value equal to the value of the minimum voxel along the ray. This technique may be useful to display lesions involving air-filled structures such as the tracheobronchial tree, lungs, and bowel to best advantage.

1.8 CT Image Quality and Radiation Dose

Spatial resolution is one of the key parameters used to describe image quality. It is a measure of how sharp, or conversely how blurred, the image is. The better the spatial resolution, the closer small high-contrast objects can be placed next to each other and still be discerned as separate entities. The limiting spatial resolution is related to the size of the smallest objects that can just barely be visualized, and is usually quoted in terms of line pairs per centimeter (lp/cm). Typical values for axial CT are around 12 lp/cm. As previously mentioned, the filter used is chosen in part based on its resolution properties.

Spatial resolution in the z axis direction, perpendicular to the tomographic plane, is primarily determined by the detector element size, pitch, and z interpolation algorithm. Smaller detectors or a smaller pitch result in better resolution. The z interpolation method takes data from a region and generates new data to reconstruct a slice at the center of that region. In very general terms, the larger the region used, the broader the sensitivity profile becomes and the worse the spatial resolution is. For example, a 360° algorithm has poorer resolution than a 180° algorithm.

Noise on the images is manifest as random variations of signal in otherwise uniform regions of the patient. The noise in the image limits the ability to visualize low-contrast objects and is measured by determining the number of x-rays used to create a single slice of the patient. Thus, the x-ray acquisition parameters (such as tube voltage, tube current, and exposure time), gantry rotation time, pitch, slice thickness, imaging field of view, image matrix (typically 512 × 512), z interpolation method, and reconstruction algorithm affect image noise, as well as spatial resolution and radiation dose levels. For example, a decrease in tube current alone will increase noise and decrease radiation dose without affecting the image contrast or CT numbers, whereas a decrease in tube voltage alone increases noise and decreases radiation dose while altering CT numbers and improving image contrast. In general, there is a trade-off between spatial resolution and noise, where improvements in resolution come at the cost of increased noise.

The radiation dose to the patient that is necessary to produce a given noise level will depend on the design of the scanner. Of the parameters that the operator of the CT scanner normally controls, tube current (in milliamperes (mA)) and exposure time (in seconds (s)) are two important ones that affect dose, as radiation dose is linearly dependent on each of these two parameters. Radiation dose also depends on kVp since the number of photons produced changes in proportion to the square of the change in kVp.

The dosimetric quantities used in CT can be divided into doses to standard physical test objects (called phantoms) and patient doses. At the present time, the doses measured from physical phantom studies are used to infer patient doses. However, this practice is under review, and improved patient dose estimation methods are undergoing research evaluation.

The radiation dose to standard phantoms is measured in terms of the CT dose index ($CTDI_{100}$). The $CTDI_{100}$ is typically measured at the surface and the center of the CT dose phantom under axial scanning conditions. These dose values are combined to create an estimate of the dose averaged over the phantom cross-section.

$$CTDI_w = (1/3)*CTDI_{100}(\text{center}) + (2/3)*CTDI_{100}(\text{surface})$$

To estimate the dose under a helical acquisition, the dose over the volume of the phantom is calculated as

$$CTDI_{vol} = CTDI_w/\text{Pitch}$$

The CTDI is indicative of the dose of a slice through the phantom, typically in units of milligray (mGy). To calculate the dose of a CT procedure, it is necessary to know both the dose per slice and the number of slices acquired. This is quantified by the *dose length product* (DLP), which is calculated as the product of the $CTDI_{vol}$ and the length of the scan, *l*.

$$DLP = CTDI_{vol}*l$$

The DLP is typically in units of mGy•cm. A typical $CTDI_{vol}$ is in the range of 10–20 mGy. For a 30 cm scan, the DLP would then be 300–600 mGy•cm.

A gross estimate of the effective dose, E, to patients, typically given in units of millisieverts (mSv), can be calculated from the DLP with knowledge of the body parts scanned.

$$E = DLP * k$$

where *k* is given in Table 1.2. The effective dose is a summary measure of the risk of inducing cancer from the radiation dose of the CT scan. Calculated as described here, this is the effective dose to the standard phantom based on the clinical technique. The actual effective dose to a patient is dependent upon the actual size and age of the patient. Patient dosimetry is an active area of research at the current time, and more representative patient dose measures are anticipated.

Table 1.3 lists the typical effective radiation doses for common CT scan procedures. For comparison, the average background radiation dose for the United States is also provided. The radiation dose from a typical CT scan is comparable to the background radiation dose obtained over the course of one to three years. However, in any discussion of radiation dose, one must generally compare the risks of exposure to the radiation dose to the benefits of the potentially clinically useful diagnostic information that may be obtained from CT. Careful selection of patients to be imaged with CT should be a priority of the radiologist and referring physician in order to avoid unnecessary radiation dose exposure.

Minimization of radiation dose during CT scanning of patients is encouraged, and may be achieved in various ways. Decreases in tube voltage, current, or exposure time, and increases in pitch may be useful

Table 1.2 Conversion Factors between DLP and Effective Dose

Scan location	K
Head	0.0023
Chest	0.017
Body	0.015
Abdomen-Pelvis	0.017
Pelvis	0.019

Table 1.3 Common CT Scan Effective Radiation Doses

Average US background radiation	~3–3.6 mSv / yr
Head CT	1–2 mSv
Chest CT	5–7 mSv
Abdomen CT	5–7 mSv
Pelvis CT	3–4 mSv
Abdominopelvic CT	8–11 mSv
Low-dose chest CT	1–2 mSv

approaches to decrease CT radiation dose while maintaining image quality. Automatic tube current modulation (also called automatic exposure control (AEC)), which automatically modulates tube current in both angular and longitudinal directions in response to the size and attenuation characteristics of the body parts scanned, can also be useful to reduce patient dose exposure by 20–60 percent while maintaining predefined image noise or image quality characteristics. Organ-based tube current modulation (TCM), in which tube current is decreased as the x-ray tube passes over the anterior surface of the body and increased over the posterior surface of the body, may also be used to decrease dose to anterior superficial radiosensitive organs such as the breast, thyroid gland, and eye lens by up to 50 percent without compromising image quality. Newer reconstruction techniques, such as adaptive statistical iterative reconstruction (ASIR) and model-based iterative reconstruction (MBIR), have shown promise to reduce dose. Through use of ASIR, CT dose can potentially be reduced by up to 65 percent in adults without compromising image quality. Similarly, model-based iterative reconstruction (MBIR) can allow for up to 80 percent reduction of CT dose, although the prolonged processing time may limit its routine use in clinical practice. Finally, use of alternative nonionizing radiation imaging technologies such as ultrasonography (US) and magnetic resonance imaging (MRI) can reduce radiation dose exposure to patients.