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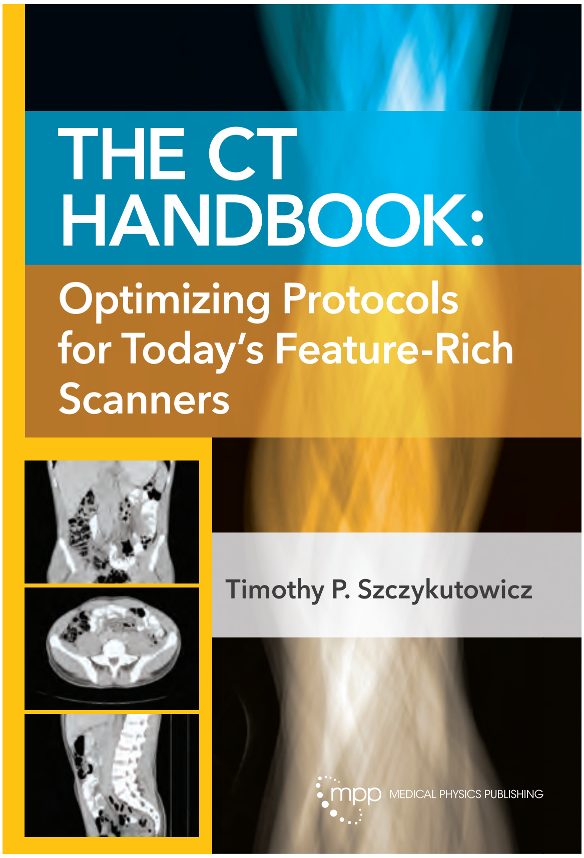

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CHAPTER LIST:

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1

Introduction to CT

"It is possible that this technique may open up a new chapter in X-ray diagnosis. Previously, various tissues could only be distinguished from one another if they differed appreciably in density. In this procedure absolute values of the absorption coefficient of the tissue are obtained."

–Sir Godfrey Hounsfield in his original paper on CT [1]

Key References for This Chapter

The seminal paper introducing CT as a modality: "Computerized transverse axial scanning (tomography): Part 1. Description of system" by Hounsfield [1].

A more recent paper describing CT's role in medicine: "Physicians Views of the Relative Importance of Thirty Medical Innovations" by Fuchs and Sox [2].

1.1 How Does CT Work

Computed tomography (CT) has enjoyed an ever-expanding number of clinical applications since its debut in the 1970s [1,3]. Throughout the history of CT, clinical applications have been enabled by technological improvements motivated by clinical needs. This cycle of innovation's response to need has transformed the CT scanners of the 1970s into the modern scanners we use today. The predominant modern scanner geometry is the so-called generation III, in which an x-ray tube and a detector rotate around a patient on a gantry with the patient moved into the x-ray beam via a translatable couch. The physical appearance of the most common multi-detector CT (MDCT) scanners used in hospitals and imaging centers today is more similar to magnetic resonance imaging (MRI) scanners relative to c-arm-based CBCT scanners. However, the physics governing the data acquisi-

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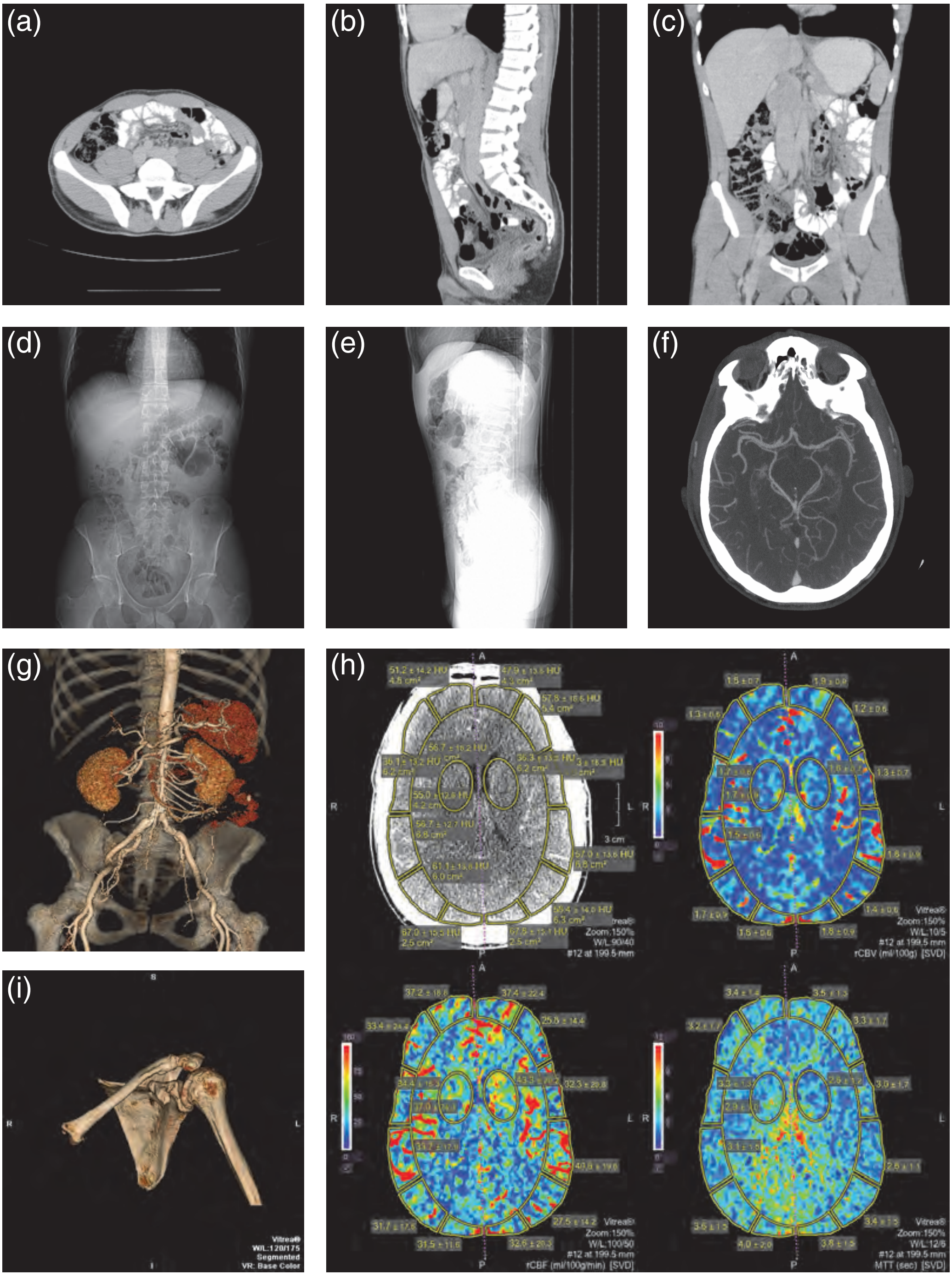


Figure 1.1 Example CT images. (a) Axial CT image of the abdomen. (b) and (c) Sagittal and coronal reformats of the patient shown in (a). (d) and (e) Anterior–posterior and lateral CT localizer radiographs of the patient shown in (a). (f) MIP image of the brain in a contrast-enhanced exam shown in the axial plane. (g) Surface-rendered image showing abdominal organs, major arteries, and bony anatomy. (h) Perfusion map images. (i) Surface rendering of the shoulder joint. Most commonly, images like (a)–(e) are made for every diagnostic CT exam. Additional processing—and usually a specific diagnostic or surgical planning purpose—is needed to realize the images shown in (g)–(i).

tion and image reconstruction of MDCT is much more closely related to c-arm-based CBCT than MRI. While the majority of CT scanners are used for general diagnostic radiology applications, CT also has several niches within today's health care system. This will be elaborated on in section 1.3. In addition to health care, CT is also used in a number of industrial settings, from scanning baggage to looking for defects in metal parts.

All the different modalities of CT have one thing in common: they all need to measure the amount of x-rays passing through a patient from a wide range of angles. In this way, CT is like acquiring radiographs of an object from many different projections. The range of angles needed to reconstruct a tomographic slice depends somewhat on the way the data is collected, but varies between about 200 and 360 degrees. Since the most efficient way to collect such data is to take measurements in a circle (this shape minimizes the x-ray source trajectory length), most CT modalities are shaped like circles or parts of circles.

CT images typically represent maps of linear attenuation coefficient. These "maps" of the patient are the images we collect with CT. However, with the now widespread use of dual-energy CT, this is not always the case (see chapter 9). Using dual-energy CT, images of electron density, effective atomic number, and material density (i.e., iodine or bone) may be obtained. Additionally, perfusion-related maps (i.e., blood volume, blood flow, and blood arrival time) can be obtained from CT scans. In industrial settings, where only one material is imaged, CT image values can be directly correlated with material density. At a high level, however, all images coming from a CT scanner can be thought of as being derived from maps of linear attenuation coefficient. For perfusion metrics, linear attenuation maps of the same anatomy acquired at different time points are fed into algorithms that output perfusion metrics like mean transit time, cerebral blood flow, and cerebral blood volume. For dual-energy CT, linear attenuation maps representing different energies are fed into algorithms that can produce material density images, increase image signal-to-noise ratio, reduce artifacts, or suppress iodine content. Figure 1.1 displays a wide range of images that can be output from a CT scanner. In addition to images of patients, CT scanners can output images containing data in graphical or numerical formats. For example, Figure 1.2 displays data resulting from a CT scan including ionizing dose and contrast dose information.

As mentioned, all flavors of CT acquire projection images of a patient or object from many angles. The detected signal with no object in the beam is measured *a priori*. Therefore, after an object is scanned, it is possible to calculate exactly how much of the beam was stopped, or attenuated, by the object by taking the ratio of the signal with no object present to the signal with an object present. This ratio is then transformed into a quantity referred to as the *sinogram*. The sinogram is then fed into a number of data correction processes before being used to reconstruct the final image. A more detailed description of this process—including the mathematics needed to robustly describe this process—is the topic of many excellent books in our community. The interested reader is referred to [4,5,6,7]. Figure 1.3 describes CT image reconstruction at a high level.

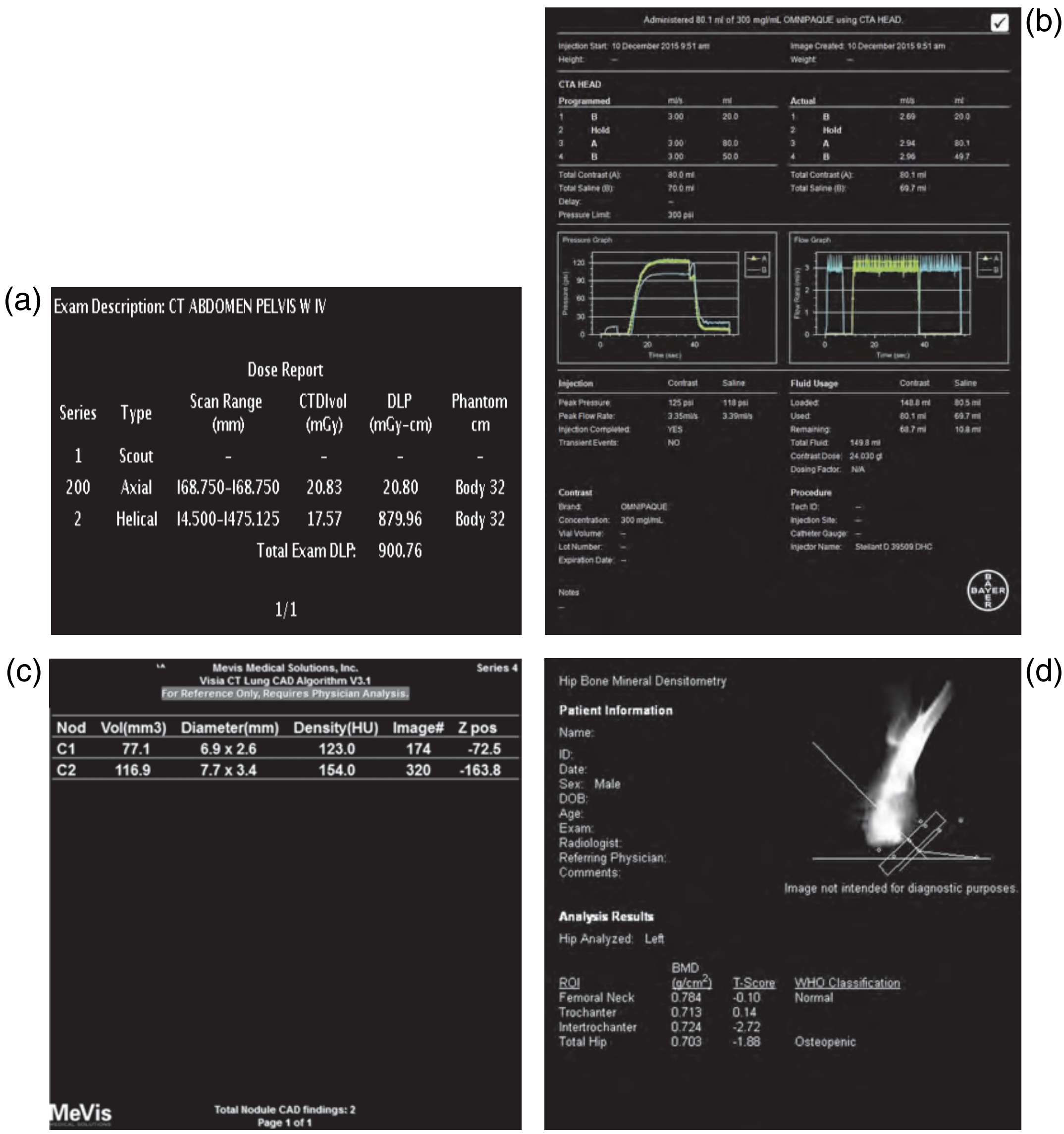


Figure 1.2 Example non-patient images commonly associated with CT exams. (a) "Dose slide" DICOM image containing ionizing radiation dose information in an image format. Newer MDCT scanners also send this information off scanner in the structured report "SR" format, as discussed in chapter 15. (b) Contrast dose injection information created by the power injector company and sent to PACS along with the image data from the MDCT scanner. The data in (b) will tell the interpreting physician important aspects on the delivery of contrast, like the amount of contrast agent and chaser used for the exam, when it was injected, how fast (i.e., flow rate) it was injected, if a test injection was performed, if two boluses had to be given (i.e., a repeat study was needed), or if a split bolus technique was used. (c) Shows the tabulated result from a computer-aided diagnosis (CAD) tool, in this case a tool identifying possible lung nodules. (d) Depicts an example bone mineral density qualification report summary.

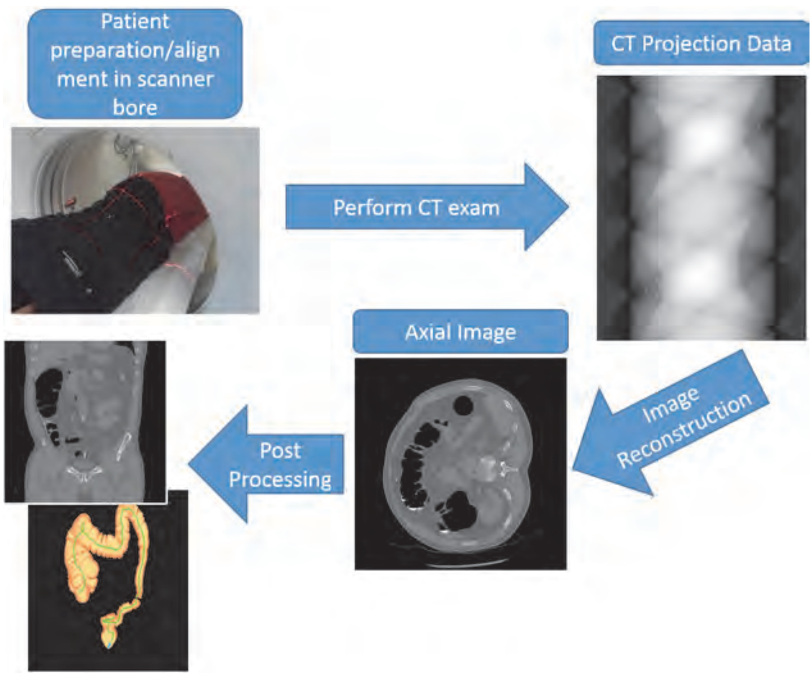


Figure 1.3 Steps involved in acquiring, processing, and reconstructing CT data. Each of these steps is composed of many sub-steps with optional or additional variations. In general, for all flavors of CT, a patient or imaging object is placed inside a CT scanner, an x-ray source illuminates the patient/object, and the amount of x-rays passing through the patient/object is recorded by a detector. The resulting projection data is referred to as a sinogram and is fed into a reconstruction algorithm that creates axial images slices. Depending on the application, many different versions of the axial image slices may be made, and some of the axial slices may be used to create views of the patient/object in different imaging planes or post-processed to produced volume-rendered datasets.

1.2 Data Collection

Across the spectrum and within different flavors of CT, it is possible to have multiple modes of data acquisition processing and reconstruction. The modes discussed in this section will all likely be found on most MDCT scanners, whereas other CT flavors usually only implement one or two of them. Some of the more exotic flavors of CT (usually non-clinical or industrial, but reserved for research purposes) use methods not described here (for example, synchrotron-based CT methods).

For the main data acquisition modes of axial and helical/spiral, Figures 1.4–1.8 are used to describe the x-ray source trajectory during data acquisition. We highlight how "good" each method is at acquiring projection data by noting how close each view angle is to the ideal cone angle of zero degrees as a function of view angle. A cone angle of zero is needed for perfect artifact-free image reconstruction [8]. In practice, however, we can use small cone angles in medical CT since the artifacts caused by small cone angles normally do not interfere with most soft tissue imaging applications. In general, as the detector size increases in the z-direction (i.e., for a patient lying on the CT couch, the superior/inferior direction) and as the pitch increases, more and more data must be used for image reconstruction that has a higher cone angle (i.e., more data is used for image reconstruction that was not acquired in the plane of reconstruction).

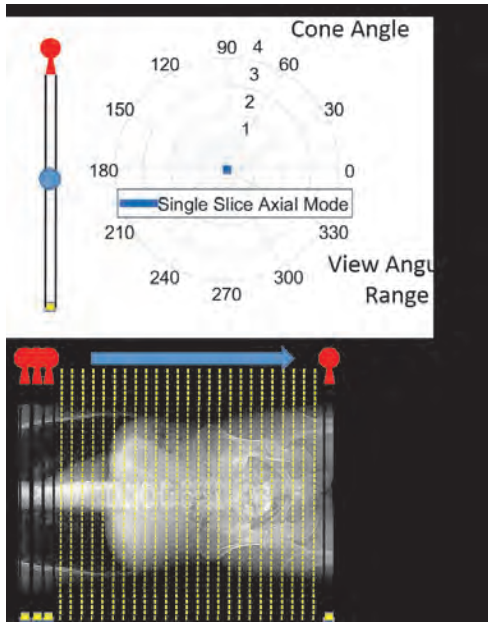


Figure 1.4 Data acquisition overview for single slice axial mode. In this mode, a point in the reconstructed volume (shown in the upper left of the figure with the blue circle) always lies along a line from the x-ray source to the detector during data acquisition. In other words, for all of the acquired view angles, the cone angle between the iso-ray and points inside the final reconstructed volume is zero. In order to scan a large z-coverage, as shown in the bottom of the figure, many scans must be acquired. Between each scan, the patient must be translated through the CT scanner. Compared to other data acquisition modes, this mode produces the best projection data, albeit it requires the longest time to scan and is the least dose efficient, making it impractical to meet the demands of today's clinical MDCT needs.

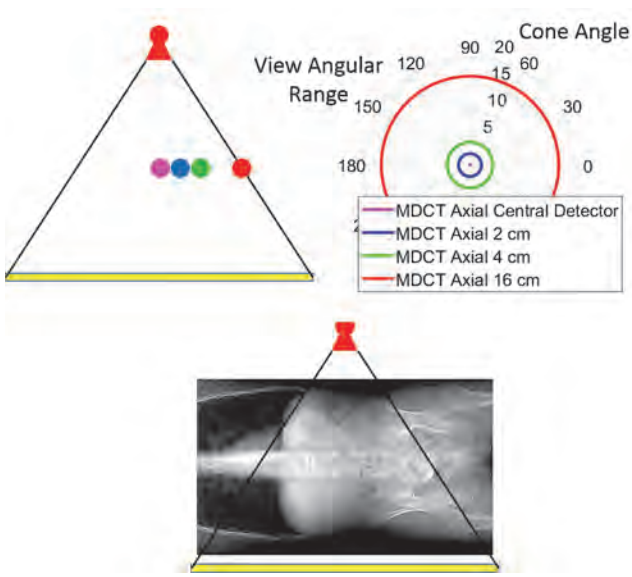


Figure 1.5 Data acquisition overview for wide axial mode. In this mode, only points in the center of the reconstructed volume (shown in the upper left of the figure with the purple circle) lie along a line from the x-ray source to the detector during the entire data acquisition. The blue, green, and red points always lie off the iso-ray line and, therefore, have a constant cone angle for all acquired view angles, as shown in the upper right of the figure. In other words, for all of the acquired view angles, the cone angle between the iso-ray and points inside the final reconstructed volume is only zero for points reconstructed in the center of the volume. The cone angle for points at 2, 4, and 16 cm from the center of the detector at the image plane are 2.13, 4.25, and 16.56 degrees, respectively (assuming a source- to-isocenter distance of 538 mm). In order to scan a large z-coverage, as shown in the bottom of the figure, only a single scan, or perhaps a few scans, need to be acquired, as these scan modes on modern MDCT scanners allow ≈16 cm to be scanned in a single rotation. Between each scan, the patient must still be translated through the CT scanner.

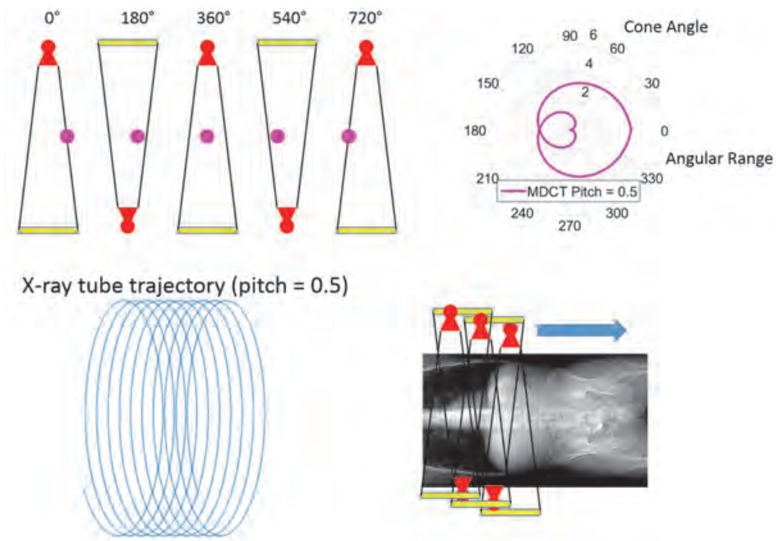


Figure 1.6 Data acquisition overview for MDCT-based helical scanning using a pitch of 0.5. In this mode, a point in the reconstructed volume (shown in the upper left of the figure with the purple circle) only lies along the iso-ray for a single projection. Since the patient is translated through the CT scanner during tube rotation, the sampling pattern (i.e., the cone angle between a reconstructed point in the volume and the iso-ray) changes as a function of view angle/time. In order to scan a large z-coverage, as shown in the bottom of the figure, the patient is just translated through the scanner while the x-ray tube is rotating until the volume has passed through the scanner. This translation is usually performed using a continuous constant velocity. The maximum cone angle for points within a reconstructed volume will depend on the collimation size. For a typical 4 cm detector array and a source-to-isocenter of 538 mm, the maximum cone angle would be 4.25 degrees.

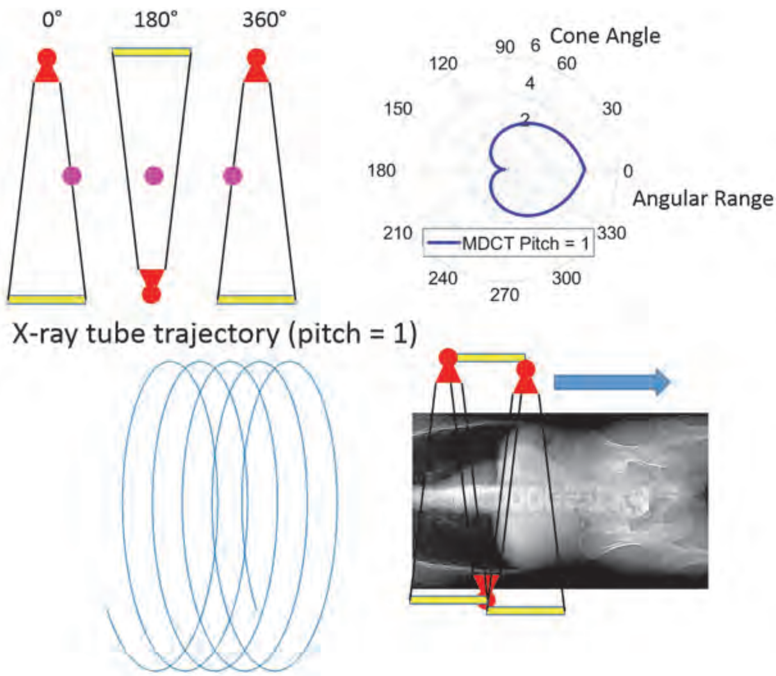


Figure 1.7 Data acquisition overview for MDCT-based helical scanning using a pitch of 1. This mode is quite similar to scanning with a pitch of 0.5, as shown in Figure 1.6. The major difference is in the amount of data acquired. With a pitch of 0.5, points inside the reconstructed volume get to "see" the x-ray tube and detector for two full rotations, as shown in Figure 1.6. With a pitch of 1, the scanner moves through sampling each point in the reconstructed volume in a single rotation, hence the difference between the sampling plots of this figure and Figure 1.6. In this mode, a point in the reconstructed volume (shown in the upper left of the figure with the purple circle) only lies along the iso-ray for a single projection, just like with a pitch of 0.5. Since the patient is translated through the CT scanner during tube rotation, the sampling pattern (i.e., the cone angle between a reconstructed point in the volume and the iso-ray) changes as a function of view angle/time. In order to scan a large z-coverage, as shown in the bottom of the figure, the patient is just translated through the scanner while the x-ray tube is rotating until the volume has passed through the scanner. This translation is usually performed using a continuous, constant velocity. The maximum cone angle for points within a reconstructed volume will depend on the collimation size. For a typical 4-cm detector array and a source-to-isocenter of 538 mm, the maximum cone angle would be 4.25 degrees. This maximum cone angle is the same as the pitch 0.5 case, but in the pitch 1 case, more data is acquired at higher cone angles relative to the pitch 0.5 acquisition.

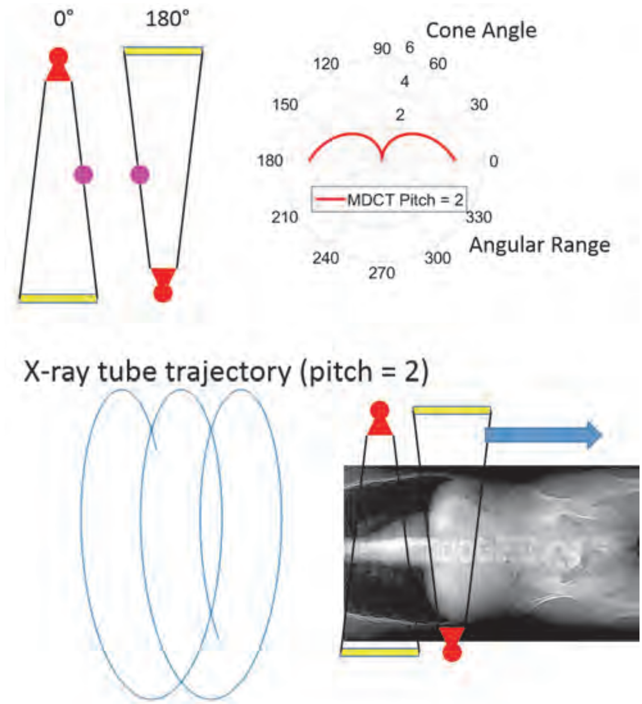


Figure 1.8 Data acquisition overview for MDCT-based helical scanning using a pitch of 2. In this mode, a point in the reconstructed volume (shown in the upper left of the figure with the purple circle) only lies along the iso-ray for only a single projection. Since the patient is translated through the CT scanner during tube rotation, the sampling pattern (i.e., the cone angle between a reconstructed point in the volume and the iso-ray) changes as a function of view angle/time. In order to scan a large z-coverage, as shown in the bottom of the figure, the patient is just translated through the scanner while the x-ray tube is rotating until the volume has passed through the scanner. This translation is usually performed using a continuous, constant velocity. The key issue with such a high pitch is that the x-ray source/detector only samples a given point for less than a short scan angular range's worth of data. This is the major source of artifact if a pitch higher than 1.5 is attempted without using a second x-ray source. The maximum cone angle for points within a reconstructed volume will depend on the collimation size. For a typical 4-cm detector array and a source-to-isocenter of 538 mm, the maximum cone angle would be 4.25 degrees. The highest possible pitch when using a single x-ray tube is 1.5, so this example data acquisition would not produce artifact-free CT images. Using two tubes, pitches higher than 1.5 are possible, and that is why they are used on high-end, dual-source MDCT scanners.

For most clinical CT applications, relatively large cone angles can be tolerated quite well since the human body only exhibits small changes in attenuation over the z-direction. Many industrial applications require imaging of high-contrast objects with very small image details changing rapidly over the z-direction. For these industrial applications, one often needs to use low pitch or small cone-beam data acquisition methods. The interested reader should refer to the Feldkamp/Davis/Kress (FDK) [9] image reconstruction method to understand the assumptions made in image reconstruction with a non-zero cone angle.

1.2.1 Axial/Sequential Mode

The key features of an axial data acquisition are that data is acquired with no translation occurring during tube rotation, and that only one rotation's worth of data is acquired. Figure 1.4 graphically depicts an axial data acquisition. In an axial operation mode, a single set of projection views are acquired at rest with respect to the plane of measurement. In other words, the patient couch remains stationary while the x-ray tube and detector spin around the patient. A single set refers to a scan angular range between about 200 and 360 degrees. After a single set of data is acquired for a given location within the object, the object may be translated (i.e., the couch is moved) to a new location so another scan can be acquired. When a large volume of data is desired (larger than can be acquired in a single rotation) multiple data acquisitions using axial mode can be "stitched" together in order to reconstruct a large scan range. Each set of projection views will allow for the reconstruction of one sub-volume of CT data. This volume is sometimes referred to as a "slab." The z-axis coverage for a single axial scan is limited by the width of the detector array. For example, a system with a 4-cm detector can only acquire 4 cm worth of data in a single axial pass. If a scan length of 16 cm was desired, four 4-cm axial scans would have to be acquired side by side, and the scanner's reconstruction engine would "stitch" the slabs together to create a single 16-cm scan volume.

1.2.2 Helical/Spiral Mode

In this operation mode, there is continuous translation of the patient/object with respect to the x-ray source and detector during data acquisition. This is by far the most common mode used in clinical MDCT scanning today, as it allows for fast scan times of large extents of patient anatomy. This scan mode is referred to as *helical* by some CT vendors and *spiral* by others.

Pitch values are unitless quantities. They are the ratio of how far the couch/table moves in one gantry rotation to the collimation (i.e., the beam width or collimator aperture at isocenter, which is equal to the number of acquired slices times the slice thickness and will normally be 2–8 cm for modern MDCT scanners) of the x-ray beam at isocenter in the z-dimension,

$$Pitch = \frac{\text{table increment in 1 gantry rotation}}{\text{beam width}}. \tag{1.1}$$

Pitch values range from ≈0.1 for retrospective cardiac and respiratory-gated exams to 3.2 for high-speed, dual-source scanning. Commonly, a pitch value near 1 is used for routine scanning.

Recently, some vendors offering wide detector coverages are suggesting the use of *axial mode* over helical/spiral. Helical/spiral CT is usually faster than axial scanning because of the wasted time (i.e., no scan data is being collected) it takes to move the patient/object between couch positions in order to "stitch" together multiple axial couch positions. When the detector is large, however, the time lost in moving the patient can be small since large volumes of the patient can be scanned in a single axial pass, reducing the number of patient/object translations.

1.2.3 Cine/Perfusion Mode

This mode is identical to axial mode, except multiple scans are acquired at the same location. The sampling of the data is usually adjustable, meaning one is free to adjust the time between scans and the total number of scans. Some scanners allow for one to prescribe axial scans with time delays as well. Therefore, the delineation between axial and cine is blurred on some systems.

1.2.4 Shuttle Mode

The word shuttle has a dictionary definition of "a form of transportation that travels regularly between two places." This is exactly what the couch of an MDCT scanner does in *shuttle mode*. Shuttle mode is a combination of a cine scan with either an axial or helical scan in which the same portion of the body is scanned multiple times. The scan range is larger than the beam collimation, so in order to cover the entire volume, the scanner has to move back and forth repeatedly. This mode is used to acquire perfusion data, usually of the brain, over relatively long periods of time (e.g., 45 to 70 seconds). Since most MDCT scanners only have a beam collimation (i.e., the width of the beam in the patient's superior to inferior dimension) of ≈4 cm, if one were to acquire a cine scan without moving the couch, only ≈4 cm of the patient could be scanned. If perfusion maps of the entire head are desired, multiple perfusion cine runs and multiple boluses of CT contrast would be needed. To avoid this, shuttle mode is used to alternate the couch between two positions—usually spaced apart by a distance equal to or double the maximum beam collimation—for each cine image acquisition. The resulting image volume contains images spread over a distance double to three times the thickness of the beam collimation in only a single administration of CT contrast bolus. The one drawback is that the temporal sampling for each bed position is about half that relative to a cine scan with no couch movement.

Most modern scanners employ ≈4 cm beam collimation, but double to triple this, 8–12 cm, is not wide enough to image the entire head. Luckily, however, the important regions of the brain to check for stroke-related abnormalities are contained within a subregion of the head most of the time. Wide collimation scanners, with up to 16 cm beam collimation, allow for almost the entire head to be imaged, mitigating the need for shuttle perfusion mode. However, surgical treatment for occlusions in the vessels located at the extreme superior regions of the head may not be possible, making coverages less than 16 cm sufficient in many cases. Section 13.1 treats scan modes and beam collimation specific to perfusion imaging of the brain.

1.2.5 CT Fluoroscopy

CT fluoroscopy (CTF) is the term used to describe a number of different methods for interventional imaging performed on MDCT scanners. Vendors usually sell MDCT scanners with CTF packages that offer software to guide CT interventions (e.g., needle trajectory guidance and planning) and hardware (e.g., real-time CT imaging with in-room monitors and scanner controls facilitating table/couch side control and visualization). Physician use of these systems, however, varies greatly, depending on local practice culture. Here is a breakdown of the major ways in which physicians use MDCT to perform interventional CT:

- ***CT guided:*** Axial/sequential or helical/spiral images are taken during a procedure with all staff outside the room. Imaging and intervention are interleaved throughout the procedure, giving the physicians a volumetric view of their progress without being in the room during x-ray exposure. In this scenario, the in-room controls of the CTF mode are not used to control the scanner, but may be used to review images. At the time of this writing, this is the most common form of CTF as it (1) offers the largest volume coverage and (2) essentially provides a negligible operator dose. Drawbacks of this method include (1) longer procedure times and (2) more patient dose due to the larger scan volumes typically used for this type of CTF.
- ***Real-time CTF:*** Cine images are acquired in real time with the operator next to the patient simultaneously controlling the scanner (i.e., using a foot pedal to activate x-rays) and progressing with the intervention (i.e., pushing a biopsy needle to a tumor site). At the time of this writing, this is the least common form of CTF as it (1) offers a volume just large enough to perform the procedure and (2) provides the highest operator dose. Usually, such real-time guidance is not needed, as physicians can plan their next device push using existing static images.
- ***CTF quick check:*** A single axial/sequential image is acquired with the operator next to the patient. The images are immediately presented to the operator using in-room monitors, and the operator then uses those images to progress with the intervention (i.e., pushing a biopsy needle to a tumor site). This form of CTF is preferred as it offers the advantages of shorter exams times, like real-time CTF at a lower dose.

Pay attention to the fact that the three modes of CTF described here use three different modes of scanning acquisition. Therefore, be careful when discussing patient dose, operator dose, volume coverage, and exam time for CTF. Each of these quantities are different for each CTF type.

1.2.6 Gantry Tilting

X-rays at clinical CT beam energies (here we are not referring to micro, industrial, or megavoltage CT) are attenuated greatly by high-Z (i.e., high effective atomic number) and high-density materials, which can commonly be found in the body. Luckily, the largest pieces of metal in the body are usually found in the extremities, in the form of orthopedics, and not near the majority of the solid organs within the abdomen, the heart/lungs, or the brain. Metal implants from extremity orthopedic devices cause a lot of artifacts, as shown in chapter 16. Probably the most common form of metal in the body, however, is dental amalgam (e.g., dental fillings). Dental amalgam artifacts are so common that most CT manufacturers have engineered a hardware solution to deal with them. On most diagnostic CT scanners, there is an option to change the orientation of the scan plane with respect to the patient. Figure 1.9 depicts how MDCT scanners can tilt their scan planes. This tilting allows metal artifacts to be "thrown out" of the plane of important structures within the body. The most common use for gantry tilting is for throwing dental amalgam artifacts away from the base of the brain, as shown in Figure 1.9c. Gantry tilting is also commonly used in order to allow better physician access to the patient and optimal visualization of needle/device paths in CT fluorography, as shown in Figure 1.9d.

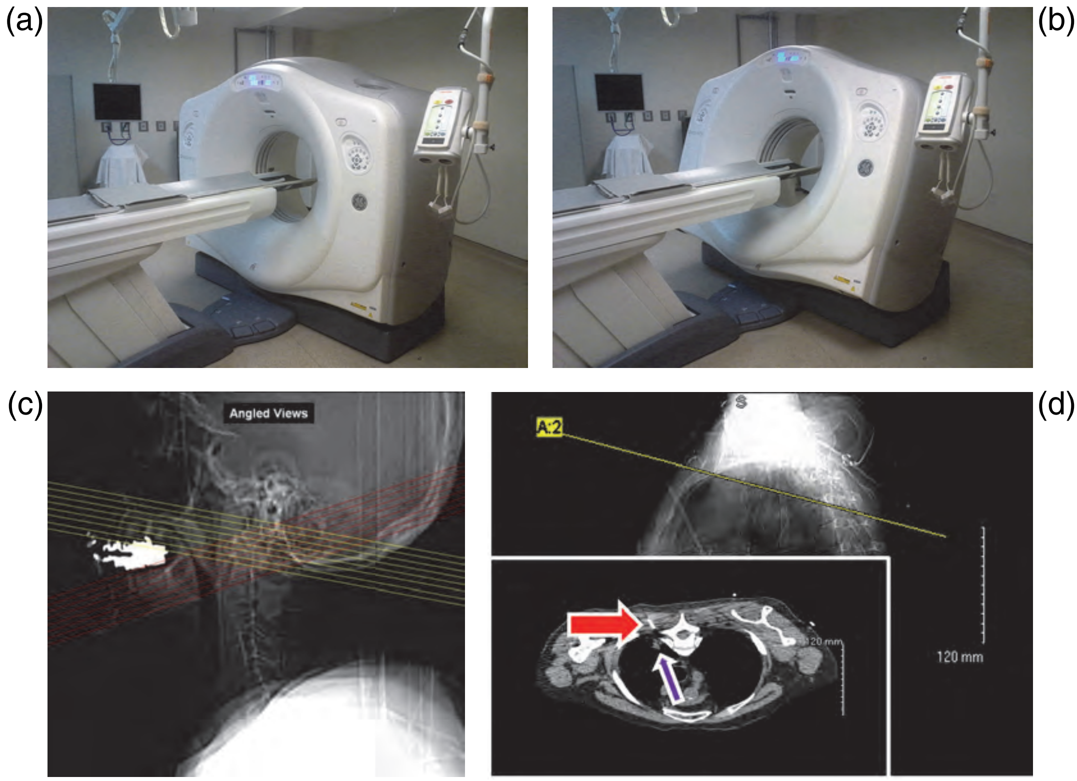


Figure 1.9 Most diagnostic CT scanners in use today feature gantry tilt. This feature is mainly used to move artifacts from dental fillings away from the brain and in interventional CT. (a) Depicts a scanner in the usual 0 degree gantry tilt position, while (b) is shown at a typical clinical tilt of 20 degrees. (c) Demonstrates how acquiring two sets of tilted images can avoid the artifacts from dental fillings. In interventional CT, where physicians must work between the patient and scanner gantry (a very small space), tilting of the gantry can facilitate easier access to the patient. (d) A lateral scout (yellow line depicting the plane on the axial image shown in the inset) and an axial image slice showing a biopsy needle (large red arrow) and a tumor (smaller purple arrow).

1.2.7 Scan Angular Range

CT enjoys a robust mathematical foundation. In other words, there exists a formula that allows one to take projection data of an object and use it to reconstruct the object [5]. This framework specifies a CT data acquisition must acquire 180 degrees plus the fan angle's worth of projection data. For most CBCT and MDCT systems, this comes out to be about 200 and 230 degrees worth of data, respectively. Figure 1.10 shows examples of full- and short-scan angular ranges.

For almost all routine imaging performed by CBCT and MDCT systems, angular scan ranges of about 200 and 360 degrees are used for image reconstruction. The minimum is used in CBCT imaging due to the slow movement of the c-arm gantry and patient access requirements. Since the scan time will scale linearly with the scan angular range, using the minimum needed angular range is an easy way to reduce scan times. It also reduces the mechanical design constraints on the c-arm gantry. For MDCT, helical scanning is the most common form of data acquisition. In helical mode, the x-ray tube is

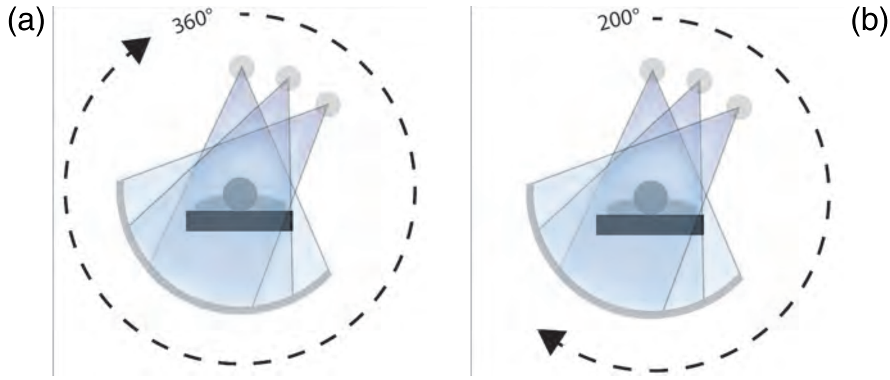


Figure 1.10 Depiction of (a) a full scan angular range and (b) a short scan angular range. Data collection in MDCT usually uses a full scan range, but may use only a short scan range's worth of data for image reconstruction to reduce motion artifacts. CBCT systems usually can only acquire data over a short scan range.

on continuously for tens of seconds as thousands of millimeters of the body (i.e., in the superior/inferior direction) are scanned. In MDCT, due to the common use of helical scanning and the need for uniform data sampling for each plane of the body, it is most common to use 360 degrees of data for image reconstruction. The end user typically does not get to choose the angular range used or data reconstruction on a clinical MDCT scanner. Scanner vendors will use different angular ranges depending on what mode the user selects the scanner to operate in (i.e., different pitches or gating modes), but the end user is rarely informed on how these modes change the angular scan range.

When high image quality is needed in CBCT, some scanners allow for special extended angular range scans to be acquired. In MDCT, when high *temporal resolution* (see section 3.5 for a definition of temporal resolution) is required, for example in coronary imaging, the minimum angular range of ≈230 degrees is used. In all flavors of CT, the use of 180 degrees plus the fan angle's worth of data for acquisition is referred to as the "short scan" angular range. The use of a short scan angular range by MDCT vendors is why the temporal resolution of some MDCT scanners will be cited as being shorter than the gantry revolution time. Since a complete gantry revolution of 360 degrees is not required for image reconstruction, the vendor can quote the ratio of ≈230/360 times the rotation time as the temporal resolution. In practice, there are other mathematical weighting schemes that are applied to reduce the amount of data used for short scans. These weighting schemes are beyond the scope of this text [10].

Up to this point in this section on scan angular range, we have not considered the use of two x-ray tube/detector arrays. Such systems are referred to as *dual-source* systems. These systems have the two x-ray sources imaging the same plane offset by about 90 degrees. Such a design does not change the data collection requirements for image reconstruction, but it does allow the data collection process to proceed faster. To cover the short scan range, a dual-source system, therefore, only needs to rotate the gantry for about 140 degrees (i.e., ≈230–90 degrees, where 230 degrees is the short scan range and 90 is the offset between the two tubes). Therefore, a dual-source system can acquire the needed data for a short scan in ≈140/230 the time needed for a single-source system. Caution should be applied to this comparison since vendor-to-vendor differences in reconstruction algorithm, system geometry, and the clinically needed field of view will all impact this comparison.

There are also experimental systems with designs for multiple x-ray sources along the z-axis (i.e., along the superior-inferior dimension) of the patient, but such systems are not commercially available for clinical use at the time of this writing [11,12]. There are baggage scanning devices with fixed numbers of stationary x-ray tube/detector arrays, but we will not consider such systems in this text. The scanning beam electron catheterization (i.e., cath) lab system, which has been shown to enable tomographic image reconstruction, can also be used to acquire tomosynthetic images of human anatomy from only a single view angle, allowing for temporal resolution on the scale of about 30 frames (i.e., tomosynthetic volumes) per second [13].

1.2.8 Detector Coverage

The 1990s was a period within the MDCT community known as "the slice wars" (see Figure 1.11). During this period, vendors in the MDCT community competed to achieve wider and wider beam collimation. The reason for this was twofold: (1) wider beam collimation meant more of the beam could be used to image the patient, meaning less radiation was wasted by the tube, and (2) since more of the patient could be imaged at a time, the total exam time was decreased, allowing for a plethora of new indications to be imaged. These new indications required faster scanning to view contrast dynamics (e.g., performing CT angiography of an entire organ group or vascular system within a single bolus of contrast). Tube heating was a concern because tubes have a finite amount of time they can run at the higher mA values. Increasing the beam collimation is analogous with increasing the maximum tube output. At a wider beam collimation, more x-rays from the tube are not blocked by collimation, so more x-rays reach the patient and can be used in image formation. Therefore, at wider beam collimations, x-ray tubes can deliver more dose to the

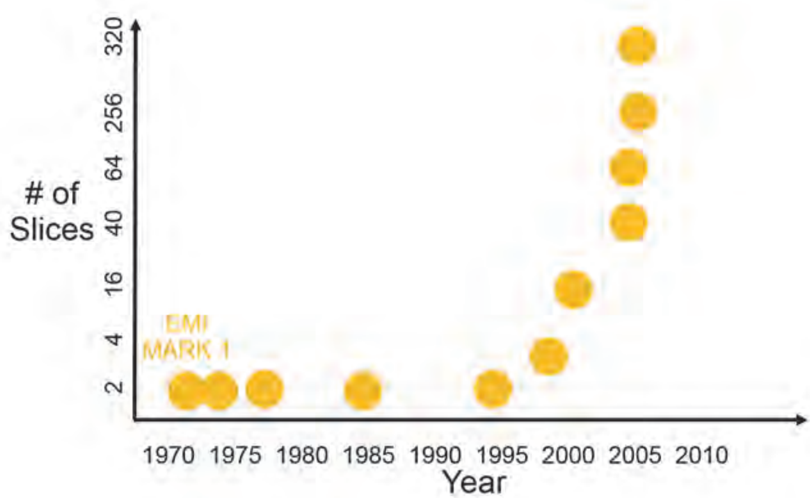


Figure 1.11 Evolution in the number of slices. (Adapted from Professor Willi Kalender [6].)

patient at a given mA value. For a fixed image quality need at a narrow versus wide collimation, this translates into faster scanning at wider collimations.

1.2.9 CBCT

For CBCT systems, early detectors were sized according to the needs of the interventionalist. Since interventional procedures can be performed with a relatively narrow field of view, early CBCT systems employed detector sizes that could not image the entire cross section of a patient's body. In other words, a cardiologist operating on the heart only needs to see the heart, a few centimeters surrounding the heart, and the vessels they use to access it. In practice, depending on the intervention, often only a subregion of the heart is needed to be visualized. For example, during an interventional electrophysiology procedure, one only needs to visualize the catheter ends. The cardiologist does not need to see the entire mediastinum, lung fields, lateral edges of the chest cavity, etc. For CBCT, however, when a detector is not large enough to cover the patient's entire cross section, truncation artifacts are created [14,15].

Typical collimated beam sizes in CBCT for body imaging on a high-end system are about 40×30 cm, and for dedicated head systems about 20×20 cm. There are also in-between collimations available from different vendors, like 30×30 cm. The collimation size is totally dependent on the size of the detector, which is one of the more expensive pieces of a CBCT system. With CBCT systems using flat panel technology today, smaller detectors usually allow for higher frame rates. All sizes of commercially available CBCT systems usually have approximately the same resolution (e.g., detector element sizes of

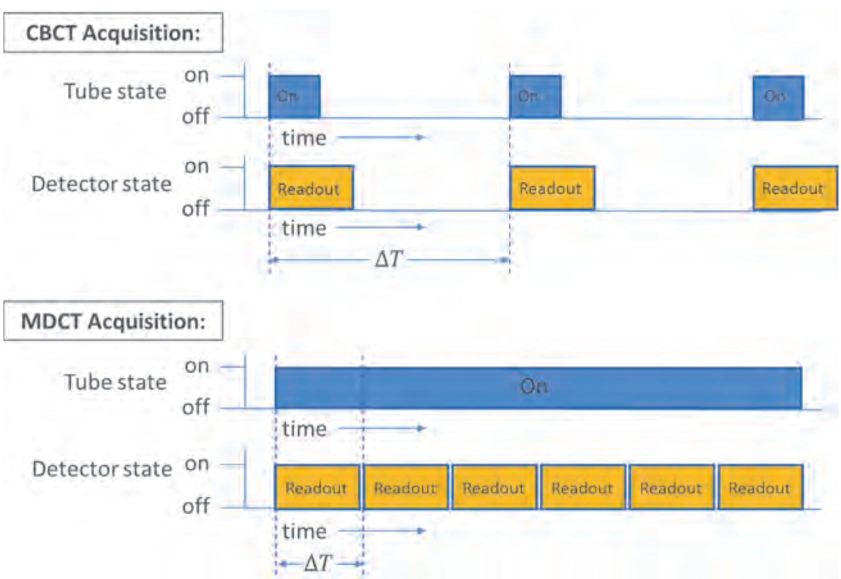


Figure 1.12 A comparison of "x-ray" on and detector readout times for MDCT compared to CBCT. Note that the ΔT sampling time period for a CBCT may include times where the tube is neither on nor data is being collected. In general, the total data collection times for CBCT are on the order of tens of seconds relative to MDCT, which is usually sub-second for a complete short scan (≈220 degree c-arm sweep) or 360 degrees of data collection, respectively. (Figure provided by Dr. Adam Budde, GE Healthcare.)

0.15 to 0.3 mm). Different from some MDCT scanners, the detector elements are actually the same size over the entire detector in CBCT.

Also, be aware that on some systems, the minimum detector size on a CBCT detector can not be used at the maximum field of view. This is similar to some MDCT systems, which force one to use wider slice widths when the full beam collimation is used. In CBCT, the use of a lower resolution at the full detector collimation is referred to as *binning*. Binning refers to the reading out of neighboring detector elements together. In other words, the signal from adjacent detector elements is added, making the element's size appear larger and reducing the amount of acquired projection data. Binning reduces the amount of data from the detector and is usually the reason vendors employ binning at large collimations in CBCT. An additional difference relative to MDCT is how the x-ray beam is pulsed during data acquisition, as explained in Figure 1.12.

1.3 Flavors of CT

This section will review all of the implementations of CT technology present in health care and beyond. Usually, only one flavor of CT is used in the same environment. There are procedures, however, that benefit from the interventional flexibility of a CBCT system and the high image quality of MDCT. Such procedures can be done in special rooms fitted with both flavors of CT, as shown in Figure 1.13.

Most MDCT scanners had beam collimations equal to or less than 4 cm. CBCT uses beam collimations of 10 to 40 cm [16]. Therefore, separating MDCT from CBCT systems was easily done using beam collimation. However, the delineation between cone-beam and convention CT is becoming ill defined and more difficult to answer as premium MDCT scanners employ detectors equal to or larger than those in some CBCT

Figure 1.13 It is not uncommon to find different flavors of CT used in the same environment. This image shows the combination of an interventional CBCT system coupled with an "MDCT on rails." Notice the metal strips on the floor in front of the MDCT scanner. Those strips are guides allowing the MDCT scanner to move over the patient, enabling volumetric image acquisition where the patient remains stationary, and the scanner translates over them. (Image provided by Siemens Healthcare.)

systems. The detector technologies differ between MDCT (i.e., solid state indirect conversion) and CBCT (i.e., flat panels). The detector size in the fan angle is always greater on MDCT compared to CBCT, but for the cone angle, some MDCT systems are equal to or larger than some CBCT systems. The data acquisition and reconstruction is also quite similar. For both MDCT and CBCT, an x-ray source coupled with an x-ray detector rotate about a patient and acquire projection data. The projection data is treated almost identically between these two modalities, each differing somewhat due to the geometrical and scan range differences. Since the introduction of flat-panel-based CBCT, however, modern CT scanners have been developed that have beam collimations larger than those in some CBCT systems. For example, Toshiba/Canon and GE make CT scanners with z-axis collimations of 16 cm; this coverage is larger than that found on all dental CBCT systems. Therefore, using the size of the z-axis collimation to delineate MDCT from CBCT is not a reliable distinguishing detail. A better metric would the assembly that the x-ray tube and detector reside on. If this assembly forms a complete ring around the patient, the system would be referred to as MDCT. If the assembly is a c-arm, then the system is a CBCT system. While it might also be possible to define MDCT versus CBCT using detector type, this may not be as reliable and is not as easy to ascertain via visual inspection. The multiple flavors of CT scanners, and some of their basic geometrical and scanning acquisition details, are outlined below.

1.3.1 Diagnostic Radiology

Scanner Name or Acronym

MDCT is the accepted name for these CT scanners.

General Comments

The most common type of CT in use today. Performs the most clinical exams relative to the other CT modalities listed here. Baggage scanners likely scan the most objects given some of them run continuously. Figures 1.14 and 1.15 show examples of the appearance of the mechanisms inside a modern MDCT. The scanners used for radiation therapy treatment planning (i.e., CT simulation) and interventional CT usually have the same hardware as the MDCT scanners. Figure 1.16 shows some common equipment that can be found inside an MDCT imaging suite.

Typical Bore Size

50 to 80 cm, with 70 cm being the most common. A bore size of about 80 cm is considered a "wide bore" and is used on dedicated bariatric, radiation planning, or interventional CT applications within a department. At the present, several CT vendors offer high-end CT scanner models that come standard with wide ≈80-cm bore openings.

Typical Detector Coverage

2 cm (8–16 slices) to 16 cm (256–320 slices) with 4 cm (64 slices) being the most common. Some vendors that use a deflected focal spot in the z-direction refer to their 64-slice systems as having 128 slices.

Typical Rotation Time or Data Collection Speed

MDCT scanners have rotation times in the range of 0.2 to 2 seconds. Scan times vary on body part and required dose level. Scan times usually range from sub-second (i.e., gated heart or high-pitch chest and pediatric exams) to tens of seconds (i.e., multi-body-part angiography or perfusion exams).

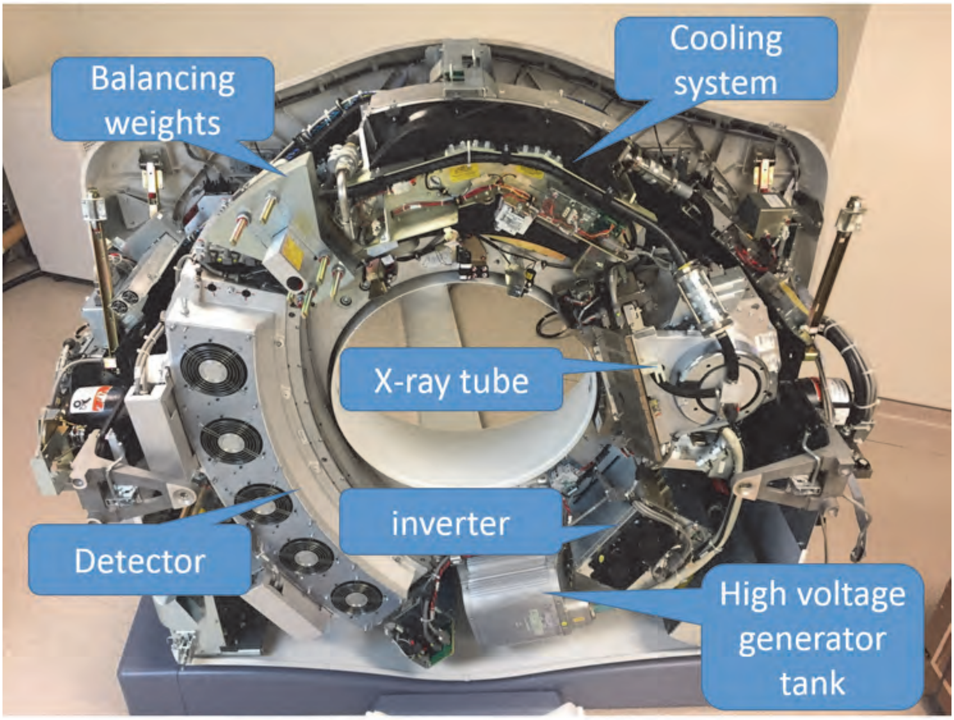


Figure 1.14 A modern MDCT scanner without its cover is shown. Different from many c-arm-based CBCT units, the high-voltage generator and cooling system resides on the gantry.

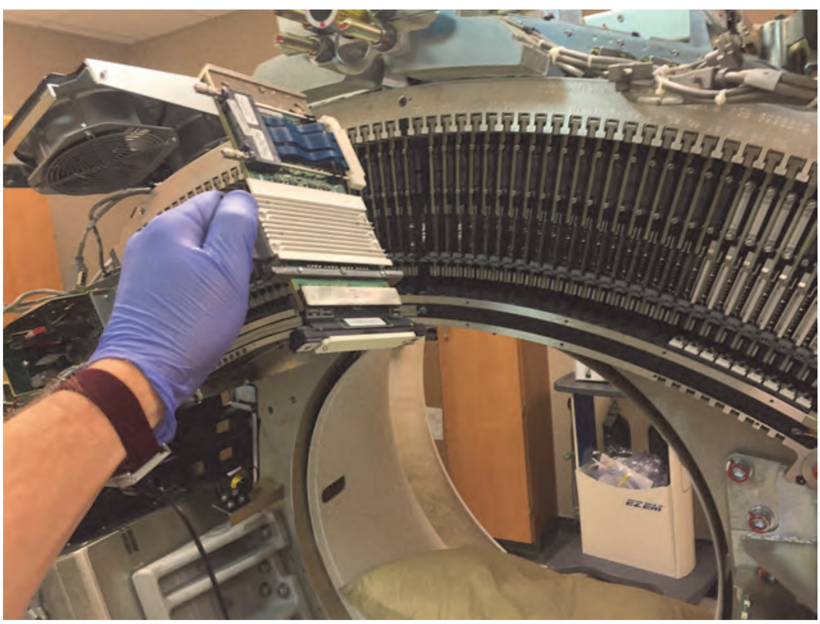


Figure 1.15 A detector module being replaced on a 64/128 (i.e., ≈4 cm coverage) channel CT scanner. This detector module is composed of multiple parts (scintillator, data acquisition system, analog-to-digital converter, heater) which a field engineer usually does not take apart. This particular module is in the center of the detector array and is being replaced because a ring artifact was observed that could not be calibrated away.

(a) (b)

Figure 1.16 (a) Any CT exam requiring gating based on the ECG signal must have an ECG monitor system for the CT scanner. Usually, special pads and connectors are needed for cardiac-gated CT scanning. In other words, the leads common to routine monitoring of a patient are not compatible with the input needed for a CT scanner. (b) An insufflator device used to administer air or CO_2-based rectal contrast agent common for colonography (i.e., virtual colonoscopy) exams.

1.3.2 Interventional Radiology CT Fluorography

Scanner Name or Acronym

MDCT (sometimes called a "wide-bore" scanner) is how these scanners are referred to in the literature.

General Comments

An MDCT system used for interventional procedures—such as ablations, arthograms, or biopsies—is not referred to differently than a diagnostic MDTC system. Typically, the most distinguishing feature of an interventional MDCT scanner is the option to perform CT fluorography, which is usually not a standard option on a diagnostic MDCT scanner. These scanners also come with special software options to facilitate the planning, acquisition, and viewing of CT fluorography data. These scanners may also include options such as needle guidance, which can predict the future location of a needle or ablation tip. However, it is common for some sites not to actually use the CT fluoroscopic modes of these scanners for interventional procedures. In such cases, staff are usually worried about physician exposure and choose to utilize the interventional planning and viewing software, but retreat to the control room while CT images are acquired.

Typical Bore Size

≈80 cm

Typical Detector Coverage

This is usually smaller than a diagnostic MDCT scanner since a large detector size is not needed for CT fluorography in most applications. In other words, it would not make sense to buy both a wide-detector MDCT scanner and the CT fluoroscopy options since a large portion of an interventional CT scanner's schedule would not be devoted to routine diagnostic scanning, where a wide detector would be most beneficial. Albeit, now vendors are providing both wide detector arrays and large-bore openings on their high-end systems.

Typical Rotation Time or Data Collection Speed

Same as diagnostic MDCT

1.3.3 Baggage Scanning CT

Scanner Name or Acronym

An accepted or widely used acronym for these scanners does not exist.

General Comments

In general, the specifications for these scanners are classified. In the United States, the federal government is the main buyer for these scanners. These scanners get used in airports, rail stations, shipping centers, and other places where there is a high risk of malicious intent. They often come with advanced segmentation algorithms and use dual-energy imaging methods to label potential threats. Given the volume of baggage these scanners see, some form of automated detection is required or the burden on the humans running the scanners would be too great.

Typical Bore Size

Models of screening scanners are available that range from tabletop screening scanners for carry-on luggage/suite cases to scanners used for shipping cargo. For example, a carry-on baggage scanner may allow scanning of objects roughly 60×50 cm in cross section. The scanners that the public does not see—those used for checked bags—typically allow objects with cross sections up to 100×60 cm. Note: there are multi-view systems with larger openings, but they do not allow for volumetric CT imaging (i.e., two-view x-ray systems exist that can image objects with cross sections in the meter range).

Other Technical Details

Northeastern University runs the Alertness and Localization of Explosives-Related Threats (ALERT)† program in cooperation with the U.S. Department of Homeland Security. They publish workshop proceedings that provide a good source of information on the ever-changing face of threat detection using CT scanners and other methods.

1.3.4 Interventional Radiology C-arm (CBCT)

Scanner Name or Acronym

CBCT is the accepted acronym for these cone-beam CT units.

General Comments

CBCT is most commonly performed on systems primarily used for fluoroscopic guidance or digital subtraction angiography. On “bi-plane” systems that use two c-arms which can be positioned independently of each other during procedures, only one of the c-arms would be used for CBCT imaging. Historically, interventional radiology used image intensifier-based c-arm systems. After the switch to flat-panel technology, CT was able to be realized on these systems. It is not uncommon to never use the 3D (i.e., CT) imaging mode during many procedures performed using CBCT-capable interventional systems. See chapter 14 for more information on CBCT systems.

†http://www.northeastern.edu/alert/

Typical Bore Size

Since the CBCT scanner design is inherently more open because of the “c” geometry and the detector-to-isocenter or source-to-isocenter distances are adjustable (see Figure 14.1), the concept of bore size is not well defined for this scanner type. The actual patient size that can be “fit” inside the “c” of the scanner is much larger than that of the “donut opening” found on MDCT scanners. Relative to MDCT scanners, however, these systems usually suffer from truncation artifacts when imaging anything but the head in adult patients. In other words, larger patients/objects can be imaged on CBCT relative to MDCT, but smaller reconstruction fields of view can be realized relative to MDCT.

Typical Detector Coverage

20–40 cm image volume is possible, determined by the size of the flat panel detector used.

Typical Rotation Time or Data Collection Speed

CBCT data acquisitions are commonly referred to as “sweeps,” as opposed to “rotations” in MDCT. CBCT sweeps are on the order of 10 seconds for a single scan. Scans are referred to as “sweeps” due to the sweeping motion of the c-arm assembly moving around the patient.

1.3.5 Dedicated Head Scanner

Scanner Name or Acronym

An accepted or widely used acronym for these scanners does not exist. An example of a dedicated head scanner is the Ceretom® (Samsung-Neurologica, Danvers, MA, USA).

General Comments

Dedicated head scanners are usually smaller, less powerful (i.e., lower maximum mA limits) scanners compared to MDCT. Dedicated head scanners still employ a third-generation CT gantry design. Many are mobile to allow the scanner to be brought to the patient in the setting of an ICU or dedicated neurology department. Such scanners have also been placed inside ambulances for acute stroke work-up on the way to a treatment center.

Typical Bore Size

Being intended for imaging of just the head, these scanners employ bore sizes suited for fitting adult heads, typically ≈32 cm. The Ceretom allows a reconstructible FOV of 25 cm.

Typical Detector Coverage

These units use ≈1 cm detector coverages (Ceretom uses 8×1.25 mm slices) in the z-direction and can support non-truncated head reconstruction. A full acute stroke work-up, including ruling out aortic injury, is not possible with these units as they cannot image down into the chest. These units do not have the field of view to support non-truncated imaging, a bore size to allow the scanner to go inferior to the patient’s head, or a scan range long enough to cover from the top of the head to the mid chest.

Typical Rotation Time or Data Collection Speed

Scan times are ≈1 to 2 seconds per rotation. On mobile units, it is harder for vendors to enable faster rotation times given fast rotation times require a well-balanced gantry, which is difficult to ensure due to the fact that the unit itself (i.e., gantry housing) moves.

1.3.6 Mobile CT

Scanner Name or Acronym

An accepted or widely used acronym for these scanners does not exist. At the time of this writing, an example of a mobile CT scanner intended for diagnostic imaging is the BodyTom® (Samsung-Neurologica, Danvers, MA, USA). An example of a surgical planning mobile CT scanner is the O-Arm (Medtronic, Dublin, Ireland).

General Comments

These units are battery powered and can perform many scans without being plugged in to recharge. Specially sized doorways and rooms (i.e., layout and size) are needed for efficient operation of these units. In general, they lack many of the special features available on MDCT scanners and are best suited for routine head and torso imaging or for surgical device localization. These two uses, diagnostic and surgical planning, represent two different types of mobile CT scanners that should not be confused with each other. Dedicated surgery mobile scanners do not offer image quality that is good enough to allow for diagnostic scanning. In general, the image quality of even those mobile CT scanners intended for diagnostic imaging is inferior to a modern MDCT scanner.

Typical Bore Size

≈70–100 cm. These scanners' bore sizes are usually larger than a department's MDCT scanner since these units are primarily used for interventions in an operating room. While these systems have large bore openings, the surgical planning units usually support only a small reconstruction field of view, which causes the resulting images to suffer from truncation artifacts.

Typical Detector Coverage

2–4 cm, usually with fewer channels than MDCT systems

Typical Rotation Time or Data Collection Speed

Scan times are usually greater than or equal to 1 sec. On mobile units, it is harder for vendors to enable faster rotation times given fast rotation times require a well-balanced gantry, which is difficult to ensure due to the fact the unit itself (i.e., gantry housing) moves.

1.3.7 Dental CT

Scanner Name or Acronym

An accepted or widely used acronym for these scanners does not exist. Within the dental community, dental CT is referred to as CBCT, 3-D Radiography, or cone-beam radiography. An example dental CT system is the NewTom VGI Evo (Via Silvestrini, Verona, Italy). The market space at the time of this writing has over a dozen vendors offering CBCT-based dental CT units.

General Comments

These scanners represent a major investment for a dental office. These units typically cost around $100,000. A "general" dentist practice would usually not have the need for one of these units as 2D imaging methods suffice for routine dental work. The scanning geometry of dental CT is notably different from other CBCT units used in the health care setting. In dental CT, the patient usually sits down or stands, and the x-ray source and detector are rotated around the patient's head. Interestingly, while most dental CT scanners use flat panel detector arrays, some employ the use of image intensifiers.

Typical Bore Size

Dental CT scanners usually employ an "open geometry" like an interventional c-arm or linac-mounted CBCT system, in which the x-ray source and detector can be seen moving around (i.e., sweeping) the patient. These scanners vary in the view or angular range needed for image reconstruction: on the small end they sweep out a short scan range of 200 degrees, but some complete a full 360-degree sweep. Source-to-object and source-to-detector distances vary considerably across the many dental CT offerings on the market today. It is safe to say all dental CT scanners will allow for adult human access for the modes and geometries advertised by the vendors. In other words, imaging may not be possible for the neck or shoulders on a dental scanner. The angular sweep and size of the source and detector housing may collide with the patient when the scanner is used for imaging a body region it was not designed for.

Typical Detector Coverage

Detector coverage for dental CT usually will not allow a patient's entire head/neck to be imaged without truncation. The maximum field of view on these units varies by model, with the largest having an FOV of 24×30 cm. Most units will have an FOV of 10×10 cm (ranging from 4×4 cm to 15×15 cm with most units, allowing a variable FOV based on resolution needs). The geometrical magnification, focal spot sizes, and detector element sizes allow for spatial resolutions in the range of 80 to 300 microns, much smaller than MDCT and most other clinical CT flavors.

Typical Rotation Time or Data Collection Speed

Typical data collection times are on the order of 5 to 30 seconds. Image reconstruction, however, can range from nearly real-time to 5 to 10 minutes when large-volume, high-resolution images are required.

1.3.8 CT Simulation for Radiation Therapy

Scanner Name or Acronym

These scanners are commonly called "simulators" or "CT sim," reflecting their use to simulate the position of a patient receiving a radiation therapy treatment. See section 14.3 for more information.

General Comments

The imaging needs of a radiation therapy clinic are mainly (1) obtaining a three dimensional volume of the patient's mass density in the exact position the patient will be in during treatment and (2) obtaining images of enough quality to define the tumor/target volume. The former requires special "flat tabletops" that duplicate the flat tables/couches used on linear accelerators. The latter requires images reconstructed using techniques good enough to define mainly soft tissue lesions. Prior image data sets from other imaging modalities are usually present to aid the dosimetrist and radiation oncologist in defining the treatment target. It is not uncommon for a radiation oncology clinic to use CT contrast agents to allow better target visualization. Dedicated radiation oncology scanners will also come with 4DCT (four dimensional CT) imaging modes so respiratory motion can be quantified and used to either gate or define the size of the treatment volume for tumors in the lungs.

Typical Bore Size

These units are usually the same as a vendor would offer for diagnostic radiology and will, therefore, have the same bore size as MDCT scanners. In the radiation oncology environment, however, patients usually are positioned not to optimize image quality, but to mimic their position during radiation treatment. These positions commonly force patients to have anatomy or positioning/immobilization devices lie outside of the traditional ≈70-cm bore opening. For this reason, most vendors that sell diagnostic MDCT scanners re-brand their wide-bore MDCT scanners as radiation oncology scanners. These scanners will have the vendor's largest bore sizes, usually used for interventional MDCT, and come with special software options, like metal artifact mitigation, wide field of view reconstruction algorithms, and extended CT number reconstruction. They also will have options for in-room hardware, like additional laser packages and respiratory gating systems.

Typical Detector Coverage

2 cm (8–16 slices) to 16 cm (256–320 slices) with 4 cm (64 slices) being the most common. Some vendors that use a deflected focal spot in the z-direction refer to their 64-slice systems as having 128 slices (see section 17.1.4).

Typical Rotation Time or Data Collection Speed

MDCT scanners have rotation times in the range of 0.2 seconds to 2 seconds. Scan times vary on body part and required dose level, but usually range from sub second (i.e., gated heart or high-pitch chest and pediatric exams) to tens of seconds (i.e., multi-body-part angiography or perfusion exams).

1.3.9 Image-Guided Radiation Therapy

Scanner Name or Acronym

The imaging systems on these scanners are usually referred to as CBCT or as "OBI" (onboard imager). The flat panel CBCT-based system ubiquitous on modern linacs (linear accelerators) is one of the most common methods to realize image-guided radiation therapy (IGRT). Other imaging methods—such as MRI, ultrasound, and planar megavoltage imaging—may also be used to realize IGRT. X-ray-based vendor examples of this technology include the Varian OBI, the Elekta XVI, and the Siemens in-line kV-CBCT.

Some radiation therapy treatment systems employ the use of "CT on rails." These systems consist of a marriage between a radiation therapy machine (i.e., a linac) and an MDCT scanner. They are connected rigidly by a track in the floor over which the MDCT unit can be moved to the patient for imaging. These rail designs are also used in combination c-arm interventional suites with MDCT scanners. The specifications of MDCT scanners coupled to radiation therapy units in this way are not considered in this section, as they are covered in the MDCT section.

General Comments

These systems have a geometry similar to and use similar hardware (i.e., detectors and x-ray tubes) as CBCT systems used in the interventional setting. However, their gantry is more rigid because the CBCT imaging system is located on the linac. Linac gantry movement is accurate to sub-millimeter precision in order to assure minimal therapy dose delivery error. CBCT is usually just one of several modes these OBI systems offer to the radiation therapy environment. In addition to CBCT, these systems can also acquire 2D radiographic images and fluoroscopic image sequences to aide in patient positioning and target localization. Patient and target position is usually confirmed by correlating, or registering, the images obtained from the OBI to the planning CT acquired using the clinic's CT simulator.

Typical Bore Size

The concept of bore size is ill-defined for the c-arm geometry of CBCT and image-guided radiation therapy implemented via CBCT. While there may not be a bore, an x-ray source and detector still rotate about the patient with source-to-detector distances of ≈150 cm. The source-to-isocenter (i.e., source-to-axis) distance on these units is always 100 cm. The Siemens in-line kV CBCT system uses a movable source and detector that lie along the same direction as the treatment beam, but reversed 180 degrees. In other words, the Siemens in-line system moves a flat panel detector to lie in front of the linac source and an x-ray source to the position where an MV imaging detector would normally reside. This inverted geometry mitigates the need to limit patient access via an x-ray source and detector rotated 90 degrees to the treatment head.

Typical Detector Coverage

The largest flat panels—large flat sheets of hydrogenated amorphous silicon-based (a-Si:H) thin film transistors (TFTs) with a scintillating material of thallium doped cesium Iodide—are about 40×40 cm. The coverage in the z-direction (superior-inferior) at isocenter is ≈27 cm. In the axial plane (i.e., x-y plane), the coverage with the ≈40-cm detector centered on the x-ray source is only ≈27 cm. This is too small to avoid truncation artifacts in most patients and body regions. Linac vendors allow for a so-called "offset" to be applied to the detector which lets one increase the detector coverage and expand the reconstructible field of view to ≈50 cm. The image quality obtained from these systems is far below that of MDCT, but sufficient for the patient registration tasks needed to realize an IGRT work flow.

Typical Rotation Time or Data Collection Speed

These linac-based CBCT systems rotate even slower than CBCT systems found on c-arm systems in interventional suites. They typically rotate around 3 degrees/sec. They can acquire a full 360 degrees of data, which provides better image quality and is needed when a detector offset is applied. They can also acquire data for a short scan (i.e., "half scan") range of data. In both cases, data acquisition times are on the order of minutes.

1.3.10 TomoTherapy® and MVision®

Scanner Name or Acronym

TomoTherapy® is a radiation therapy treatment system made by Accuray (Sunnyvale, CA). It is the only commercial solution using megavoltage x-rays to produce clinical CT images in a helical fan-beam geometry. MVision® is a radiation therapy imaging system realizing CT using a CBCT geometry at megavoltage energy levels. See section 14.3 for more information.

These two units have been separated from the other IGRT and industrial CT flavors as they are the only clinical CT scanners currently using a megavoltage energy spectrum.

General Comments

Each of these systems uses the same therapy x-ray production hardware to produce the imaging spectrum. They both also, therefore, use the same geometry as the treatment beam (i.e., there is no offset from these imaging chains' view angles from the actual treatment beam, as in kV CBCT systems). Since these units use megavoltage beam energies, the image contrast is dominated by the Compton Effect, with essentially no contribution from the photoelectric effect. This means images obtained from such megavoltage systems are proportional to the electron density of the patient.

Typical Bore Size

For the TomoTherapy H series, the bore size is 85 cm, larger than even wide-bore MDCT units. For the MVision system, the source-to-detector distance is 145 cm and the source-to-isocenter distance is 100 cm.

Typical Detector Coverage

For the TomoTherapy H series, a xenon ion chamber array is used which supports a reconstructible field of view of 39 cm. This relatively small FOV means most patients will suffer from truncation, but enough anatomy can be collected for patient registration.

On the MVision system, the detector size is 41×41 cm, limiting the reconstructible field of view (radius over which there will be no truncation artifacts) to 27.4 cm. Most all anatomical regions and patients will be truncated with this FOV, but it is large enough to allow for meaningful patient registration to be made near the treatment volume.

Typical Rotation Time or Data Collection Speed

For the TomoTherapy H series, a mid-range level of image quality takes ≈2 minutes per 15 cm of z-axis coverage for data acquisition. On TomoTherapy systems, image quality, scan time, and dose are all functions of the table feed per rotation, which can be set to coarse, normal, or fine, corresponding to feed rates of 4, 8, and 12 mm per rotation, respectively. The rotation time is 10 seconds, which is very slow relative to other clinical CT systems. For example, a 100 cm coverage would take 830 seconds using coarse mode, which has a scan speed of 1.2 mm/s.

For the MVision system, the scanner acquires data over a short scan range of 200 degrees. For an imaging dose of 10 monitoring units (MU), the scan time reported by the vendor is 45 seconds.

1.3.11 Dedicated Breast CT

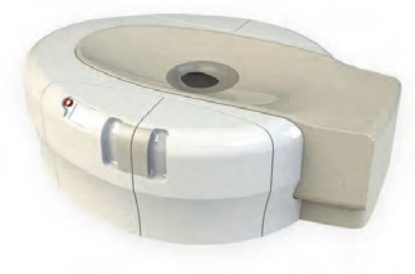
Scanner Name or Acronym

These units are not currently widely used, but several companies are producing units in this space. The Koning Corporation manufacturers a dedicated breast cone-beam imaging system that, as of 2016, has received U.S. FDA, CE Mark, China CFDA, Health Canada, Australia TGA, and New Zealand approvals [17]. At the time of this writing, there is another company, Advanced Breast CT GmbH, which was CE labeled on 9/24/2018, that uses a CdTe detector and a spiral/helical scan mode, allowing for scan times of 6–12 seconds at 80 micron resolution.

General Comments

These systems image the breast using a pendant geometry that does not require compression. The CT geometry is similar to CBCT, except the imaging plane is tipped over such that the breast is imaged along coronal planes. In other words, the patient lies down in a prone position such that her breast tissue falls into the scanner. The scanner's axial plane is then aligned with the patient's coronal plane, allowing cross-sectional images of the breast to be acquired as shown in Figure 1.17.

Figure 1.17 Breast cone-beam CT imaging system. This system has its imaging plane aligned to Earth's level plane. Patients lie on the scanner, and their breast tissue falls into the scanner to be imaged without compression. The image shown here is of a Koning Corporation system.



Typical Bore Size

It has been reported the bore opening on the Koning system is 39 cm with no insert in place [17]. Inserts are used of various sizes to aid in patient positioning and comfort.

Typical Detector Coverage

It has been reported that the reconstruction field of view is 16×28×28 cm on the Koning system [17].

Typical Rotation Time or Data Collection Speed

It has been reported the scan time is 10 seconds per breast on the Koning system [17].

1.3.12 Synchrotron CT

Scanner Name or Acronym

Synchrotron-based CT [18,19], also known as x-ray microtomography, is primarily a research and development tool. The name x-ray microtomography is also used to refer to nonsynchrotron-based CT modalities [20].

General Comments

Like all flavors of CT, these systems produce CT images with values equal to the linear attenuation coefficient (i.e., HU). Their x-ray source and imaging geometry, however, are quite different from other flavors of CT. The x-ray source is not a traditional x-ray tube; instead, a synchrotron produces the x-rays. In contrast to most other flavors of CT where the source and detector assembly rotate about the sample, the use of a synchrotron means the sample must rotate. To create synchrotron radiation, charged particles are forced to travel within in a large "storage" ring. As the charged particles move in a circle, they give off electromagnetic energy in the form of x-ray radiation. This radiation is focused onto a sample to be imaged. Detectors measure the amount of radiation that passes through the sample, like in all flavors of CT. The samples are rotated in order to acquire projection angles from different view angles. Very small resolution scales can be imaged at a synchrotron facility because the effective focal spot of these systems can be made much smaller relative to using a traditional x-ray tube. The beam energy from these systems may also be made to be monochromatic, in contrast to traditional x-ray-based systems that produce polychromatic spectra. These systems are usually characterized by the "brightness" of the x-ray source. A high-brightness x-ray source is one that can achieve shorter imaging times than a lower-brightness source. The brightness is analogous to effective mAs in CT flavors using x-ray tubes.

Typical Bore Size

The bore or sample size is usually quite small, with ≈5–10 mm being typical. However, it is common for each facility offering synchrotron-based CT to have a unique bore size.

Typical Detector Coverage

It is common for each facility offering synchrotron-based CT to have a unique detector configuration.

Typical Rotation Time or Data Collection Speed

Data collection for these systems depends on the facility and the desired level of image quality. It varies from minutes to hours.

1.3.13 Small Animal Imaging

Scanner Name or Acronym

MicroCT (or μCT) is used to refer to these units in the literature. An example is the Inveon® (Siemens Medical Solution, Forchiem, Germany).

General Comments

These scanners are used mainly on small animals and in industrial research. They are commonly used for *in vivo* imaging of animals undergoing therapy research, where scans over the course of an intervention are acquired over time. Some units rotate the sample, keeping the x-ray source and the detector position fixed, while others use the traditional third-generation geometry common to MDCT, where the x-ray source and detector rotate about the sample. They use flat panel, usually CCD-based detectors, which allow in a subset of modes (i.e., there are many modes in which the field of view, focal spot size, tube output, source-to-detector distance, etc. can all be changed). A spatial resolution of ≈10 mm can be obtained. These units are typically quite small and can fit within any normal lab space. They are self-shielded, so their placement can be made without concern for radiation shielding. In other words, the room they are placed in does not need to have lead shielding. The scanner unit itself closes around the imaging object and blocks x-rays to a level sufficient to mitigate the need for room shielding.

Typical Bore Size

The maximum sample size will vary by scanner model and whether the scanner rotates the sample or uses a third-generation geometry. Maximum sample sizes are in the range of about 10×10 cm. The maximum reconstructed field of view, however, is usually smaller than this size. At maximum spatial resolution (e.g., ≈10 μm), the reconstructed field of view may be only ≈1.5 cm. At lower spatial resolution levels (e.g., ≈50 μm), the reconstructed field of view may be ≈5 cm.

Typical Detector Coverage

This varies greatly, as described above, due to changes in spatial resolution. These changes are made by changing many system parameters, one of the most important being the geometric magnification. The geometric magnification is changed by increasing/decreasing the x-ray source-to-detector distance and the location of the sample with respect to the x-ray source and detector. For example, while a typical x-ray source-to-detector distance may be 20 cm, in a high-resolution imaging mode the sample may only have a source-to-object distance of 7 cm. For a full field of view mode (i.e., a lower spatial resolution mode) the sample may have a source-to-object distance of 18 cm.

Typical Rotation Time or Data Collection Speed

These times are very long relative to clinical CT imaging modalities. Scan times will vary by vendor and scan mode, with high-resolution, high-image-quality modes taking the longest. Scan times are usually on the order of minutes, with times over 10 minutes not uncommon for high-quality imaging (larger mAs values and full detector readout, i.e., no binning of detector data). Reconstruction times are also quite long relative to clinical CT systems, taking on the order of minutes to tens of minutes depending on the volume and voxel sizes.

1.3.14 Industrial CT

Scanner Name or Acronym

An accepted or widely used acronym for these scanners does not exist. Examples of this CT flavor vary widely, from µCT-like benchtop systems, to MDCT scanners placed inside lead-walled containers for use on factory floors, to industrial CT scanners such as the Nikon XT H 450 capable of imaging using a 450 kV spectra (Nikon Metrology Inc., Leuven, Belgium).

General Comments

Industrial CT spans the needs of industrial practices. On the small end of the spectrum, industrial CT is used to image small (i.e., a few cm in total object size) electronics/ceramics/plastics/etc. using micro-focus x-ray tubes with spatial resolution like that of µCT. On the large end of the spectrum, industrial CT images large metal machined parts requiring penetration of centimeters of high-density metal, while simultaneously demanding high sub-mm spatial resolution. Throughput in industrial CT varies greatly, from hours or days in a research and development setting for data acquisition and reconstruction to seconds or minutes for routine quality assurance and integration into manufacturing processes.

Typical Bore Size

Bore size will vary greatly, depending on the unit. On the small end, the bore size is equal to that of a µCT unit (i.e., ≈10×10 cm) to those seen in baggage scanning (≈100×100 cm).

Typical Detector Coverage

The detector type, size, and element size will all vary greatly depending on the model. There are even companies making industrial CT systems that use “CR” (computed radiography) film to acquire CT data sets that require manually processing for each view angle.

Typical Rotation Time or Data Collection Speed

The rotation and data collection speeds are, in general, slower than clinical CT flavors, owing to the fact that most industrial applications do not require high throughputs. Industrial CT units also often require high tube outputs to penetrate thick, highly attenuating objects (i.e., metal machined or cast parts) or to obtain adequate signal-to-noise ratios to support extremely high spatial resolutions.

1.4 The Most Common Misconceptions in Medical CT Today

CT technology and the number of applications using CT have been enjoying a rapid rate of growth. This has resulted in a wide range of different CT flavors present in health care and beyond (see Figure 1.18). Between specific flavors of CT and within a CT flavor, many differences exist. Therefore, it is not surprising that many misconceptions on fundamental aspects of CT exist. This short list is meant to illuminate some of the major mis-

Figure 1.18 Examples of CT scanning "flavors" from a variety of practice settings. All of these scanners use x-ray projection data acquired from a plurality of angles about an object to reconstruct tomographic images.

conceptions CT faces in the setting of health care today. False statements are highlighted **in bold type** in this section.

1. **High pitch values provide a lower dose.**
   This is true when an automatic tube current modulation/tube current modulation (AEC/TCM) system is not being used or is not available. However, all major CT vendors in the diagnostic CT market provide AEC/TCM systems that aim to maintain the effective mAs selected by the user if the pitch is changed. AEC is also ubiquitous on new interventional CBCT, interventional CT, and CT simulator scanners. See chapter 7, which covers AEC systems.
2. **Standards and regulations, such as complying with ACR accreditation criteria, Joint Commission performance elements, NEMA XR25/26/29, etc., ensure safe use of CT within your clinic.**
   In reality, a diagnostic CT scanner may have hundreds of CT protocols. ACR accreditation, in practice, requires review of about four of them. A Joint Commission auditor may not even review all of the required elements for CT. The cited NEMA standards ensure nothing, as the user can have them installed, but no agency actually checks or requires that they are turned on and set up [21]. See chapters 11 and 12.

3. **In dual-energy CT (DECT), signal on a material density map is proportional to the amount of basis material in that voxel. In other words, everywhere one sees signal on an iodine density map can be interpreted as being due to iodine being present.**
   This is an incorrect interpretation of material density maps [22]. See chapter 9, which covers DECT.
4. **Size-based protocols are only needed for the change from pediatrics to adults, but not within each patient population. For head imaging, AEC is not needed for both pediatrics and adults.**
   These statements are false. In reality, to maintain a constant level of image quality with patient size, scanner output must change with varying patient size. In CT, very small changes in patient size necessitate large changes in CT output. For example, ≈4 cm of soft tissue increase would require the scanner output to double in order to maintain image quality! See chapter 7 and Figures 13.3 and 13.4.
5. **$\mathrm{CTDI_{vol}}$/DLP reflect patient dose. $\mathrm{CTDI_{vol}}$/DLP measurements reflect measurements made on phantoms of equal size to the scanned patient.**
   These statements are false. See chapter 4.
6. **The spatial resolution of a CT image is defined by the in-plane (e.g., x-y dimension) and through-plane (e.g., z-dimension) voxel/pixel size.**
   Especially in industrial CT, vendors and researchers use voxel size to denote spatial resolution. This misrepresentation of spatial resolution is common throughout all flavors of CT. In reality, one can arbitrarily choose pixel/voxel size independently of the spatial resolution of the CT system. A good CT system will use a pixel/voxel size on the order of the maximum spatial resolution of the underlying image data; this is why some people think the spatial resolution is always determined by the pixel/voxel size. See chapters 3 and 5.
7. **Slice numbers reported by vendors reflect the actual number of detector rows.**
   False. Some vendors report double the actual number of detector rows present on their scanners. There is a legitimate reason for this. Some vendors employ the use of a flying focal spot, which toggles the focal spot location back and forth in the z-dimension (along the direction of the detector rows) during the scan. This provides a doubling of the effective number of slices. The important question to ask a CT vendor of any flavor is, "What is the maximum beam collimation in mm, and at that collimation, what is the smallest possible reconstructible slice thickness?" This question will make the actual number of slices and beam collimation clear. See chapters 6 and 17.
8. **Dose from a single diagnostic low dose (about <100 mSv) CT exam will cause cancer to be formed in a fraction of patients.**
   At the present time, this statement cannot be backed up in any scientific capacity. Risk estimations based on large epidemiological studies have yet to provide statistically significant results at low dose levels, so our current estimates for stochastic risk from CT is based on "educated extrapolation" from high doses to low doses. This extrapolation is commonly referred to as the "linear no-threshold hypothesis," which states that any amount of irradiation will provide a finite chance of causing cancer. The data supporting this model is accurate at doses much higher than found in computed tomography. See chapter 4.

9. **The CT number can be trusted across different scanner makes/models and between phases and visits (i.e., prior image comparison) of a single patient.** CT scanners are calibrated to provide a CT number of –1000 HU for air and 0 HU for water. The measurement conditions are usually specified using a small (≈20 cm) water or water-equivalent cylindrical phantom. In clinical practice, the presence of oral/ICV contrast, patient position, patient size, beam energy, vendor, scanner model, slice thickness, contrast dynamics (i.e., scan phase), reconstruction options (i.e., metal mitigation or beam hardening algorithms) will all change the CT number. See chapter 9.
10. **It is good practice to ensure patients receive a certain effective dose by limiting the scan range (which can be used to adjust the exam DLP) or by using a manual technique (which can be used to ensure all patients receive the same clinically appropriate CTDID_{vol}/DLP).** These types of dose-limiting practices are not optimal. They are analogous to giving medication doses of the same amount/volume to all pediatric patients regardless of their individual body weights and ages. Scan length should be determined based on the clinical indication. CT technologists should NOT be instructed to scan more or less of a patient in order to hit a dose target. Sites using manual mA techniques in order to ensure their protocols deliver pre-determined doses ignore the fact that x-ray attenuation is highly nonlinear with patient size. In other words, bigger patients need more dose to have a similar level of image quality. Modern AEC systems allow dose tailoring as a function of patient size. AEC should be used to ensure larger patients receive more dose relative to smaller patients, and proper AEC settings should be used to adjust the relative dose level appropriately based on the indication being imaged. See chapter 7.

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