**Book chapter**

**Recent advances in Computed Tomography**

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***Abstract:***

***CT is a vital diagnostic imaging tool for examining a wide range of clinical disorders. Several important improvements in CT technology have had or are projected to have a substantial therapeutic impact in recent years, including extreme multidetector CT, iterative reconstruction methods, dual-energy CT, cone-beam CT, portable CT, and phase-contrast CT.*** ***It does not necessitate hospitalisation and has also lowered the average length of pre-operative stay by eliminating the need for invasive neuroradiological investigations, which are significantly more expensive per patient than CT. Plain CT is recognised as a non-invasive technology, with radiation doses ranging from less than 1 rem to as high as 9 rem detected at the skin for a single segment. Drip infusions and biphasic approaches, which are more difficult and time-consuming, do not generate intense improvement. Polytomography is superior to CT in showing some extremely minute anatomical characteristics, although it is being phased out in favour of high-resolution CT.*** ***CT is a vital diagnostic imaging tool for examining a wide range of clinical disorders. Several important improvements in CT technology have had or are projected to have a substantial therapeutic impact in recent years, including extreme multidetector CT, iterative reconstruction methods, dual-energy CT, cone-beam CT, portable CT, and phase-contrast CT. This article reviews and illustrates these strategies as well as their clinical uses. Furthermore, upcoming technologies that solve shortcomings in these modalities are explored.***

**Keywords:** Cone-beam CT; dual-energy CT; High Resolution Computed Tomography; extreme Multi-detector CT; iterative reconstruction; phase-contrast CT; portable CT.

**Introduction**

The first of the contemporary slice-imaging modalities, X-ray computed tomography (CT), was introduced into clinical practice in 1972. Reconstructing images mathematically from measured data and displaying and archiving them in digital form was innovative at the time, but it is now standard. CT has showed a consistent rising trajectory in terms of technology, performance, and clinical use, despite forecasts and expert judgments that it would be fully superseded by magnetic resonance imaging in the 1980s. CT not only survived but thrived as a result of the advent of spiral scanning, which marked the transition from slice-by-slice imaging to full volume imaging. CT imaging of complete organs or tissues is now possible because to the emergence of array detector technology in the 1990s. CT is a highly sophisticated and demanding diagnostic imaging modality. Radiologic technologists must be well-versed in technology in order to optimize dose and image quality while providing great patient care. CT scans employ ionizing radiation, or x-rays, in conjunction with an electronic detector array to create an image of a "slice" or "cut" of tissue. Within the scanner, the x-ray beam revolves around the object, causing several x-ray projections to pass through it. CT is founded on the fundamental idea that the attenuation coefficient can be used to calculate the density of the tissue traversed by the x-ray beam.

**Principle of Computed Tomography:**

CT scans are made using a sequence of x-rays, which are a type of electromagnetic radiation. The scanner delivers x-rays from various angles towards the patient, and detectors in the scanner quantify the difference between x-rays absorbed by the body and x-rays transmitted through the body. This is known as attenuation. The density of the scanned tissue determines the amount of attenuation, which is assigned a Hounsfield Unit or CT Number.

High density tissue (such as bone) absorbs more radiation, and the scanner on the opposite side of the body detects less of it.

Low density tissue (such as the lungs) absorbs less radiation, resulting in a stronger signal measured by the scanner.

Conventional x-rays provide the radiographer a two-dimensional image and require the patient to be moved manually to capture the same location from a different angle. CT, on the other hand, can picture the three-dimensional planes of the human body because to the complex mathematical processes involved. This is performed by gathering projections from various angles, and the three-dimensional data is displayed on a two-dimensional monitor via a process known as reconstruction. The data acquired is never an exact reproduction of what is being scanned, but it is close enough to be used for medical diagnostic reasons. CT scans can be utilized with or without contrast, depending on the structure being scanned. Intravenous radio fluorescent contrast administration into the bloodstream can be utilized for a variety of diagnostic applications, including: Used to visualize the cardiovascular system (for example, to look for aneurysms, dissections, or atherosclerotic disorders).To determine whether a tumors is malignant.

The contrast begins to leave the body via the urinary system around 7 minutes after an intravenous infusion of iodinated CT contrast. The contrast can be seen in the ureters as they enter the bladder, resulting in a CT Urogram, which is widely used to replace the classic intravenous pyelogram seen in radiography.

The density of bodily tissue determines the degree to which x-rays are attenuated. As a result, the brightness and contrast of the imaged tissues are affected.

Tissues with high attenuation coefficients (high absorption) appear white, whereas those with low attenuation coefficients (low absorption) appear black. The Hounsfield Scale of radio density measures this.

Tissues with a high Hounsfield score have a high attenuation coefficient, which causes them to appear white:

| **Substance** | **Hounsfield Value** |
| --- | --- |
| Air | -1000 |
| Fat | -70 |
| Water | 0 |
| Blood | 70 |
| Bone | 1000 |

**Advances in Technology**

Most CT systems now support "spiral" (also known as "helical") scanning in addition to the formerly more common "axial" mode. Furthermore, many CT systems can image several slices at the same time. Such advancements allow for the imaging of larger volumes of anatomy in less time. Another technological innovation is electron beam CT, often known as EBCT. Although the premise of obtaining cross-sectional images is the same for single- and multi-slice CT, the EBCT scanner does not require any moving elements to generate the individual "snapshots." As a result, the EBCT scanner acquires images faster than conventional CT scanners.

**Detector Configuration**

By the mid-1990s, helical scans had become speed constrained due to mechanical forces involved with sub second gantry rotation durations and the output needs of x-ray tubes to provide enough flux for a sufficient signal-to-noise ratio. The next performance boost came by collecting data at different body levels in simultaneously, employing more than one row of detectors at the same time. This advancement enabled an increase in volume acquisition speed proportionate to the number of detector rows. In this method, the x-ray tube creates a broad beam of x-rays rather than a small slice; by broadening the collimation to illuminate multiple rows of detectors, more measurements are obtained from the same tube output. At first, two- or four-row multi-detector row CT (MDCT) scanners, but the number of detector rows has gradually increased, with 64-detector row devices now allowing for very high volume coverage. Picture data measurements no longer match to rays orthogonal to the scan axis due to the increased longitudinal breadth of the x-ray beam with MDCT; hence, new reconstruction techniques are necessary to maintain picture quality and prevent distortions.

Each individual detector row in single-detector row CT (SDCT) acts as a single unit and gives projection data for a single region every rotation. Different section widths are obtained in SDCT by modifying the x-ray beam's pre-patient collimation. The detectors of MDCT are further separated along the z-axis, allowing for the capture of numerous sections each rotation. With narrower section widths, MDCT delivers greater and faster z-axis coverage each rotation.

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| https://radiologykey.com/wp-content/uploads/2016/07/C1FF6-1.gif  **Configurations of multi detector computed tomography** |

When four-channel MDCT scanners were introduced in the late 1990s, CT manufacturers used three different detector configurations: (A) 16 detector rows with a uniform thickness, known as uniform array (General Electric); (B) eight detector rows with variable thicknesses, thinner rows centrally and wider rows peripherally, known as adaptive array [Siemens and Philips]; and (C) 34 detector rows with two fixed thicknesses, four thinner rows centrally and 30 thicker rows peripherally, known as adaptive array [i.e., It is worth noting that detectors in four-channel MDCT systems are organized into eight to 34 rows along the z-axis. Nonetheless, because these systems only have four data channels, the number of sections acquired per rotation is limited to four. When a scan with a narrow collimation is desired, the data is measured using four distinct center detector rows, with a narrowly collimated x-ray beam directed over these central detector rows (e.g., 4 1 mm).

A broadly collimated x-ray beam is employed to generate scans with greater section widths, and outputs from two or more nearby detector rows are electronically combined into a single thicker each of the four data channels has its own detector row. Two 1-mm detector rows, for example, can be combined to form a single detector row for 2-mm collimation (4 2 mm), three 1-mm detector rows for 3-mm collimation (4 3 mm), and so on.

For 16-channel MDCT, all CT manufacturers used a hybrid array architecture, with detector rows that are somewhat less than 1 mm thick in the Centre and significantly thicker in the periphery. The length of the z-axis coverage and the number of detector rows, on the other hand, vary greatly amongst CT manufacturers.

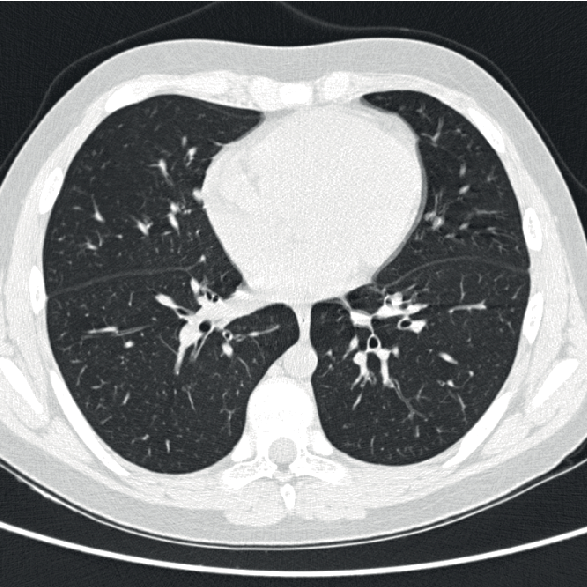
For 64-channel MDCT, CT manufacturers adopted a common detector row design, this time a uniform array with uniform detector row thicknesses. However, like with 16-channel MDCT, the overall number of detector rows and z-axis coverage differ significantly between CT manufacturers.

**Dual energy CT**

Dual energy CT, also known as spectrum CT, is a computed tomography technique that employs two distinct x-ray photon energy spectra to interrogate materials with varying attenuation qualities at different energies. Unlike traditional single energy CT, which yields a single image set, dual energy data (attenuation values at two energy spectra) can be utilized to reconstruct a variety of image types:

1. Pictures with weighted averages (simulating single energy spectra)
2. Virtual mono energetic pictures (attenuation at a single photon energy as opposed to a spectrum)
3. Pictures of material degradation (mapping or eliminating components with recognised attenuation properties, such as iodine, calcium, or uric acid)
   * non-contrast virtual pictures (iodine eliminated)
   * Iodine concentration (maps of iodine)
   * calcium suppression (removal of calcium) o uric acid suppression (removal of uric acid)
4. Maps of electron density.

**High Resolution Computed Tomography**



**High Resolution CT For Chest**

The radiological technique to detecting lung illnesses has been transformed by high-resolution computed tomography (HRCT). Understanding the architecture of the secondary pulmonary lobule is essential for HRCT interpretation. Many aspects of the secondary pulmonary lobule can be seen on HRCT in both normal and diseased lungs. Volumetric scanning uses continuous data capture while the subject is advanced at a consistent rate across the CT gantry, avoiding any inter-scan delay, to provide an accurate assessment of lung disease. HRCT and volumetric helical CT can be used in conjunction to assess focused lung illnesses and airway diseases.

HRCT may identify anomalies that are not clinically significant; consequently, its specificity should be considered in the context of the specific clinical condition. Notably, sarcoidosis staging was determined by CXR findings, and pulmonary involvement determined by HRCT should not be confused with the aforementioned classification. When morphologic changes on paired CXR and paired HRCT scans in sarcoidosis patients were compared, changes depicted on serial HRCT had a stronger agreement with pulmonary function test trends than CXR changes in the characterization of disease improvement or progression.

**Triple phase CT**

The triple-phase liver CT procedure can be used to evaluate localized liver lesions, hyper vascular liver metastases, and endocrine tumors.

It entails the acquisition of a dedicated late arterial phase, portal venous phase, and delayed phase. Not to be mistaken with a four-phase, which incorporates a non-contrast series.

Because of the uniformity of the liver tissue on CT, differentiating hepatic lesions on non-contrast scans is challenging; nevertheless, this exam helps to alleviate that difficulty. Because the portal vein supplies 75% of the liver's blood supply and the hepatic artery supplies the remaining, a later arterial phase is required for the optimal improvement of the parenchyma.

To aid in the classification of hyper vascular liver lesions' vascularity. This test is most commonly used to distinguish a hepatocellular carcinoma from other lesions.

A hepatocellular carcinoma, a very vascular initial lesion, will show hyperenhancement in the arterial phase and venous or delayed phase washout, whereas a hemangioma should match the blood pool in each phase (as the aorta does in the arterial phase, for example).

**COMPUTED TOMOGRAPHY IMAGE FORMATION**

**X-Ray Images**

X-ray imaging is the process of producing x-rays, transmitting those x-rays through material objects, and detecting the beam energy that departs the item. The attenuation of x-rays within an item is determined by atomic-scale interactions, in which each molecule in the object has a cross section for interacting with each x-ray. The x-ray flux reduces by a particular proportion for each unit distance travelled through the object as a result of this interaction. As a result, if a 60 keV x-ray passes through 1 mm of water, it will survive 97.4% of the time. The survival probability multiply at a 95% rate in 2 mm of water.The transmission probability, as expressed by the Lambert-Beer equation, is thus an exponentially decreasing function of the total amount and type of material present:

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where S is the number of surviving signal quanta, I is the number of incident quanta, i is the linear attenuation coefficient for each material, and i is the amount (thickness) of that material present.

The image in projection x-ray imaging is made up of the relative changes in the signal S over a viewing area. The survival chance for a single quantum for a 70-kg person with an abdomen roughly similar to a 20-cm thickness of water is about 2%. An extra 2 mm of aberrant structure would reduce the survival chance to 1.98% (a 1% difference).Given this minor difference in the midst of many overlapping bodily components, projection radiography is clearly limited in its ability to depict anatomic features. In CT imaging, S is measured from many projections, and i is computed from these values for direct display. This approach produces significantly greater relative contrast between nearby structures. A 2-mm calcified nodule, for example, may have a 200% difference in attenuation coefficient relative to adjacent tissue, making it considerably more visible than on a projection radiograph.In Eq. 1, projection x-rays are displayed as a brightness proportional to variations in the transmitted signal S for image viewing. The image attenuation map in CT is displayed in Hounsfield units (HU), which are relative to the attenuation coefficient of water.

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| https://radiologykey.com/wp-content/uploads/2016/07/C1FF7-1.gif  An abdominal slice and its sinogram. |

**Image Reconstruction Using Data from a Two-Dimensional Projection**

The reconstruction of a 2D image section from projection measurements demonstrates the fundamentals of CT image production. An x-ray source and a collection of detectors circle around the patient, measuring x-ray transmission through the body. Each measured value is the sum of all attenuating sections of the patient along a line from the x-ray source to the detector. As a result, a uniform circular disc with a circular profile will have the greatest attenuation at its centre. Prior to reconstructing images, raw projection data is obtained by collecting line measurements from various view angles during one round of the gantry. A sinogram is produced by the raw projection data. The sinogram can be shown graphically, with the y-axis (rows) representing detector measurements and the x-axis (columns) indicating detector measurements at one gantry point. The sinogram image has a fascinating pattern, but the overlapping shapes make it difficult to analyse. As a result, a method for determining and computing the original picture attenuation is required.

One problematic way for finding the source picture is to consider the sinogram and image as a linear algebra problem. Each measurement is an equation that sums all the image pixels along a ray to the detector; the set of all equations can then be solved for the unknown picture pixels. The scale of this problem is intimidating since it involves 512 512 (i.e., more than one quarter million) variables and 768 1,400 (i.e., more than one million) measures, necessitating matrix operations that overload even modern computers. Other mathematical methods for solving for pictures, such as iterative algorithms or maximum likelihood optimisation, can be utilised, but they are likewise too computationally demanding for widespread clinical application.

The mathematical approach that enabled CT reconstruction is known as filtered back projection.

Theoretically (5), if the projection measurements have certain properties (they all lie in one plane, they are made up of equally spaced gantry steps that cover at least one half revolution, and the detectors are equidistant and cover the entire object to be reconstructed), then the attenuation (image) at any point within the scanner field of view can be calculated by summing a certain weighted combination of the measurements. This weighted summing method is known as a kernel (for more information, read the section titled Reconstruction Kernel later in this chapter). Measurements from neighboring detectors are removed and measurements from the detector immediately intercepting the pixel are added. Depending on the application, different kernels might be constructed to deliver sharp, crisp images or to smooth out noise.

This weighted summing method is known as a kernel. Measurements from neighboring detectors are removed and measurements from the detector immediately intercepting the pixel are added. Depending on the clinical application, different kernels can be constructed to deliver sharp, crisp images or to smooth out noise. This method, which was widely used by CT makers in the early days of CT, may be executed very efficiently by computers or dedicated hardware modules, either directly or through Fast Fourier Transform techniques.

**Reconstruction of Images from Three-Dimensional Projection Data**

The filtered back projection procedure necessitates the confinement of image data to a single plane. Helical CT acquires 3D volumes rather than single data sections, necessitating the development of new reconstruction techniques.

Linear Interpolation in **Single-Detector Row Spiral Computed Tomography,** Because the patient table moves continually in spiral scanning, there are only a few (or none) perfectly equivalent gantry measurements that are aligned in the same plane for 2D filtered back projection at any given longitudinal or z-location. The greater the pitch (how quickly the CT table moves relative to the detector collimation), the more the gantry measurements separate and deviate from the plane. Missing gantry measurements are estimated by using the average of the nearest (in the z-axis) data gathered to provide a complete set of measurements for filtered back projection.

This approach is used in two variations. The first method, 360LI, averages observations separated by one revolution. To create projection data for a target image plane, this method linearly interpolates two gantry measurements on either side of the image plane that are closest to the image plane and 360 degrees apart (i.e., measured in subsequent revolutions). The 360LI approach has the problem of having a huge amount of travel in one revolution, and if structures change dramatically over this distance, blurring or partial volume averaging will occur.

The second approach, known as 180LI, takes advantage of the symmetry between the x-ray source and detector throughout the gantry, i.e., when the source-detector positions are one half rotation (180 degrees) apart, the measured ray is nominally the same. The 180LI technique takes advantage of the fact that an interpolation partner is already available for each measurement beam after about half a revolution, when the x-ray tube and detector have exchanged places. A complimentary ray is a virtual, geometrically derived ray. Because the 180LI approach uses smaller z-distances, it suffers from less blurring. (The similar trick can be used in cardiac imaging to reduce temporal blur and shrink the time window for an image snapshot.)

Z-Interpolation vs. Z-Filtering in **Multi-detector Row Spiral Computed Tomography**: The early multidetector row scanners contained two or four detector rows, with data measurements interpreted as a simple parallel stack of independent detector rows. The 360LI and 180LI utilized in SDCT spiral reconstruction methodologies can be directly extended to spiral MDCT in this scenario. The closest row measurements to the target plane could then be used to build planes of measurements using linear interpolation (either 360LI or 180LI), a technique known as advanced single-slice rebinning. The interpolation calculation may be done quickly and is essentially the same as single-row scanning. Interpolation in the 360LI interpolation approach can be accomplished utilising rays measured at the same projection angle by various detector rows or in consecutive rows. The scanner's rotations are separated by 360 degrees. For spiral interpolation in the 180LI reconstruction method, both direct and complementary rays can be employed. CT scanner manufacturers presented many mathematical algorithms, such as z-interpolation or z-filtering, for weighing and interpolating neighboring rays for the target image plane.

**Cone Beam Reconstruction Using Broad Beam Multi detector or Flat-Panel Computed Tomography**

When the number of detector rows exceeds four, the cone-beam angle between detector rows must be taken into account. For picture reconstruction, some manufacturers employ modifications and expansions of nutating-section methods. These methods divide the 3D reconstruction process into a series of standard 2D reconstructions on slanted intermediate picture planes, allowing them to take advantage of well-established and highly fast 2D reconstruction techniques. Two examples are adaptive multiplanar reconstruction (Siemens) and weighted hyperplanar reconstruction (GE Medical Systems). Other manufacturers (Toshiba, Philips) have adapted the Feldkamp technique, an approximate 3D convolution back-projection reconstruction first introduced for sequential scanning, to multisection scanning.

Using this method, the measurement rays are back projected onto a 3D volume along the lines of measurement, accounting for their cone-beam shape. Three-dimensional rear projection, on the other hand, is computationally difficult and necessitates the use of dedicated hardware to achieve acceptable image-reconstruction timeframes. A recent topic of research is the development of methods that account for the cone-beam geometry of measuring x-rays.

**Metrics Imaging**

Image quality is the ultimate measure of an imaging system, but it is difficult to define and quantify. Image quality is typically judged qualitatively and subjectively in healthcare situations. To characterise system performance, communication theory describes the fundamental parameters of information flow as signal, resolution, distortion, and noise. Image quality is typically described using quantitative and objective metrics such as spatial resolution, contrast resolution, temporal resolution, noise, and artefacts. These metrics are influenced by CT scanner apparatus and scan factors and are frequently used to evaluate a CT scanner's performance.

**Signal**

A map of some physical quantity, either directly measured or derived from measurements, is represented by a picture. The visual signal can be continuous, as in a screen-film x-ray or 35-mm photograph, or discrete, as in a medical image displayed on a computer monitor. The quantity measured in the CT acquisition procedure is the attenuation of the x-ray beam (similar to a projection x-ray), with a continuous physical electrical signal representing x-ray energy flux transformed to a discrete digital value. A digital image representing the attenuation coefficient of the substance in the object is calculated from a set of these measurements. The map is made up of pixels (picture elements), often in the form of a square array of 512 pixels on each side. When numerous slices are combined to form volume data sets, the 3D map is transformed into a collection of voxels (volume elements). The original measurements may be 16-bit data (enabling a range of values spanning a factor of 64,000), however the reconstructed images are commonly 8- or 12-bit data (allowing a range of up to 4,095). The signal is considered to be linear with the physical parameters of the exhibited item. For example, doubling the density of the contrast medium in a voxel increases the pixel value by a factor of two.

The picture signal contains information in the form of patterns of change in the image. The magnitude of such change is defined by contrast, which is the variance of local values from the surrounding values.

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