CHAPTER ONE

CT PHYSICS AND INSTRUMENTATION – MECHANICAL DESIGN

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BASIC CT UNIT ANATOMY

A computed tomography (CT) unit consists of a gantry, a patient table, hardware equipment, an operator console and optionally additional workstations.

The gantry is a doughnut-shaped ring containing the X-ray tube, the detector array and associated equipment. The central hole in the gantry accommodates the patient on a sliding table. The X-ray tube rotates around a slice of patient anatomy. This slice represents the X-Y plane, with the X-axis being horizontal and the Y-axis vertical. The isocenter of the gantry is the central point of this plane. The third dimension is represented by the Z-axis, which is along the orientation of the patient table. The patient bed is a sliding tray on a fixed table with an adjustable height and a defined capacity of forward motion. The operator console is located in another room or behind radioprotective screening, and allows operation of the CT units. Additional workstations can be used to review processed image data, but usually not raw data processing.

X-RAY TUBE

Basic anatomy of the X-ray tube

An X-ray tube is a vacuum tube that produces X-rays. It is composed of a cathode (filament) and an anode (target). The cathode cup is negatively charged and incorporates a wound tungsten filament that emits electrons when heated. The anode consists of a disk of tungsten or a tungsten alloy with an annular target, called the focal track, close to the edge. The anode disk is supported on a long stem that is supported by ball bearings within the tube. The anode can be rotated by electromagnetic induction from a series of stator windings outside the evacuated tube.

The X-ray tube is enclosed in a housing unit filled with insulating oil. This oil provides electric shielding from the tube voltage, X-ray protection and transmits heat generated in the housing unit to the unit's surface. The exterior of the housing unit is cooled with a fan, and insulating oil is cooled by passing it through a heat exchanger.

Low-power applications use stationary anode tubes, while for most mid-range and high-performance applications there is a need to utilize rotating anode tubes.

Basic physiology of the X-ray tube

A current of a few amperes (4–8 A) heats the tungsten filament that releases electrons (thermoionic emission) in the vacuum. A high-voltage power source ('tube voltage') ranging from 30 to 150 kilovolts (kV) is connected across cathode and anode to accelerate the electrons producing an electron flow ('tube current'). These electrons collide with the anode material and about 1% of their kinetic energy is converted into X-rays, usually perpendicular to the path of the electron beam. The remainder of energy is converted into heat, causing the

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X-ray tube to warm up during operation. The temperature of the focal track can increase quickly to 1000–1500°C. Heat diffuses by conduction throughout the anode body and by thermal radiation (infrared radiation) to the tube housing (80%). Heat is removed from the tube housing by convection to the surrounding atmosphere.

Many X-ray systems, including CT, have built-in safety features that will not allow the equipment to be operated in 'overheated' conditions. The temperature cannot be measured directly in the focal track. It has to be evaluated based on indirect values that characterize the ability of the anode to store the heat generated during the X-ray emission, such as the anode heat capacity, the anode dissipation/cooling rate or the tube dissipation.

Specificities for CT

Since the invention of CT, its demands regarding the X-ray source never ceased to increase and are largely superior to those of radiography. These specific requirements can be summarized with higher scan power, shorter rotation times (maximum rotation speed), shorter cool-down times and smaller focal spots without compromise on resolution and image quality. In older CTs, the generator capacities and anode disk's heat storage capacities were insufficient and long time interruptions were needed.

Scan power: typical values for the maximum power are 20/40–100 kW with the high voltage range ranging from 80 to 150 kV.

Focal spots: X-ray tubes use typical focal spot sizes of 0.5–1.2 mm. Specific innovations for CT are the 'flying focus' allowing for control of the focus position on the anode during the scan or the electromagnetic control of the electron beam, which allows switching of the focal spot position both in the fan and in the Z-direction, providing overlapping sampling.

Rotation speed: the traditional glass tube technology is not adequate in terms of required precision and stability to sustain the very high rotation speed, up to 10 500 rotation/min, of high-performance tubes. Despite its higher thermic dissipation and lower cost, glass has been replaced by the metal ceramic technology, which is more precise and better able to sustain the constraints related to the rotation speed.

Cool-down time: different approaches can be used on their own or in combination to shorten the cool-down times and improve the heat storage capacities of CT.

- The '***brute force approach***' was the main way used for three decades. This approach consists of an increase in the thermal capacity of the anode by increasing its diameter and mass. This system has obvious limitations, as it still uses radiative cooling.
- The '***material approach***' is based on a slow evolution in the materials used for the anode.

  Direct:
  - Use of circular grooves in the anode support to increase the contact and improve cooling.
  - Use of special liquid metal vacuum bearings that allow faster anode rotation.
  - Focal track made of a mixture of rhenium and tungsten. Rhenium has a higher linear expansibility than tungsten and slows the rate at which anode crazing occurs.
  - Anode 'compound'/increased thermal capacities: use of molybdenum or graphite with tungsten in the anode disk. Molybdenum has twice the specific heat capacity and half the density of tungsten. Graphite has an even higher specific heat capacity and a quarter of the density of molybdenum. It increases thermal capacity.
  - Replacement of the ball bearings by a liquid metal (gallium) that allows the evacuation of heat by conduction.

  Indirect:
  - Multiple detectors allow reduction of the heat produced via reduced scan duration by a factor approximately equivalent to the number of rows. Manufacturers developed systems with up to 1000 rows.
- The '***paradigm shift***' corresponds to innovations in X-ray tubes.

In 2000, Siemens developed the Straton tube, also called the rotating envelope tube, for its high-tech scanner. This tube uses direct convective cooling, exclusively of the anode, with a cooling oil stream at the anode's back surface. As a result, the cooling rate is vastly increased to 4.8 MHU/min, eliminating the need for large heat storage capacities of the anode disk and reducing waiting times due to anode cooling in the clinical workflow.

This lighter tube also presents a solution for the acceleration/pressure centrifuge high G (above 20 G).

Another innovation on the X-ray tube in relation to CT is the dual-source CT (two tubes, two detector fans); the main advantage of this architecture is its improved temporal resolution. In today's CT scanners, the gantry rotation time is reduced to about 0.35 s and it is mechanically challenging to reduce that time even further, which justifies the renewed interest in multi-source architectures.

COLLIMATORS AND FILTRATION

CT systems feature various collimators, filters and shielding designs, which provide filtration of the X-ray spectra, definition of the measured slices, guarding detectors against scattered radiation and general radiation protection. These vary from scan to scan but always offer the same functions.

Collimation

Collimation in CT serves to ensure good image quality and to reduce unnecessary radiation doses for the patient.

Collimators are present between the X-ray source and the patient (tube or pre-patient collimators) and between the patient and the detectors (detector or post-patient collimators).

The tube collimator is used to shape the X-ray fan beam before it penetrates the patient (restrict the X-ray flux applied to a narrow region defines the shape of the X-ray beam). It consists of a set of collimator blades made of highly absorbing materials such as tungsten or molybdenum. The opening of these blades is adjusted according to the selected slice width and the size and position of the focal spot. It defines slice thickness for single-slice CT. Tube collimators define the dose profile according to the required slice thickness. Post-patient collimators improve the slice sensitivity profile by giving a more rectangular shape. Table 1.1 shows the features of CT collimators.

Filtration

The X-ray photons emitted by the X-ray tube exhibit a wide spectrum. The soft, low-energy X-rays, which contribute strongly to the patient dose and scatter radiation but less to the detected signal, should be removed. To achieve this goal, most CT manufacturers use X-ray filters.

The inherent filtration of the X-ray-tube, typically 3 mm aluminium equivalent thickness, is the first filter. In addition, flat or shaped filters can be used. Flat filters, made of copper or aluminium, are placed between the X-ray source and the patient. They modify the X-ray spectrum uniformly across the entire field of view. Because the cross-section of a patient is mostly oval-shaped, some manufacturers use shaped (or bow-tie) filters. These filters have an increased thickness from center to periphery, allowing them to attenuate radiation hardly at all in the center but strongly in the periphery. They are made from a material with a low atomic number and high density, such as Teflon.

In some machines comb-shaped collimators close to the detector array are used to decrease the effective detector element width and thus increase the achievable geometrical resolution.

DETECTOR SYSTEMS

The detector is the system for quantitative recoring of the incident ionizing radiation. It acts in two steps.

Table 1.1
CT collimator features.

Collimator type	Tube side – detector side	Location	Fixed/ adjustable	Aim
Pre patient	Tube	Very close to the focus (tube housing)	Fixed	– To reduce the generated radiation roughly to the maximally anticipated beam for the given detector and geometry – Blocks 99% of the photons emitted by the tube
Pre patient	Tube		Fixed	– To exactly define the maximum permitted beam
Pre patient	Tube	As far as possible from the focus (close to gantry housing)	Adjustable	– Variable collimation to the desired slice width or multiple slice widths – To minimize penumbra caused by finite focus size
Post patient	Detector		Fixed	– To minimize signal contributions from scattered radiation – Width = maximal collimation
Post patient	Detector* (optional)	Between the single detector elements	Adjustable	– To minimize signal contributions from scattered radiation

*Can only be implemented in scanners with a rotating detector.

1. The reception of the incident X-ray photon via X-ray-sensitive detector elements with a specific geometrical configuration.
2. The transformation of the X-ray photon into a corresponding electrical signal, that is then amplified and converted from an analog to a digital form (via analog-to-digital converters). This step is relatively easily specified and submitted to few fluctuations.

There are two detector types.

- Ionization chambers, mostly filled with the noble gas xenon under high pressure. Gas detectors have become obsolete due to their limited detection efficiency and the difficulty in manufacturing them for multi-row design.
- Scintillation detectors, in the form of crystals such as cesium iodide or cadmium tungstate, and ceramic materials such as gadolinium oxysulfide. These detectors are now predominantly used mainly because of their short decay time, which is an essential factor in subsecond scanning times since. Ultra-fast ceramic (gadolinium oxysulfide based) have superior characteristics in this area, making them the best choice for spatial resolution and image quality.

The alternative detector concept is the flat-panel technology. Potential advantages are the possibility to scan with wider cone angles without the need to develop detectors with 1024 rows or more and the high spatial resolution, particularly in medium to large field of view scans. Flat-panel detectors were developed for digital radiography and their use for CT is currently being explored by manufacturers.

GANTRY ANATOMY

Third generation

With third-generation CT, simultaneous rotation of the X-ray tube and detector array became possible (rotate/rotate geometry, rotating gantry). Moreover, the number of detectors and the angle of the fan beam were increased considerably so that the X-ray beam could scan the entire patient. The translational motion of first- and second-generation CT scanners could therefore be eliminated, which reduced the scan times substantially.

In the beginning, third-generation scanners suffered from the problem of ring artifacts. Each detector in third-generation scanners is responsible for the data corresponding to a ring in the image. Detectors close to the center of the detector array are responsible for a ring with a smaller diameter than those detectors towards the periphery of the detector array. Because there is always a certain amount of electronic drift associated with each detector, this causes gain changes between detectors, finally leading to ring artifacts. Today, modern technology has overcome this problem so that third-generation CT scanners are free of ring artifacts.

Multislice CT systems always use a third-generation technology and they provide scan times as short as 0.5 s.

Fourth generation

Because of the problem of ring artifacts with third-generation scanners, fourth-generation CT scanners were designed. The detectors are placed separately in a stationary 360° ring around the patient and only the X-ray tube rotates (rotate/stationary geometry). Whereas in third-generation scanners, data are acquired by the detector array simultaneously, a single detector collects the data in fourth-generation CT over the period of time that is needed for the X-ray tube to rotate through the arc angle of the fan beam. Each detector also represents its own reference detector. In this way, ring artifacts were avoided in fourth-generation scanners.

This technology requires many detectors because the detector array covers a 360° angle. It is not used nowadays to design multislice CT units because of the high costs for such an immense number of detectors.

Slip-ring technology

In third- and fourth-generation scanners, the X-ray tube rotates around the object. This also applies to the detectors in third-generation scanners. In combination, this is also referred to as a ‘rotating gantry’, although not all parts of the gantry rotate. These components require a number of electrical connections for high-voltage power, data transmission and control. In most early CT systems, the connections between the components on the rotating side of the gantry bearing and the power sources, computers, etc., on the stationary side of the bearing were made using cables. They were of finite length and allowed a rotation of perhaps 700°. As a result, these systems had to stop and reverse rotation directions between images.

The alternative to this cable system is the slip-ring technology. It allows the continuous circular rotation of the X-ray tube and other components of a CT system. In a slip ring, electrical brushes allow connections between continuously rotating and stationary components. The slip-ring design made it possible to achieve greater rotational velocities allowing shorter scan times. It finally enabled the design of the modern helical CT scanner.

Multislice CT

Helical CT represents a CT system using slip-ring technology in which continuous X-ray tube rotation is used along with simultaneous and continuous table translation through the gantry. The X-ray tube describes a helical path around the object. The term 'helical CT' is equivalent to spiral CT, which is actually an inaccurate term (a spiral decreases in diameter). Helical CT scanners are named single section (single slice, single detector row), dual section (dual slice, dual detector row) or multisection (multislice, multidetector, multirow) according to the maximum number of slice images generated per gantry rotation.

Helical CT technology makes it possible to image a given volume much more quickly (e.g. 30 s for the entire abdomen). More importantly, though, it allows a volume to be imaged during a more consistent phase of contrast enhancement. It is of significant benefit for CT angiography and multiphase abdominal imaging. The extent of sequential coverage, or the total time of scanning, is generally limited by X-ray tube heating.

The relationship between the incremental table movement and the selected slice width during one rotation of the gantry is described as 'pitch'. In single-slice CT the pitch describes the ratio of the table movement per 360° gantry rotation to the collimator width (*collimator pitch*). In multislice CT, the *detector pitch* describes the ratio of the table movement per 360° gantry rotation to the detector width. The *collimator pitch*, then, defines the ratio of the detector pitch to the number of detector rows in multislice technology. The pitch influences patient dose, scan time and image quality. Increasing the pitch decreases scan time and reduces motion artifacts. However, the effective section thickness as well as image noise increases, too. For clinical studies, a pitch of 1–1.5 is commonly used.

Since the helical data set does not correspond to sequential plane data, it needs to be reconstructed via interpolation into planar image data sets before the actual CT reconstruction.

With single-slice CT, detectors are rather wide in the Z-axis, e.g. 1 × 20 mm. Almost the entire detector element is actively detecting radiation, and slice thickness is determined by the collimator width. Per default, the collimator width is always smaller than the detector width. Therefore, for single-slice CT, slice thickness can be decreased via smaller collimator width; however, utilization of the X-ray beam is lower, therefore signal-to-noise-ratio decreases as well. This may be partially compensated by increasing the mAs. As an advantage, partial volume averaging decreases and spatial resolution is improved with thinner slice thickness.

With multislice CT, detectors are much smaller (e.g. <1 × 1 mm). The detector size determines the smallest possible slice thickness and the collimators determine the number of detectors used. If only the central two detectors are used, the slice width can be reduced below the detector width. To allow for variation in slice width and to decrease scan time, the signals from multiple rows of detector elements can be combined, so-called binning. Binning can be performed during the requisition or from raw data after scanning.

There are two detector array designs in multislice systems: those with detector elements of equal width (equal-width design) in each detector row and those with detector elements of unequal width in the different detector rows (unequal-width design).

With multislice CT, a much higher anatomic coverage can be achieved with the same pitch and slice thickness than in single-slice CT. For the same anatomic coverage and scan time, with single-slice CT one has to either increase the pitch or the slice thickness. But image quality is then degraded considerably.

There are many advantages with multislice CT. Scanning is faster, providing better temporal and contrast resolution and fewer motion artifacts. Consequently, multiphase studies (e.g. arterial, venous, portal phase studies) became possible. Thinner slice thicknesses are possible, which improves spatial resolution and reduces partial volume averaging. Due to more patient length scanning per rotation, higher X-ray tube current settings may be used, which in turn reduces image noise.

Table 1.2 shows typical performance characteristics for a CT scanner in 2010.

Moving gantry

Most CT units include a moving table and a fixed gantry housing. However, for certain purposes, moving or sliding gantries have been developed. Instead of the table moving into the gantry, as in conventional CT, in

Table 1.2
Performance characteristics for a CT scanner in 2010.

Power	60–100 kW
Rotating time per 360°	0.33–0.4 s
Slice width	0.5–0.6 mm
Simultaneously scanned slices	64
Data per helical scan	200–4000 MB
Image matrix	512 × 512
z-coverage per rotation	20–40 mm
Scan times 'whole body'	10–30 s
Scan range	>1000 mm
Isotropic spatial resolution	0.4–0.6 mm
Contrast resolution	3 HU
Effective dose	1–20 mSv

this case the table is fixed and scanning is accomplished by moving the gantry over the patient. In human oncology, for example, it may be an advantage during the course of irradiation if a CT scan can be performed for treatment planning adjustments immediately prior to irradiation. For this, patients are positioned on a common and fixed treatment table, which is integrated in a combined CT and linear accelarator irradiation system. The irradiation system and the CT gantry are positioned on opposite ends of the table so that, by rotating the treatment table, linac radiotherapy or CT scanning can be performed. Moving gantry systems are also designed for usage during surgery or angiography.

SCANNING MODES

A routine scan requires a scout radiograph for anatomical orientation and scan region (slice) selection and scan performed in sequential or helical mode.

Scout radiograph (survey radiograph, localizer radiograph, scanogram, topogram, scout view)

A survey radiograph, similar to a conventional radiograph, is very useful for selection of single slices or complete scan regions. This radiograph is taken with a low dose and low spatial resolution by transporting the patient slowly through the field of measurement with the X-ray tube in a fixed position with radiation emitted continuously or in pulsed mode. Lateral scanograms are particularly useful to select the gantry tilt according to anatomy.

Sequential scanning (axial scanning, single-slice scanning)

For a long time, CT examinations consisted of scanning single slices sequentially. A single slice is scanned, then the patient is transported for a scan increment, mostly equal to the chosen slice thickness. Then, a second scan is taken and the procedure is repeated. This examination mode is relatively time-consuming and has been largely replaced by the faster helical CT. One fundamental disadvantage is that overlapping images for 3D image reconstruction are generally not available.

Modern scanners offer automated and therefore fast modes for scanning single slices sequentially. Cardiac scanning may be a future indication.

Dynamic scanning (serial scanning)

Dynamic CT is used to record temporal changes in the density characteristics of an object. Typically, dynamic scanning is used to assess contrast medium dynamics. A representative selected slice is scanned repeatedly or multiphase examination of a complete organ is performed before, during and/or after administration of contrast medium. The observed changes may represent physiological processes, such as heart motion or breathing, or pathological processes such as portosystemic shunts. Dual-phase CT angiography is a minimally invasive technique, which provides an excellent 3D representation of portal and hepatic vascular anatomy.

Material-selective scanning (dual-energy CT)

Dual-energy methods serve to obtain information about the material composition in the tissues examined. To achieve this, a selected slice is scanned with two different spectra, i.e. with different high-voltage values and possibly with different filtration. This can be done in two successive scans or by switching the high voltage rapidly from projection to projection.

TABLE DESIGN

Many CT tables are made of a carbon fiber material because it will not cause artifacts when scanned. The movement of the table is referred to as incrementation (incrementation indexing). All table designs have weight limits that if exceeded may compromise increment accuracy. The maximum table load on actual CT machines is between 200 kg and 330 kg. Various table attachments and positional aids are available for specific body parts. For large animal CT these are usually custom made (for details, *see* Chapter 39).

Proprietary CT Terminology (Table 1.3)

Table 1.3
Proprietary CT terminology.

	GE Medical	Siemens	Philips	Toshiba
X-ray tube preparation				
Warm up and calibration	Tube warm-up	Check-up calibration	Tube conditioning	Warm-up
Scanning modes				
Planning radiograph	Scout	Topogram	Pilot or surview	Scano
Planning single CT scan	N/A	N/A	N/A	S&V (scan & view)
Sequential scan mode	Axial	Sequential	Axial	S&S (scan &scan)
Thick slice reformatting of sequential thin slices	Addition	Average	N/A	StackScan
Helical scan mode	Spiral	Helical	Spiral	Helical
Dynamic CT	Dynamic cine (continuous scanning)	Cine (dynamic, serio)	Dynamic	Dynamic
Scan planning				
Table	Couch	Couch	Patient table	Couch
Tube heat	Heat units	Heat	Tube heat (load)	Tube (OLP)
Scan range, length	Range	Range	1500	Range
Scan FOV	Scan FOV	Scan FOV	Scan FOV	C FOV
Reconstruction FOV	D FOV	Recon FOV (ZOOM)	FOV	D FOV
Bolus tracking	Smart prep	Care bolus	Bolus Pro	SureStart
Collimator pitch	Pitch	Pitch	Pitch	(Thickness)
Dose management software	Smart mA & Auto mA	Care dose 4D	Dose Wise	Sure Exposure
Image reconstruction interval	Interval	Increment	Continuous, Overlapping	Interval
Image reconstruction algorithm	Algorithms	Kernels	Recon algorithm	Recon FC
Additional image filtration	E1-4	Filters	Custom Image Filters	Filter (QDS)
Matrix, reconstruction	Center X,Y (0,0)	Center X,Y (0,0)	Center X,Y (0,0)	Center X,Y (256,256)
Cone beam reconstruction algorithm	Cross beam	AMPR/ASSR	COBRA	TCOT

(*Continued*)

Table 1.3
(*Continued*)

	GE Medical	Siemens	Philips	Toshiba
Post-processing				
Archive	Image works	Browser	Patient catalogue	Directory
Visualization software	Image display	Sub-tree	Viewer	I-selector
Multiplanar reconstruction (MPR)	Direct MPR	Real-time MPR	Real-time MPR	Multiview (auto MPR)
DICOM protocols transfer	Network	Export	Copy series	Transfer
Subtraction software	Innova 3D	Subtraction Syngo neuro DSA	Subtraction	Sure Subtraction
Cardio software	CT Cardiac Imaging	Syngo circulation	Cardiac tools/Best to beat	Sure Cardio
Plaque analysis software	Color mapping	Syngo plaque MAP	Plaque Analysis	SurePlaque
Gating software application	SnapShot Pulse	Syngo circulation	Step & shoot cardiac	SureCardio Prospective
Fluoro CT software	CT fluoro	Care vision	CCT (continuous CT)	CT Fluoro SureFluoro
CT endoscopy/endo 3D	Navigator	Fly	Endoview	3D

Further Reading

Bushberg JT, Seibert JA, Leidholdt EM and Boone JM (2002) Computed tomography. In: Bushberg JT, Seibert JA, Leidholdt EM and Boone JM (eds) *The essential physics of medical imaging* (2e), pp 327–72. Philadelphia, PA: Lippincott Williams & Wilkins.