الجامعة التقنية الوسطى
كلية التقنيات الصحية والطبية/ بغداد
قسم تقنيات الأشعة

تقنيات اجهزة التصويرالمقطعي المحوسب

المرحلة الثانية / الكورس الثاني

Middle Technical University (MTU) الجامعة التقنية الوسطى
College of Health and Medical Techniques -Baghdad كلية التقنيات الصحية والطبية/ بغداد
Radiological Techniques Department قسم تقنيات الأشعة

المادة: تقنيات اجهزة التصويرالمقطعي المحوسب

Computed Tomography EquipmentsTechniques

Second stage/ 2nd coarse المرحلة: الثانية / الكورس الثاني

Title: العنوان:

History of Computed Tomography

Limitations of conventional radiography

Name of the instructor: اسم المحاضر:

م.د.لمياء فاضل عبدالحسين

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Target population: الفئة المستهدفة:

Students of second class طلبة المرحلة الثانية

Introduction: المقدمة:

Computed tomography (CT) is an imaging procedure that uses special x-ray equipment to create detailed pictures, or scans, of areas inside the body. It is also called computerized tomography, or computerized axial tomography (CAT). The term *tomography* comes from the Greek words *tomos* (a cut, a slice, or a section) and *graphein* (to write or record). Computed tomography (CT) is noninvasive and produces cross-sectional images of the body. Each cross-sectional image represents a “slice” of the person being imaged, like the slices in a loaf of bread. These cross-sectional images are used for a variety of diagnostic and therapeutic purposes.

Scientific Content: المحتوى العلمى:

Limitations of Film-Based Radiography

➥ The major shortcoming of radiography is the super- imposition of all structures on the film, which makes it difficult and sometimes impossible to distinguish a particular detail (Fig.1). This is especially true when structures differ only slightly in density, as is often the case with some tumors and their surrounding tissues.

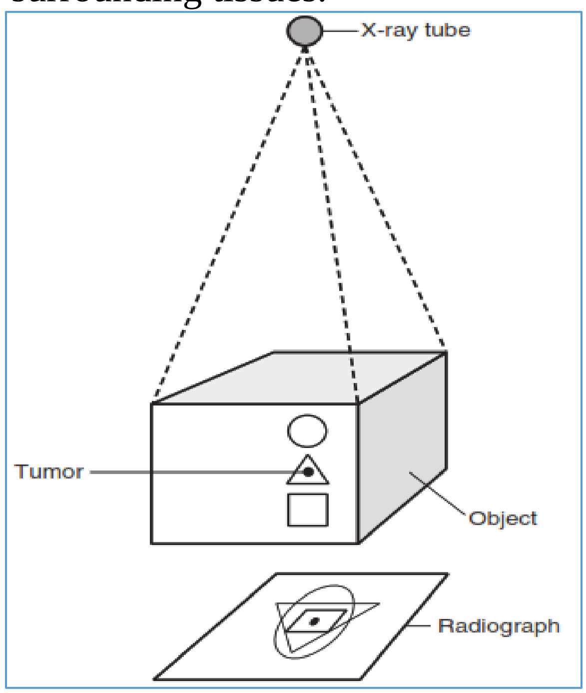

Fig. 1: The major shortcoming of radiography is that the superimposition of all structures on the radiograph makes it difficult to discriminate whether the tumor is in the circle, triangle, or square.

➥ A second limitation is that radiography is a qualitative rather than quantitative process (Fig. 2). It is difficult to distinguish between a homogeneous object of nonuniform thickness and a heterogeneous object (Fig. 2) (includes bone, soft tissue, and air) of uniform thickness.

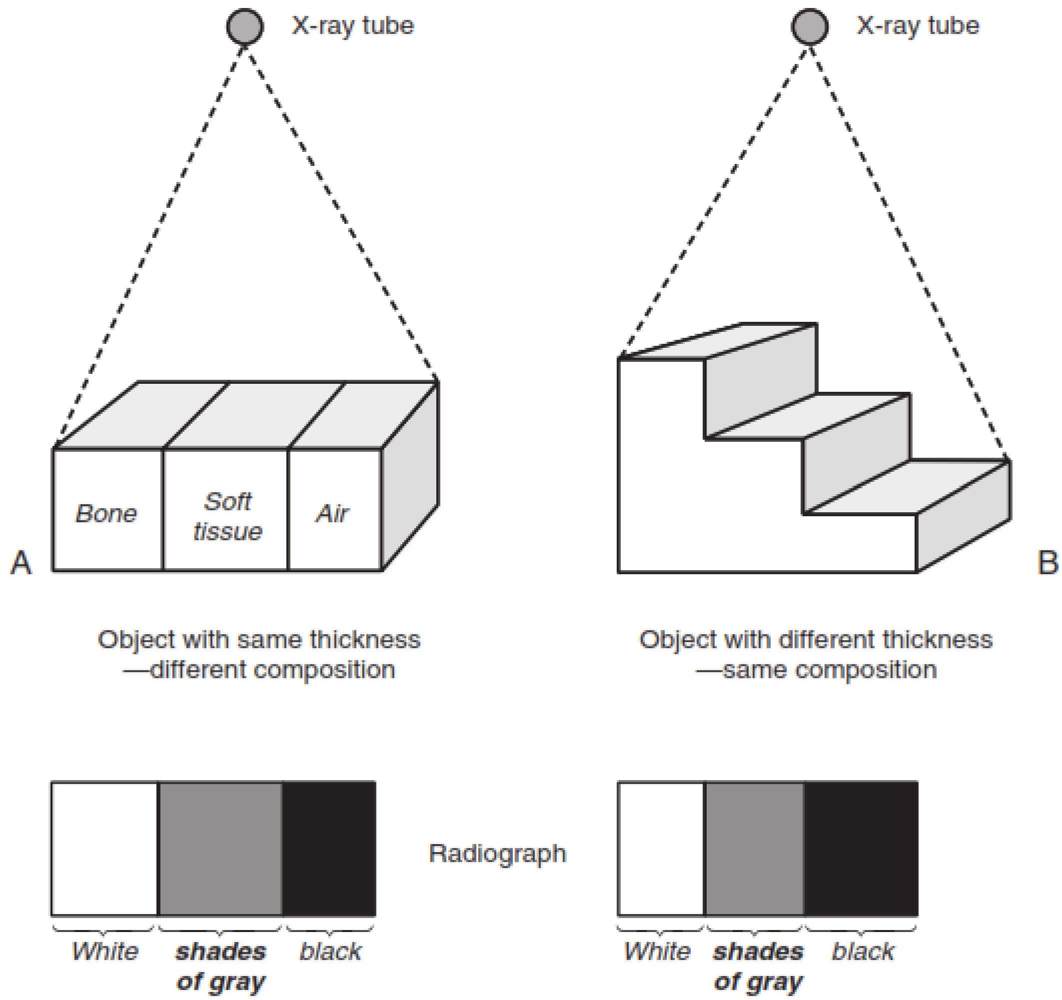

Figure (2): Radiography is a qualitative rather than quantitative procedure. Two radiographs can appear the same although the two objects, **A** and **B**, are entirely different.

Limitations of Conventional Tomography

The problem of superimposition in radiography can be somewhat overcome by conventional tomography . The most common method of conventional tomography is sometimes referred to as *geometric tomography* to distinguish it from CT (Fig. 3).

When the **x-ray tube** and film are moved simultaneously in opposite directions, unwanted sections can be blurred while the desired layer or section is kept in focus.

The immediate goal of tomography is to eliminate structures above and below the focused section, or the focal plane. However, this is difficult to achieve, and under no circumstances can all unwanted planes be removed. The limitations of tomography include persistent image blurring that cannot be completely removed, degradation of image contrast because of the presence of scattered radiation created by the open geometry of the x-ray beam, and other prob- lems resulting from film-screen combinations.

In addition, both radiography and tomography fail to adequately demonstrate slight differences in subject contrast, which are characteristic of soft tissue.

Radiographic film is not sensitive enough to resolve these small differences because typical film- screen combinations used today can only discriminate x-ray intensity differences of 5% to 10%.

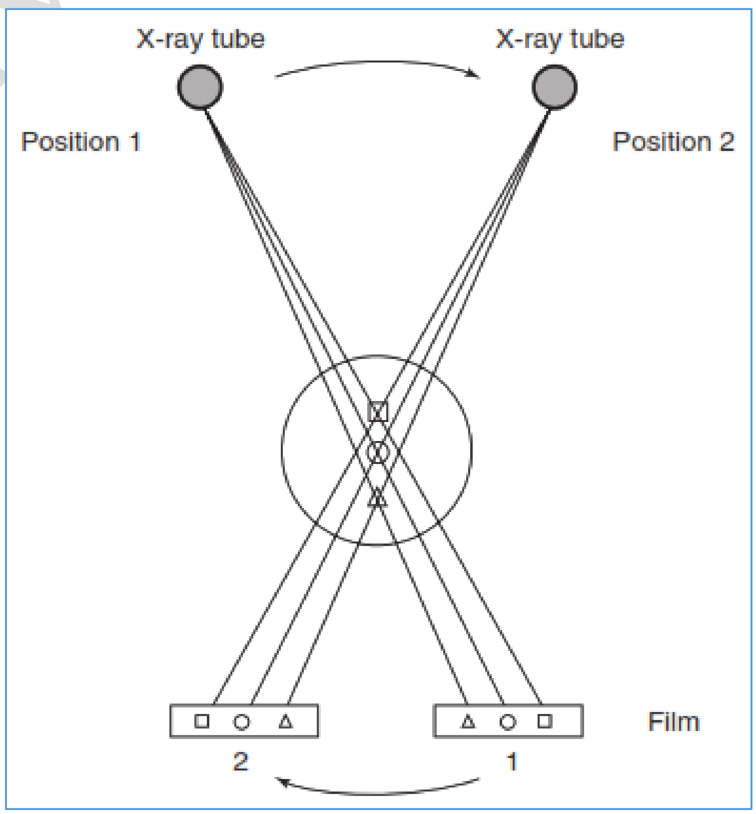

Figure 3: Basic principles of conventional tomography. The x-ray tube and film move simultaneously and in opposite directions to ensure that the desired section(○) of the patient is imaged by blurring out structures above (■) and below(Δ) the plane of interest (○).

The limitations of radiography and tomography result in the inability of film to image very small differences in tissue contrast. In addition, contrast cannot be adjusted after it has been recorded on the film. Digital imaging modalities such as CT, for example, can alter the contrast to suit the needs of the human observer (radiologists and technologists) by use of various digital **image postprocessing** techniques.

CT scans can be performed on every region of the body for a variety of reasons (e.g., diagnostic, treatment planning, interventional, or screening). The cross-sectional images generated during a CT scan can be reformatted in multiple planes, and can even generate three-dimensional images which can be viewed on a computer monitor, printed on film or transferred to electronic media. Although most common in medicine, CT is also used in other fields, such as nondestructive materials testing, to study biological and paleontological specimens.

CT differs from the conventional radiography in two significant ways:

- First, CT forms across-sectional image, eliminating the superimposition of structures that occurs in plane film imaging because of compression of 3D body structures onto the two-dimensional recording system.

- Second, the sensitivity of CT to subtle differences in x-ray attenuation is at least a factor of 10 higher than normally achieved by screen-film recording systems because of the virtual elimination of scatter.

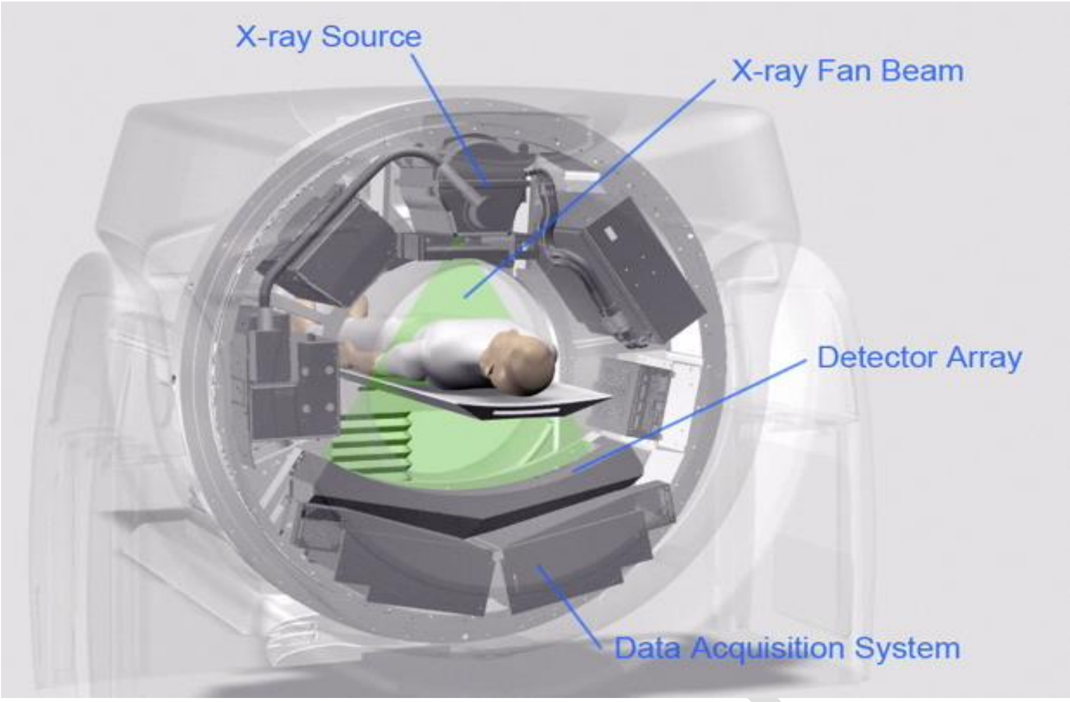

Figure (5) : Schematic of a CT system

Purpose of CT scan

CT-scans provide detailed cross-sectional images of various internal structures, for example, internal organs, blood vessels, bones, soft tissue etc., and can be used for:

- Diagnostic purposes-
- Guidance for specific treatment or further tests- surgeries, biopsies and radiation therapy
- Detection and monitoring of conditions- Cancer, heart disease, lung nodules, liver masses.

Technique

Digital geometry processing is used to generate a three-dimensional image of the inside of an object from a large series of two-dimensional X-ray images taken around a single axis of rotation.

- In the circular opening a flat “patient couch (table)” is mounted, the diameter measures between 24-28 inches. The patient lies flat onto the table and can be adjusted upwards, downward, frontwards or backwards to position for imaging.

- The table moves the patient into the gantry and the x-ray tube rotates around the patient. The scanner gantry contains the rotating portion that holds the X-ray tube generator and the detector array. As x-rays pass through the patient to the detectors, A computer system acquires and performing the necessary calculations to go from measurements to a viewable image.

- One cross sectional slice of the body is obtained for each complete rotation. Multiple shots are taken as the scanner rotates and these are called “profiles”. Within one rotation about 1,000 profiles are acquired. A two dimensional image (slice) is formed when a full set of profiles from each rotation that are analyzed by a computer are compiled.

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Computed Tomography EquipmentsTechniques

Second stage/ 2nd coarse المرحلة: الثانية / الكورس الثاني

Title: العنوان:

- **Basic principles of CT Scanners : Generations of CT**
  - **First-generation**
  - **Second-generation**
  - **Third-generation**
  - **Fourth-generation**
  - **Fifth-generation CT , electron beam (EBCT)**

Name of the instructor: اسم المحاضر:

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Target population: الفئة المستهدفة:

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Introduction: المقدمة:

Generations of CT scanners

CT scanners were first introduced in 1971 with a single detector for brain study under the leadership of Godfrey Hounsfield, an electrical engineer at EMI (Electric and Musical Industries Ltd). Thereafter, it has undergone several changes with an increase in the number of detectors and decrease in the scan time. The changes were majorly on the X-ray tube and detector arrangements.

Scientific Content: المحتوى العلمي:

A Major adjustment in the technology of CT scanners was manifested in:

- Tube orientation and shape of beam (from pencil beam through narrow beam to fan beam)
- Number of detectors (from single detectors to multiple detectors).
- Detector arrangement.

►*First Generation: (Parallel-Beam Geometry)*

First-generation CT systems are characterized by a single X-ray source (pencil beam or parallel-beam geometry). Multiple measurements of x-ray transmission are obtained using a single highly single collimated x-ray pencil beam and detector directing across the the patient isocenter. Both, the source and the detector, translate simultaneously in a scan plane, where the beam is translated in a linear motion across the patient to obtain a projection profile.

This process (translate – rotate scanning motion) is repeated for a given number of angular rotations, by approximately 1 degree, and another projection profile is obtained, until the source and detector have been rotated by 180 degrees. The advantages of this design are simplicity, good view-to-view detector matching, flexibility in the choice of scan parameters (such as resolution and contrast), and the highly collimated beam provides excellent rejection of radiation scattered in the patient.

This scanner was limited because;

1. Only head scans could be performed.
2. Generates a lot of heat, therefore, require an elaborate cooling system.
3. Scan time was very slow. About 1 minute per slice therefore the duration of scan (average): 25-30 mins.

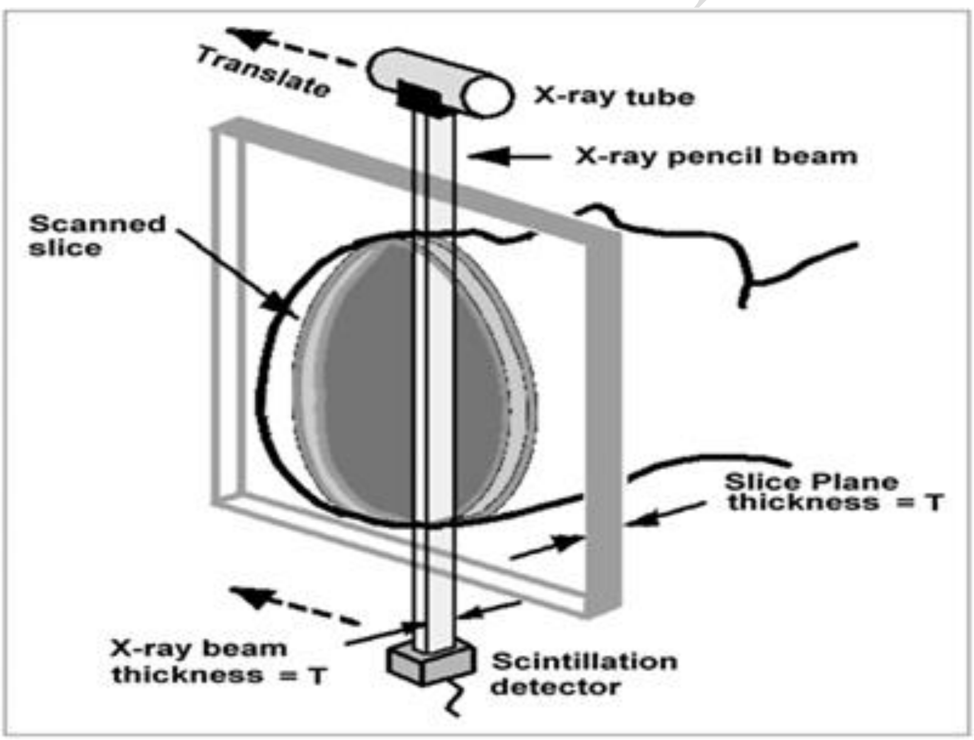

Figure (1) : First Generation: Parallel-Beam Geometry.

►*Second Generation: (Fan Beam, Multiple Detectors)*

Second-generation CT systems use the same translate/rotate scan geometry as the first generation. The difference here is that a pencil beam is replaced by a fan beam and a single detector by multiple detectors (5-30) so that, a series of views can be acquired during each translation, which leads to correspondingly shorter scanning times, about 20 seconds per slice therefore duration of scan (average): less than 90 sec. So, objects of wide range sizes can be easily scanned with the second-generation scanners. The reconstruction algorithms are slightly more complicated than those for first-generation algorithms because they must handle fan-beam projection data.

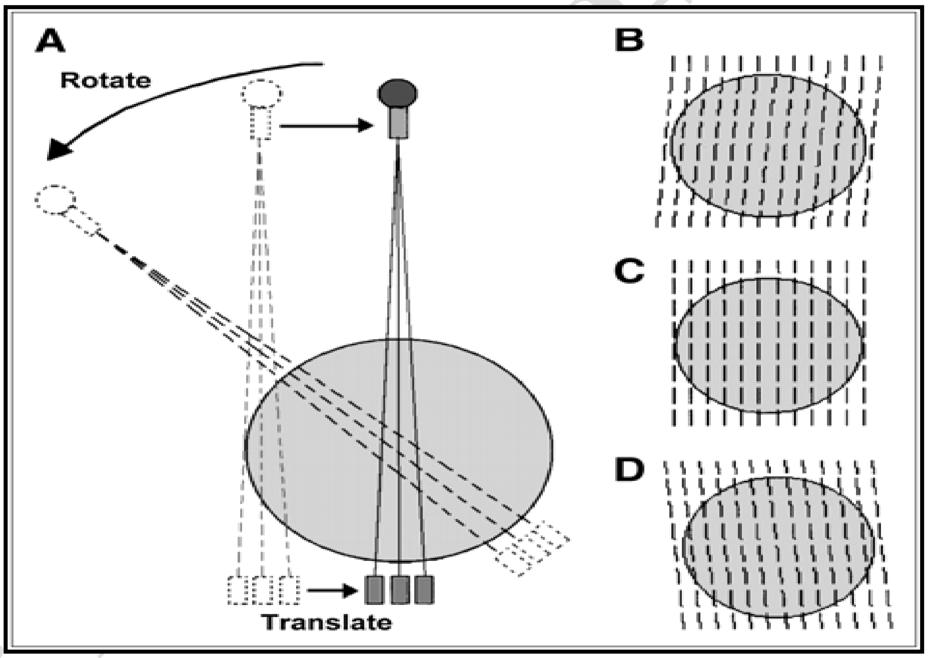

Figure (2) : Second Generation: Fan Beam, Multiple Detectors.

► *Third Generation: (Fan Beam, Rotating Detectors)*

A fan beam of x-rays is rotated 360 degrees around the isocenter. No translation motion is used; however, the fan beam must be wide enough to completely contain the patient. A curved detector array consisting of several hundred independent detectors (500-1000) is mechanically coupled to the x-ray source, and both rotate together. As a result, these rotate-only motions acquire projection data for a single image in as little as 1 s.

Typically, third generation systems are faster than second-generation systems. The detectors here have incorporated bigger amount of sensors in the detector array.

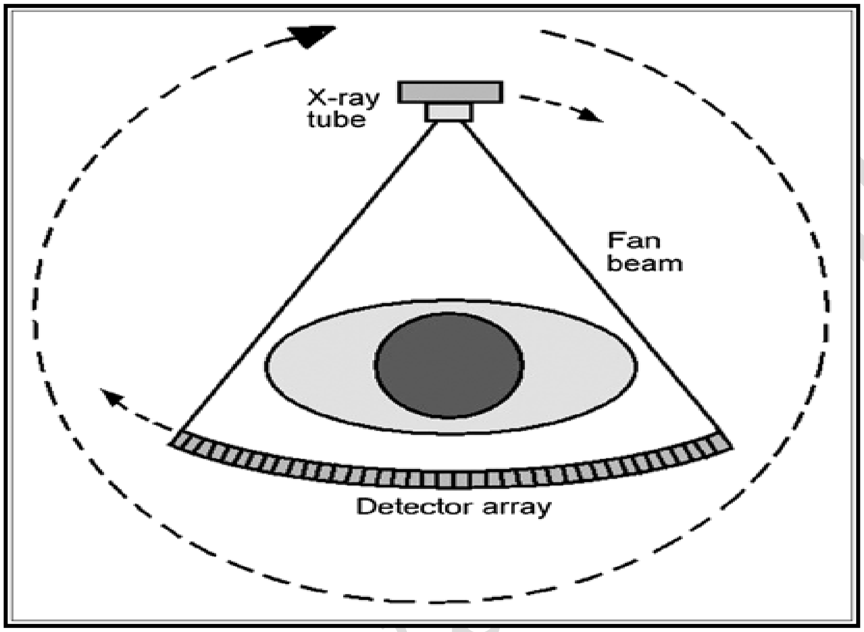

Figure (3) : Third Generation: Fan Beam, Rotating Detectors

▶ ***Fourth Generation: (Fan Beam, Fixed Detectors)***

In a fourth-generation scanner, the x-ray source and fan beam rotate about the isocenter, while the detector array remains stationary. The detector array consists of 600 to 4800 (depending on the manufacturer) independent detectors in a circle that completely surrounds the patient. Scan times are less to those of third-generation scanners (~ 2sec.).The number of views is equal to the number of detectors.

Two detector geometries are currently used for fourth-generation systems:

(1) a rotating x-ray source inside a fixed detector array and

(2) a rotating x-ray source outside a nutating detector array

Both third- and fourth-generation systems are commercially available with advanced configurations.

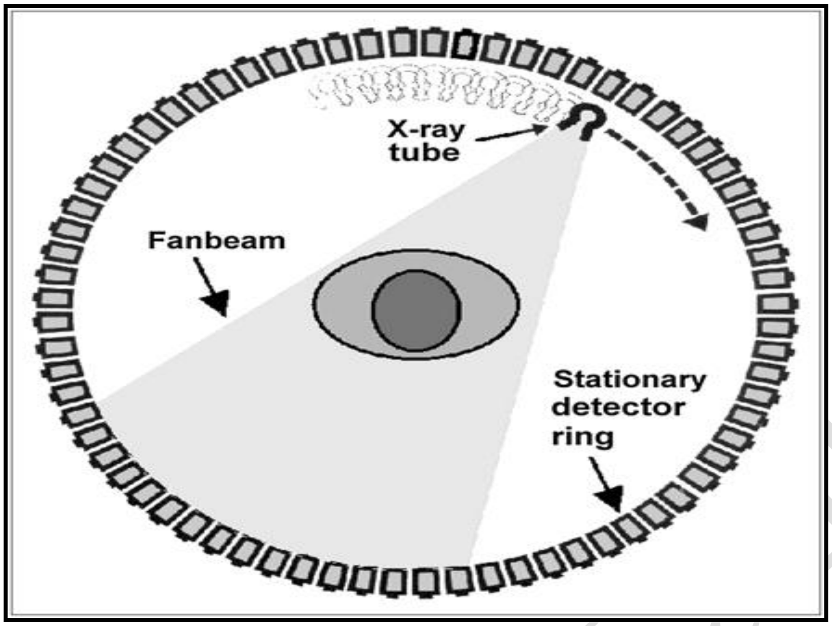

Figure (4): Fourth Generation: Fan Beam, Fixed Detectors.

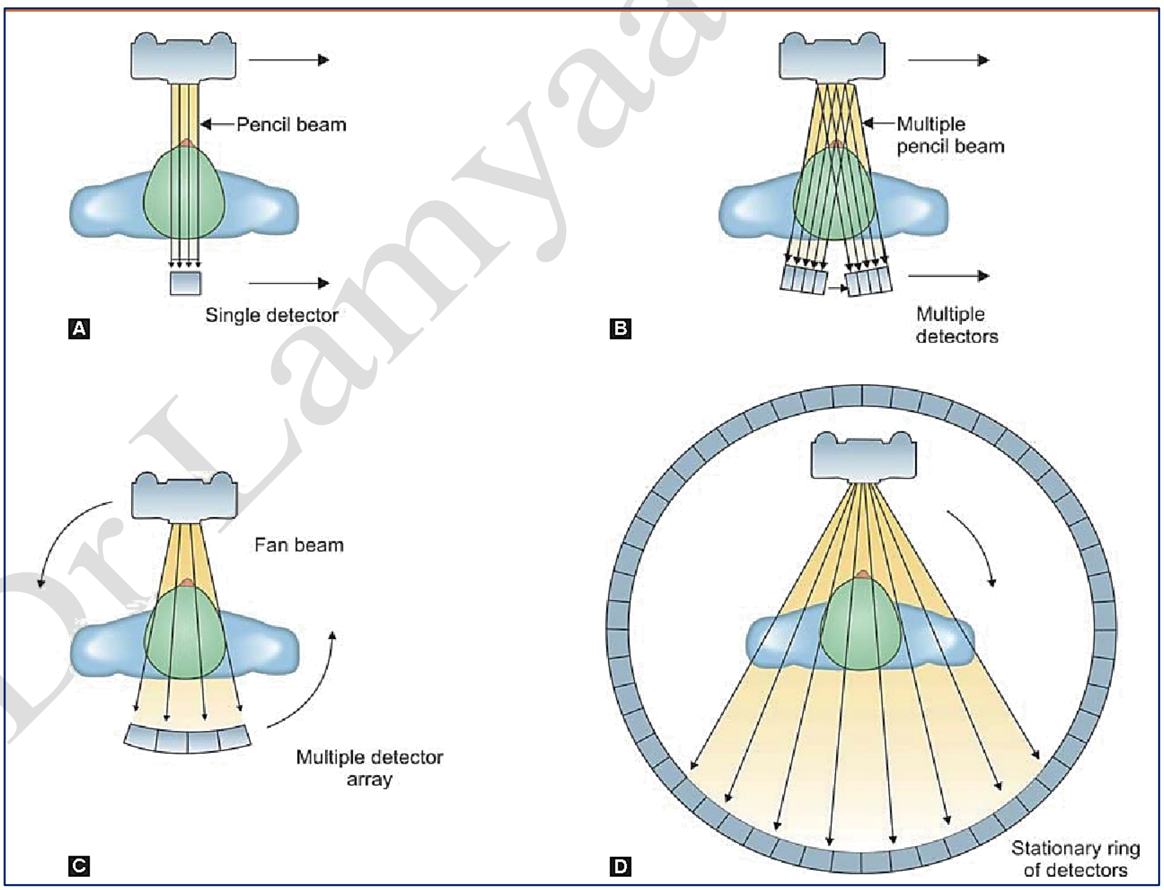

Figure (5): The Four Generations of CT scan

▶ ***Fifth Generation: (Electron beam scanning EBSCT)***

Fifth-generation scanners are unique in that the x-ray source becomes an integral part of the system design. The detector array remains stationary, while a high – energy electron beams is electronically swept along a semicircular tungsten strip anode. X-rays are produced at the point where the electron beam hits the anode, resulting in a collimated fan beam x-rays that rotates about the patient with no moving parts.

Projections data can be acquired in approximately (<50ms) and performing complete scans in a little as 10-20ms, which is fast enough to image the beating heart without significant motion artifacts. So, it was designed for ultrafast scans to freeze cardiac motion in Cardiac CT scans, where was a hurdle with previous existed generation.

The idea behind the ultrafast scanner is the large bell shaped x- ray tube. It doesn't use conventional x-ray tube, instead, a large arc of tungsten encircles the patient and lies directly opposite to the detector ring. X-rays are produced from a focal track as a high energy electron beam strikes the tungsten. The concept is known as EBCT (Electron Beam CT).

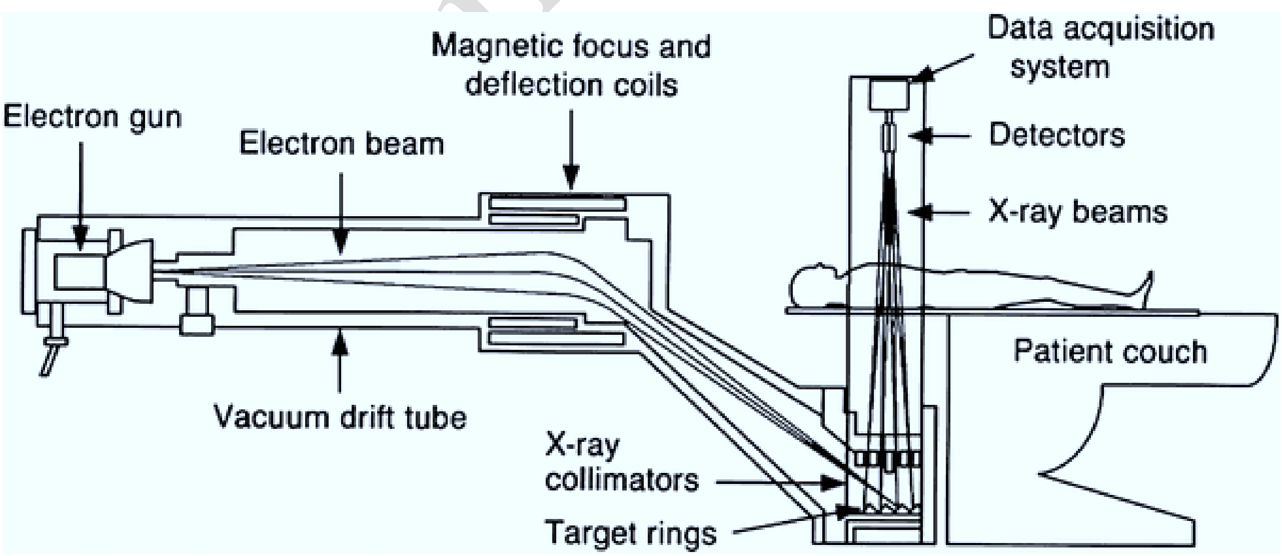

Fig. (6) : Electron beam scanning EBSCT

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Computed Tomography EquipmentsTechniques

Second stage/ 2nd coarse **المرحلة: الثانية / الكورس الثاني**

Title: العنوان:

- **Helical/spiral CT Scanners: Requirements for Volume Scanning:**
  - **Slip-ring technology**

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Target population: الفئة المستهدفة:

Students of second class طلبة المرحلة الثانية

▶ *Sixth Generation (Helical or spiral CT)*

In conventional CT (the 3rd and 4th generation CT scanners), the patient was scanned one slice at a time. The x-ray tube and detectors rotate for 360 degrees or less to scan one slice while the table and patient remain stationary. This slice-by-slice scanning is time-consuming. On the other hand, cables are spooled onto a drum, released during rotation and respooled during reversal. Scanning, braking and reversal required at least 8-10 sec of which only 1-2 sec were spent for data acquisition. The result was a poor temporal resolution and long procedure time.

Therefore, efforts were made to increase the scanning of larger volumes in less time. This notion led to the development of a technique in which a volume of tissue is scanned by moving the patient continuously through the gantry of the scanner while the x-ray tube and detectors rotate continuously for several rotations. As a result, the x-ray beam traces a path around the patient.

The development of helical or spiral CT was a truly revolutionary advancement in CT scanning that finally allowed true 3D image acquisition within a single breath hold technique. For more clarification, when the examination begins, the x-ray tube rotates continuously while the couch moves the patient through the plane of the rotating x-ray beam, (the table smoothly moves through the rotating gantry). This means that the X-ray tube and detector perform a ‘spiral’ or ‘helical’ movement with respect to the patient, generally at a rate of one revolution per second.

In this technique the data are continuously acquired or collected without pausing while the patient is simultaneously transported at a constant speed through the gantry. For this reason the duty cycle of the helical scan is improved to nearly 100% and the volume coverage speed performance can be substantially improved. This technique allows fast and continuous acquisition of the data from a complete volume.

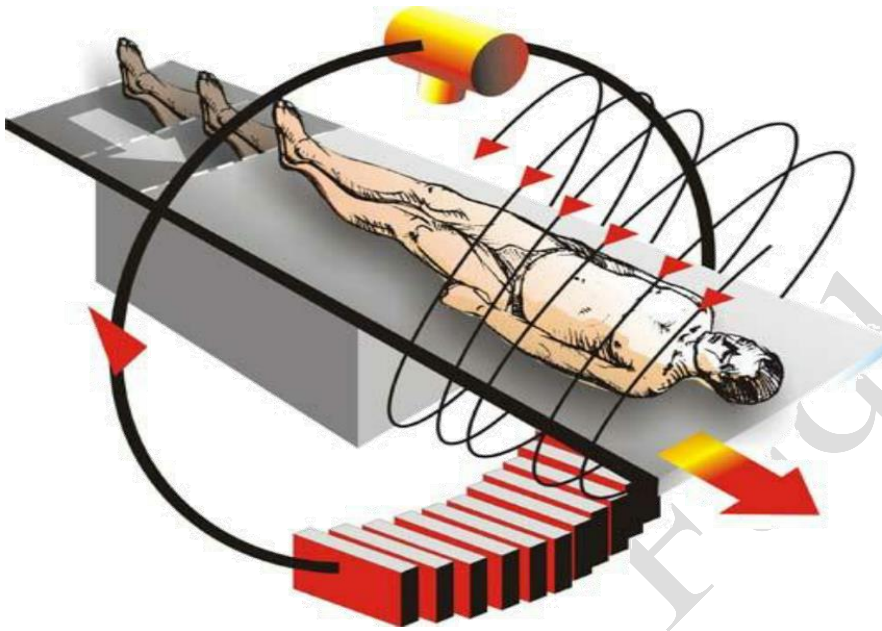

Fig. (1): Helical CT

<u>Three technological developments were required:</u>

- Slip ring technology
- high power x-ray tubes
- Interpolation algorithms

<u>Slip ring technology</u>

All generations of CT scanners (except 4^{th} gen.) required winding and unwinding of connection cables causing inter-scan delays. Slip ring was designed to eliminate this. A **slip ring** is a drum with grooves along which electrical contactor brushes slide. Data are transmitted from detectors via various high capacity wireless technologies, thus allowing continuous rotation. Eliminating interscan delays made possible by slip ring technique. A slip ring passes electrical power to the rotating components without fixed connections.

It allows the complete elimination of interscan delays except for the time required to move the table to next slice position. For eg : if scanning and moving the table each take 1s, only 50% of the time is spent acquiring the data.

Slip rings are electromechanical devices consisting of circular electrical conductive rings and brushes that transmit electrical energy across a moving interface. All power and control signals from the stationary parts of the scanner system are communicated to the rotating frame through the slip ring.

The slip-ring design consists of sets of parallel conductive rings concentric to the gantry axis that connect to the tube, detectors, and control circuits by sliding contactors (Fig. 2, 3). These sliding contactors allow the scan frame to rotate continuously with no need to stop between rotations to rewind system cables. This engineering advancement resulted initially from a desire to reduce interscan delay and improve throughput.

However, reduced interscan delay increased the thermal demands on the x-ray tube; hence, tubes with much higher thermal capacities were required to withstand continuous operation over multiple rotations.

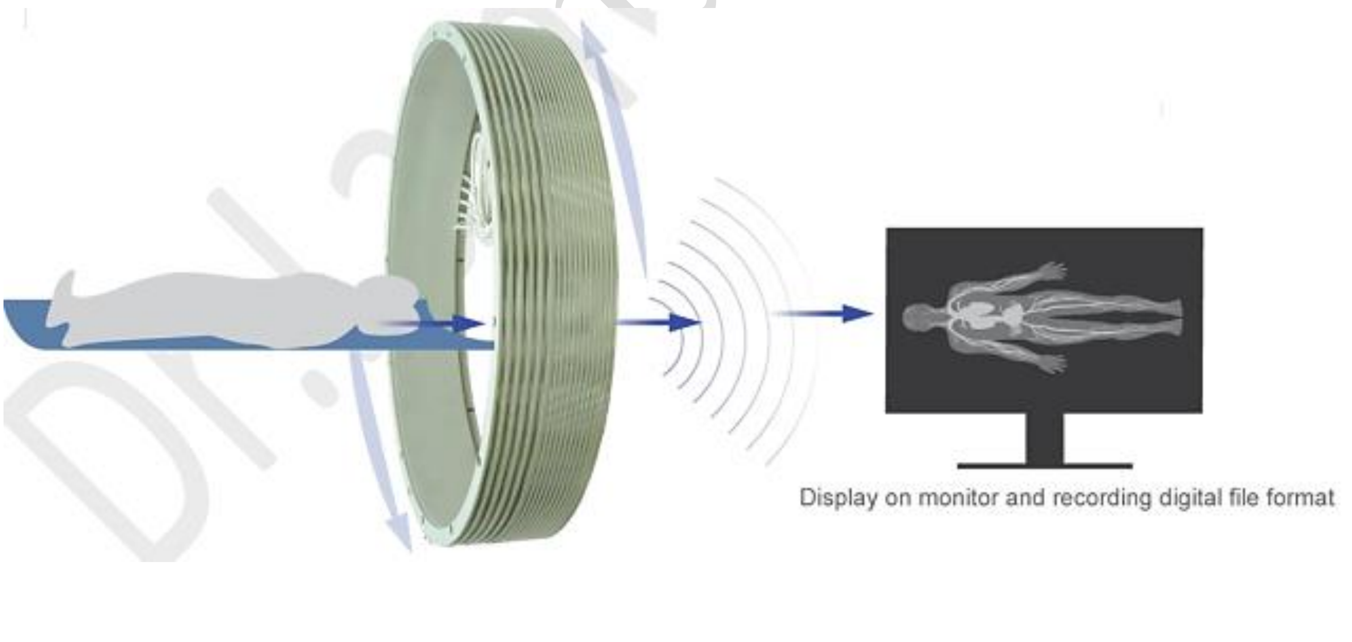

Fig. (2): Slip ring technology

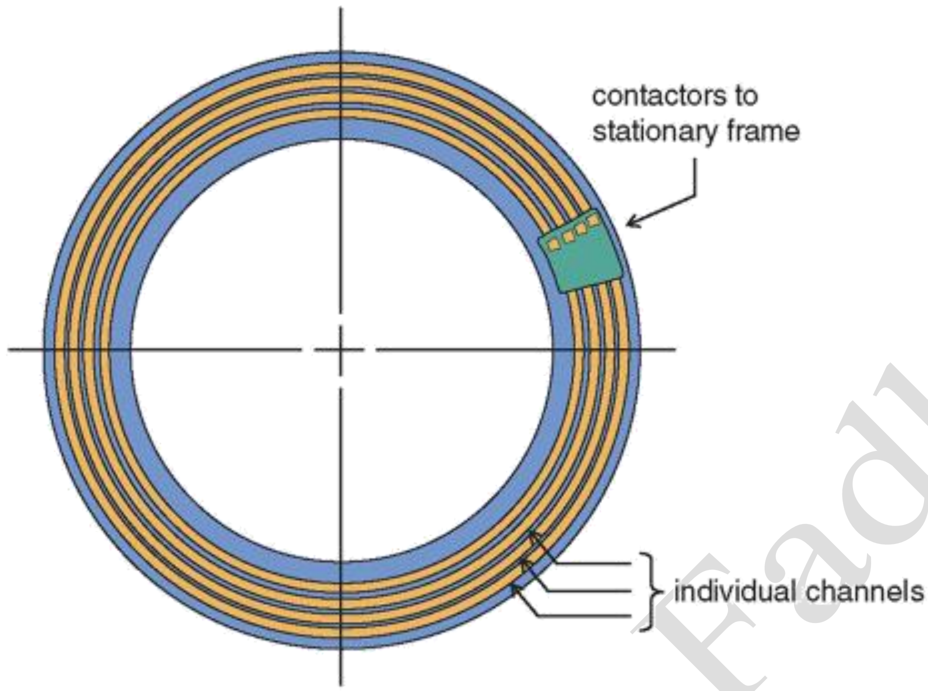

Fig. (3): To convey power onto the rotating gantry from the stationary frame, as well as to conduct signal data from the rotating gantry to the stationary frame, a slipring is used. A slipring uses gliding contacts to allow communication and power transfer between the stationary and rotating frames without the use of wires, and this enables the gantry to rotate continuously in a single direction. Sliprings have enabled gantry rotation periods to move from 3.0 s (when cables were used) to modern CT rotation periods of as little as 0.25 s.

High power x-ray tubes

X-ray tubes are subjected to far higher thermal loads in CT than in any other diagnostic x-ray application. In early CT scanners, the power level was low. Since long scan times allowed heat dissipation. Shorter scan times in later versions of CT scanners required high-power x-ray tubes and use of oil cooled rotating anodes for efficient thermal dissipation.

The introduction of helical CT with continuous scanner rotation placed new demands on x-ray tubes. Several technical advances in component design have been made to achieve these power levels and deal with the problems of the target temperature, heat storage, and heat dissipation. For example, the tube envelope, cathode assembly, and anode assemblies including anode rotation and target design have been redesigned.

As scan times have decreased, anode heat capacities have increased by as much as a factor of five, preventing the need for cooling delays during most clinical procedures, and tubes with capacities of 5–8 million heat units are available. In addition, improvement

in the heat dissipation rate (kilo–heat units per minute) has increased the heat storage capacity of modern x-ray tubes.

The large heat capacities are achieved with thick graphite backing of target disks, anode diameters of 200 mm or more, improved high-temperature rotor bearings, and metal housings with ceramic insulators (Fig. 4).

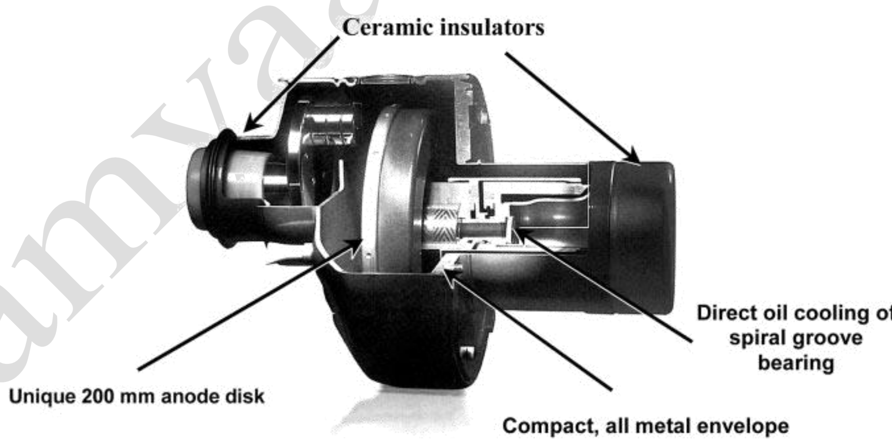

Fig. (4): CT x-ray tube.

Among other factors. The working life of tubes used to date ranges from 10,000 to 40,000 hours, compared with the 1,000 hours typical of conventional CT tubes. Because many of the engineering changes increased the mass of the tube, much of the design effort was also dedicated to reducing the mass to better withstand increasing gantry rotational rates required by ever faster scan times.

In summary:

Shorter scan time required high power of x-ray tubes and use of oil cooled rotating anodes for efficient thermal dissipation. Largest heat capacities are achieved with:

- ◣ Thick graphite backing of target disks
- ◣ Anode diameters of 200mm or more
- ◣ Metal housing with ceramic insulator.
- ◣ The working life of tubes ranges from 10,000 – 40,000 hours

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Second stage/ 2nd coarse المرحلة: الثانية / الكورس الثاني

Title: العنوان:

- **Interpolation Algorithms**
- **Pitch**

Name of the instructor: اسم المحاضر:

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Target population: الفئة المستهدفة:

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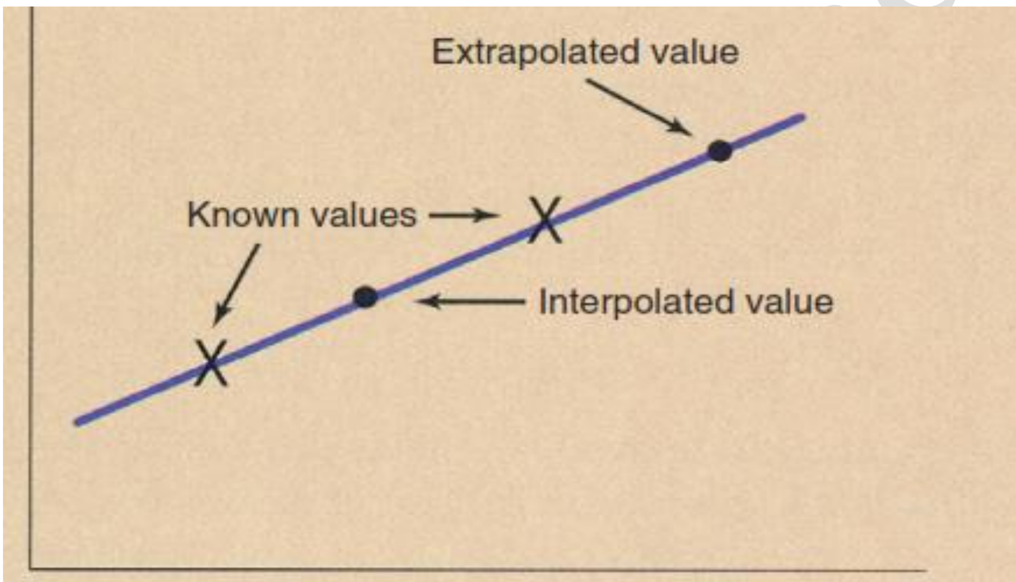
Scientific Content: :المحتوى العلمي

Interpolation Algorithms

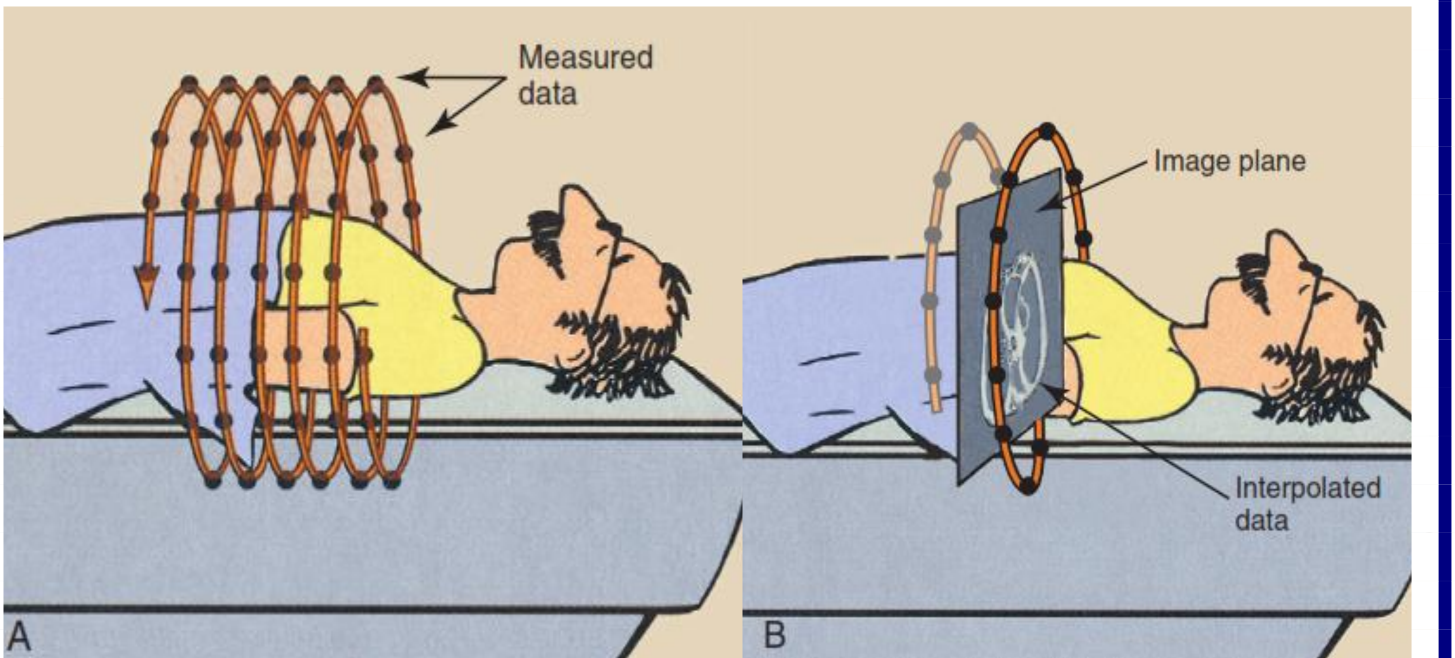
Reconstruction of an image at any z-axis position is possible because of a mathematical process called **interpolation**. Figure (1) presents a graphic representation of **interpolation** and **extrapolation**. If one wishes to estimate a value between known values, that is an interpolation; if one wishes to estimate a value beyond the range of known values, that is an extrapolation.



Fig(1): Interpolation estimates a value between two known values. Extrapolation estimates a value beyond known values.

During helical CT, image data are received continuously, as shown by the data points in Figure 2*A*. When an image is reconstructed, as in Figure 2,*B*, the plane of the image does not contain enough data for reconstruction.

The data in that plane must be estimated by interpolation. Data interpolation is performed by a special computer program called an ***interpolation algorithm***.

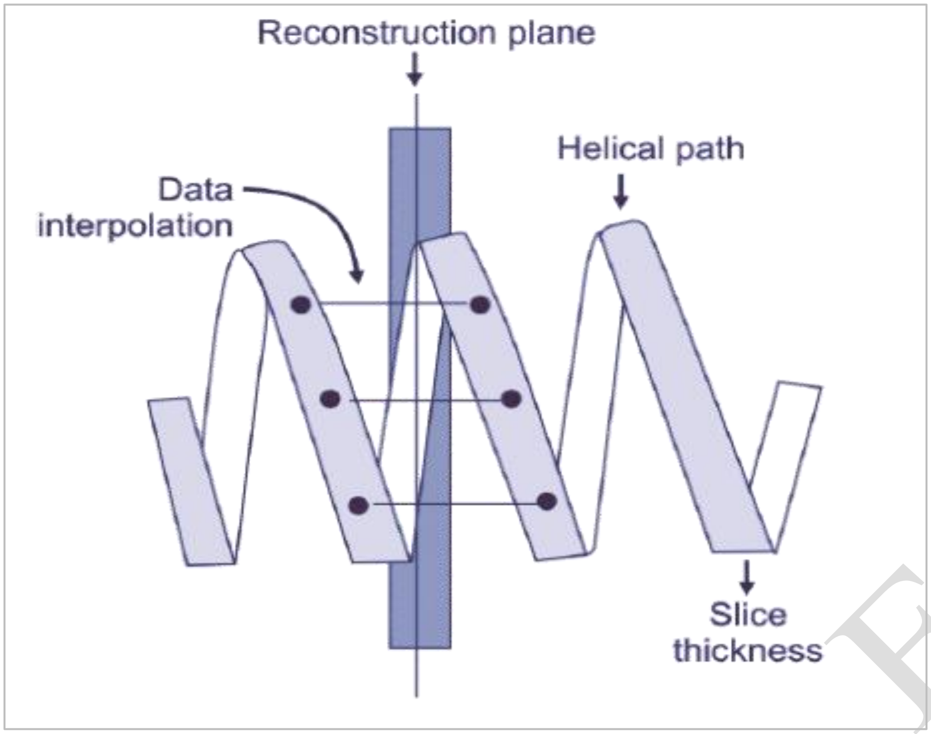


Figs.(2): A, During multislice helical computed tomography, image data are continuously sampled. B, Interpolation of data is performed to reconstruct the image in any transverse plane.

Image interpolation creates a number of new slices between known slices in order to obtain an isotropic volume image.

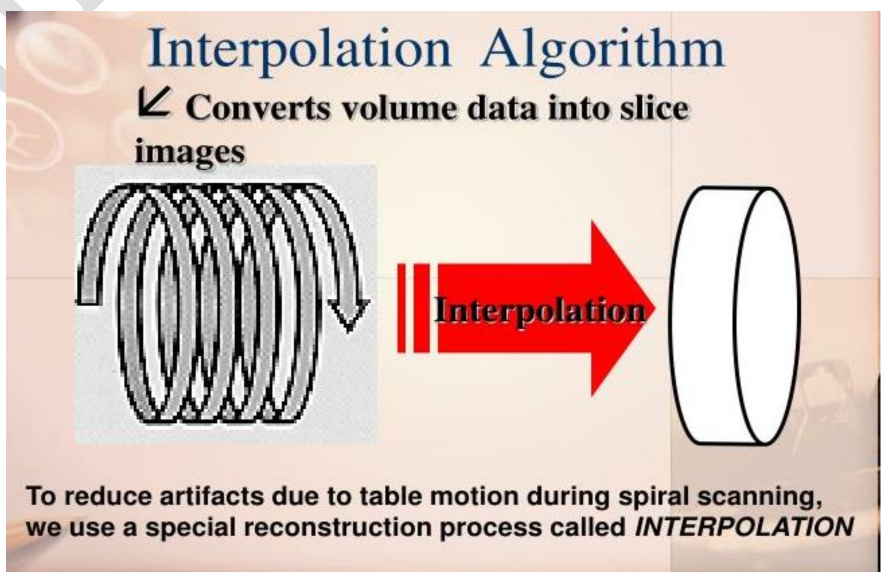
The problem with continuous tube and table motion was that projections precessed in a helical motion around the patient and did not lie in a single plane. This meant that conventional reconstruction algorithms could not work.

Helical CT scanning produces a data set in which the x-ray source has travelled in helical trajectory around the patient, (the data are acquired in a helical path around the patient). Present day CT reconstruction algorithms assume that the x-ray source has negotiated a circular not a helical path around the patient. To compensate for these differences in the acquisition geometry, before the actual CT reconstruction the helical data set is interpolated into a series of planar image data sets (the reconstruction plane of interest). Interpolation is essentially a weighted average of the data from either side of the reconstruction plane, with slightly different weighting factors used for each projection angle.



Figs.(3): Data interpolation

In summary: *Interpolation Algorithms are the mathematical process required to reconstruct axial images from the spiral volume data set.*



Pitch

During helical scans, the table motion causes displacement of the fan beam projections along the z axis; the relative displacement is a function of the table speed and the beam width. The ratio of table displacement per 360° rotation to section thickness is termed *pitch*.

Pitch is the table movement per rotation divided by beam width.

pitch = table travel / beam width

- pitch = 1 - coils of the helix are in contact
- pitch < 1 - coils of the helix overlap
- pitch > 1 - coils of the helix are separated

For example

↳ If beam width is 10cm, the table moves 10cm during one tube rotation, then pitch is 1, so, x-ray beam associated with consecutive helical loops are contiguous.

↳ If beam width is 10cm and table moves 15cm per tube rotation, then pitch is 1.5

So, a gap exists between the x-ray beam edge of consecutive loop.

↳ If beam width is 10cm and table moves 7.5cm then pitch is 0.75, so, beams and consecutive loops overlap by 2.5 (doubly irradiating the underlying tissues).

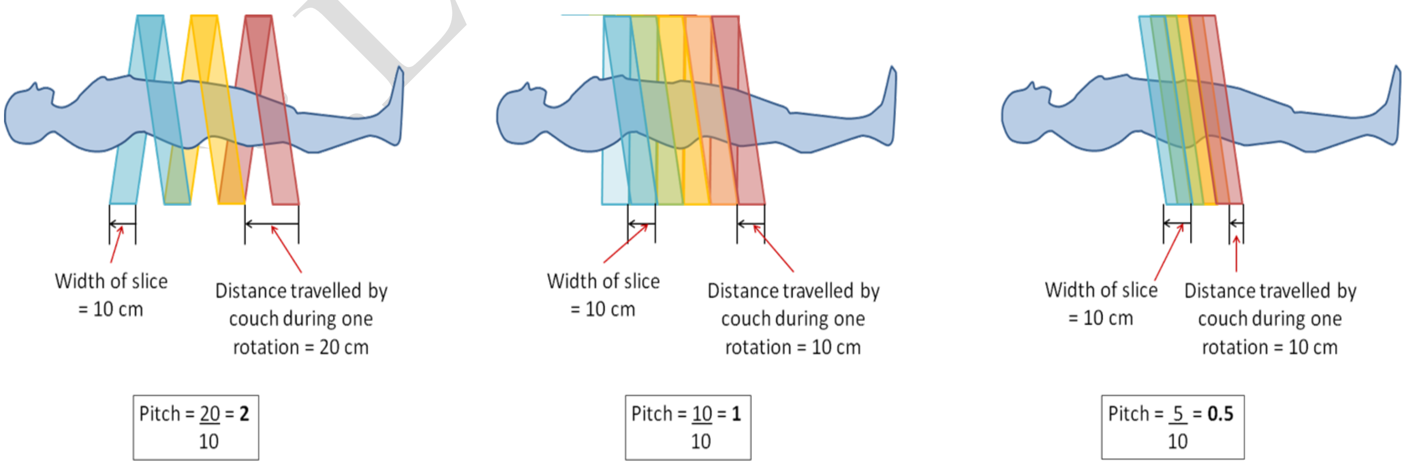


Fig (4): Illustration of pitch concepts

The relationship between the volume of tissue imaged and pitch is given as follows:

VOLUME IMAGING

$$\text{Tissue imaged} = \frac{\text{Beam width} \times \text{Pitch} \times \text{Imaging time}}{\text{Gantry rotation time}}$$

<u>Advantages of helical CT scanner</u>

- ✓ Fast scan times and large volume of data collected.
- ✓ Minimizes motion artifacts.
- ✓ Less mis-registration between consecutive slices.
- ✓ Reduced patient dose.
- ✓ Improved spatial resolution.
- ✓ Enhanced multiplaner or 3D renderings.
- ✓ Improved temporal resolution

Middle Technical University (MTU) الجامعة التقنية الوسطى

College of Health and Medical Techniques -Baghdad كلية التقنيات الصحية والطبية/ بغداد

Radiological Techniques Department قسم تقنيات الأشعة

المادة: تقنيات اجهزة التصويرالمقطعي المحوسب

Computed Tomography EquipmentsTechniques

Second stage/ 2nd coarse المرحلة: الثانية / الكورس الثاني

Title: العنوان:

- **Multislice Computed Tomography (MSCT) (multidetector-row) CT**
- **Dual sources CT**

Name of the instructor: اسم المحاضر:

م.د.لمياء فاضل عبدالحسين

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Target population: الفئة المستهدفة:

Students of second class طلبة المرحلة الثانية

▶ *Seventh Generation (MS/MD CT)*

The multislice CT (MSCT), or multi-detectorrow CT (MDCT), is a CT system equipped with multiple rows of CT detectors to create images of multiple sections. This CT system has different characteristics from conventional CT systems, which have only one row of CT detectors.

The introduction of this advanced detector system and its combination with helical scanning has markedly improved the performance of CT in terms of imaging range, time for examination, and image resolution. At the same time, the time for scanning (the time required for 1 revolution) has been shortened to 0.5 sec. and the width of the slice (tomographic plane) reduced to 0.5 mm. Thus, dramatic improvements have been made in CT-based diagnostic techniques.

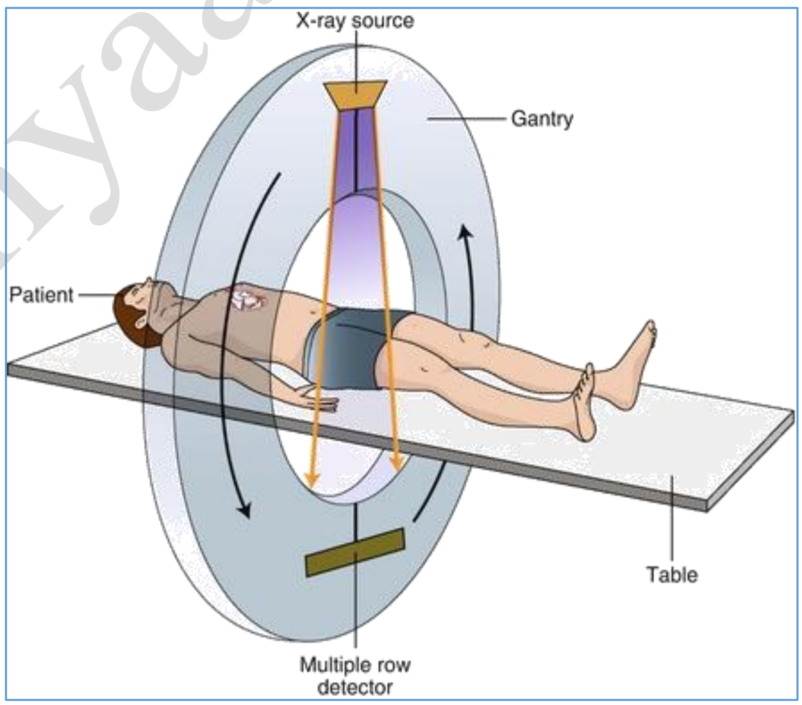


Figure (1): A cartoon depiction of a typical MDCT.

The primary difference between MSCT and SSCT is the detector arrangement (Fig.2). ***SSCT*** uses a one dimensional detector arrangement where many individual detector elements are arranged in a single row across the irradiated slice that receives the x ray signals. ***In MDCT***, there are multiple rows of detectors. By increasing the number of detector rows, the z axis coverage slab thickness increases, thereby decreasing the number of gantry rotations necessary to image the selected field of view (scan length), so reducing the strain on the x ray tube.

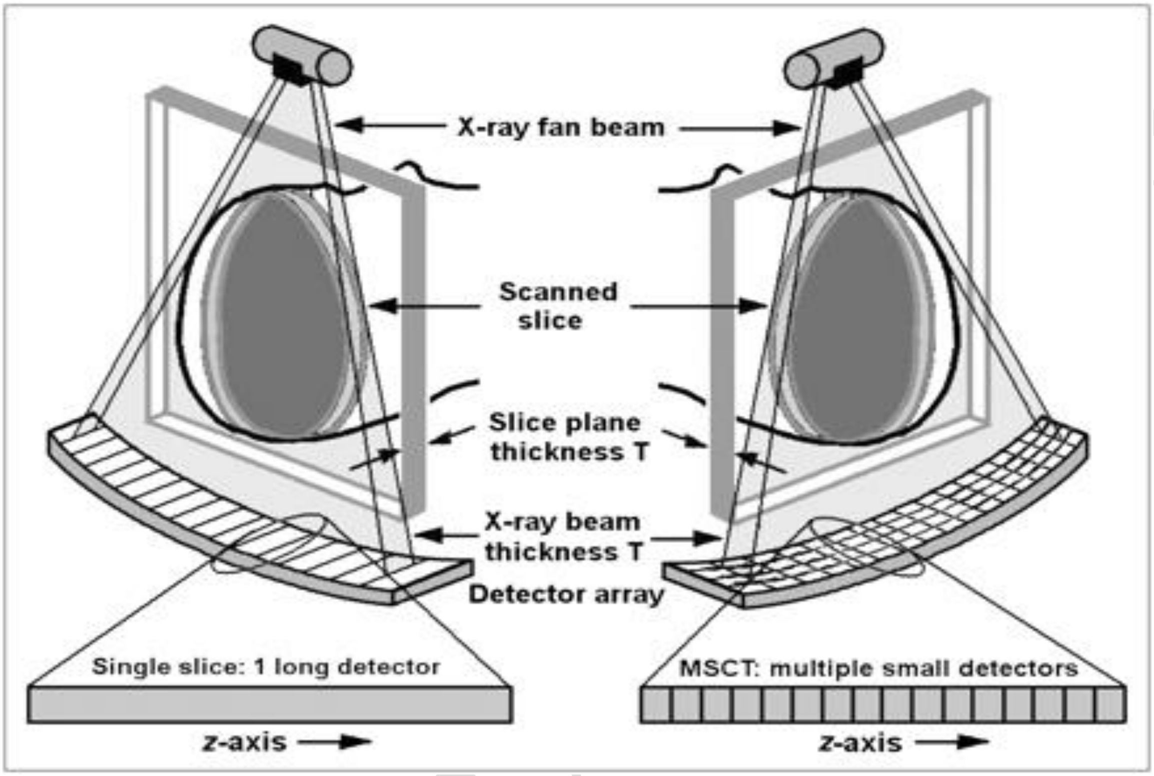


Figure. 2. A cartoon depicting a single slice scanner and a multislice scanner

For example, if each detector was 1.25 mm long and the scanner had 16 rows of detectors, the z axis coverage (slab thickness) per gantry rotation would total 20 mm. Subsequent MSCT scanners possessed increasing numbers of detector rows starting at 16 rows and moving to 64, 156 and 320 rows. The coverage (slab thickness) varies by detector row number where slab thickness per gantry rotation is directly proportional to detector row number. Figure (3) depicts the concept of slab thickness.

CT scanners of the same detector row number may have different slab thicknesses depending on the z axis size of each individual detector. Smaller detectors will cover less of the z axis per detector row per gantry rotation.

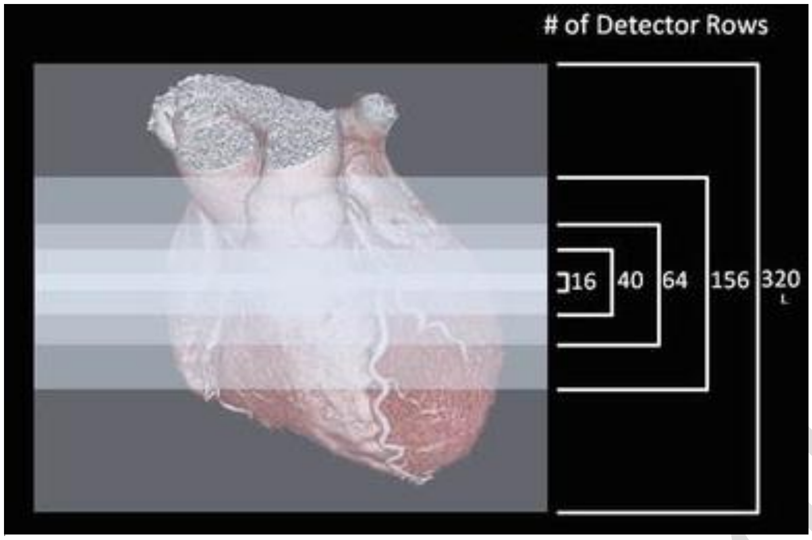


Figure (3): An artificial representation of the meaning of slab thickness or z axis coverage. Z axis coverage is directly proportional to detector row number

The entire detector array consists of groupings, each of which are connected to the mother board unit of the detection system. Each group may be selectively activated or deactivated providing various slice thicknesses which may be predetermined depending on the scan indication. In addition, detector arrays within a given row may be varied. For example, the inner detector rows, which are made up of narrower detectors than the outer rows may be selectively activated such that the slice thickness will narrow (Fig.4). Additionally, pairs of detectors may be linked to create thicker slices.

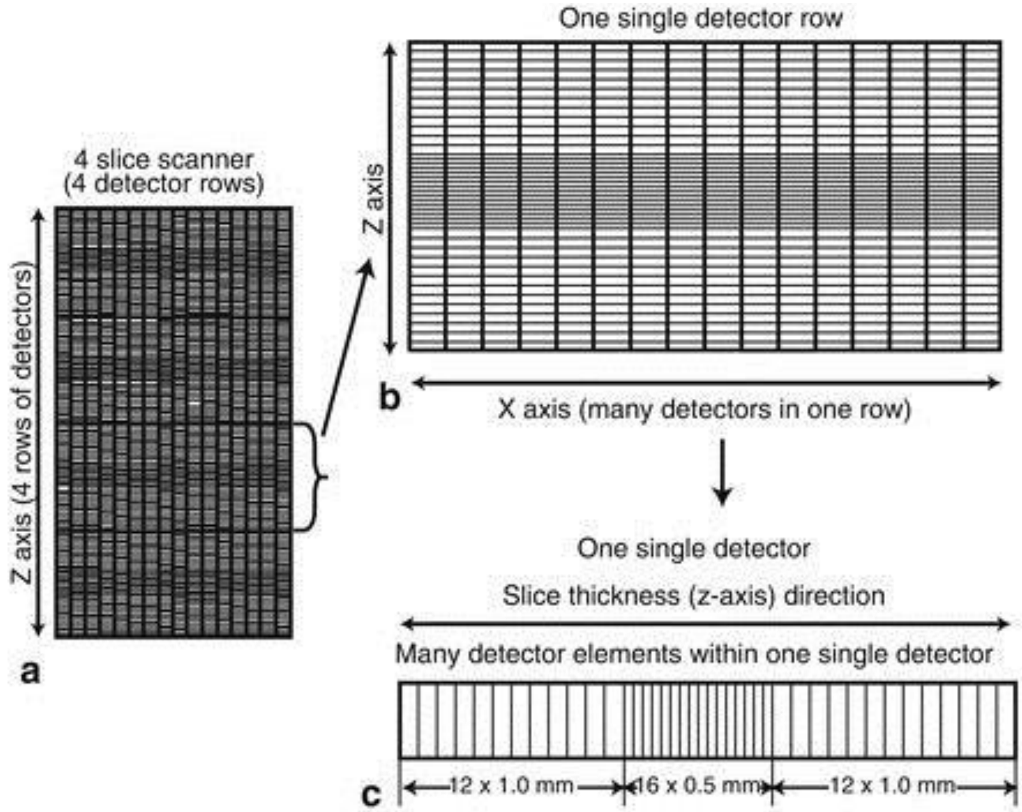


Figure. 4 Cartoon depicting a particular detector array configuration. Panel (**a**) depicts a four slice scanner (four detector rows). Panel (**b**) illustrates one single detector row. Within each row, there are multiple single detectors. There may be as man as 800 detectors per row. Panel (**c**) depicts a single detector within one single detector row. Each single detector has multiple detector elements. This particular detector contains two outer groups of twelve 1 mm detector elements and one inner group of sixteen 0.5 mm detector elements. Elements within a detector can be combined or isolated to create varying slice thicknesses and even submillimeter slice thicknesses necessary for coronary artery imaging.

There are three types of detector arrays (Fig.5):

◣Matrix detectors, which consist of parallel rows of equal thickness, (Philips).

◣ Hybrid detectors with smaller detector rows in the center, (Siemens).

◣ Adaptive array detectors that consist of detector rows with varying thickness. (Detector units with increasing widths toward both ends are arranged symmetrically), (Toshiba).

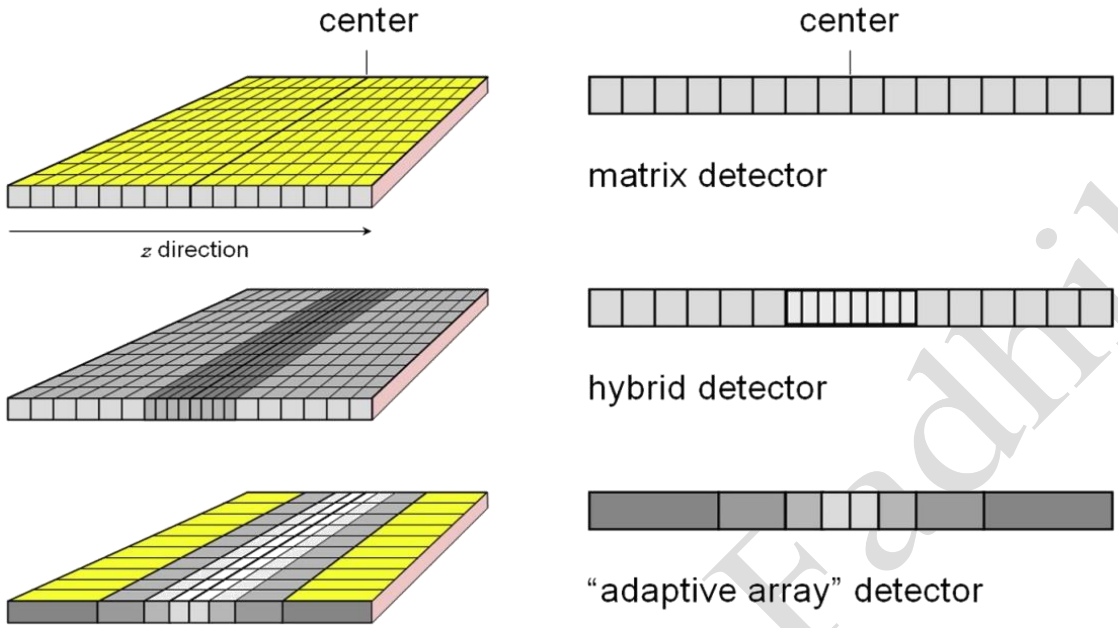


Figure. 5: types of detector arrays in MSCT

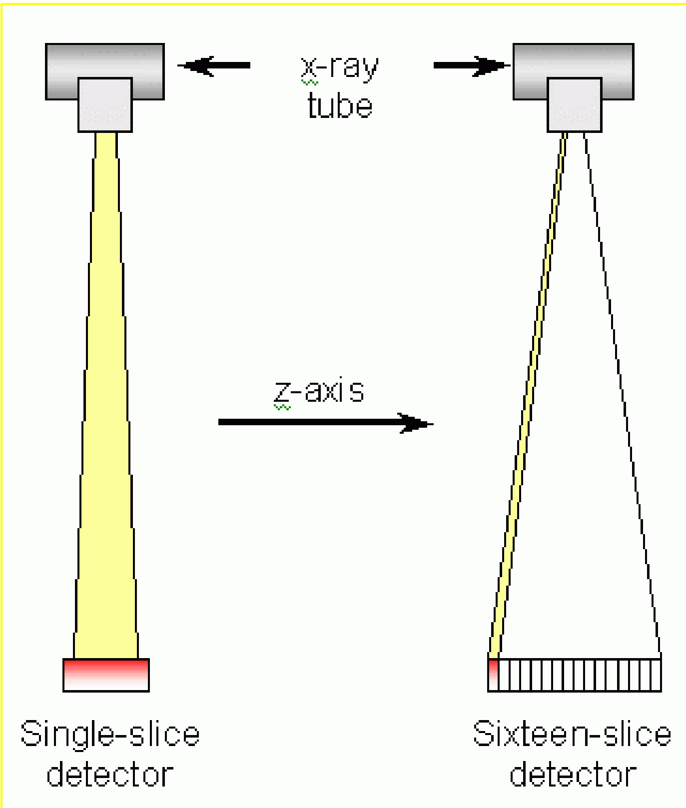
Two significant other differences exist between SSCT and MSCT:

➤ The first involves the relationship between slice thickness and x ray beam width.

In SSCT, X ray beam collimation was designed such that the z axis width of the x ray beam at the isocenter (center of rotation) was the desired slice thickness.

In MSCT, the slice thickness is determined by detector configuration and not x ray beam collimation. Since the detector width or linked detector element width determines the acquired x ray beam thickness (slice thickness), this length is referred to as detector collimation.

➤ The second relates to beam configuration effects. The effects of the dominant cone beam in *MSCT*, in comparison with fan beam shape in *SSCT*, (Figure. 6), are streak artifacts due to the divergent nature of the x ray beam emitted from the patient. This means that the z axis width of the x ray beam is wider when it exits a patient than when it enters. X ray beams 180° apart are sampling the same tissue planes, but their cone-shaped x ray beam sampling is slightly different at 0° than at 180° making the opposite, supposedly identical images, slightly inconsistent. This results in partial volume streaking, which is accentuated with wider x ray beam widths; as such, cone beam artifacts are more pronounced with MSCT than with SSCT. Cone beam artifact severity is directly proportional to the number of detector rows.



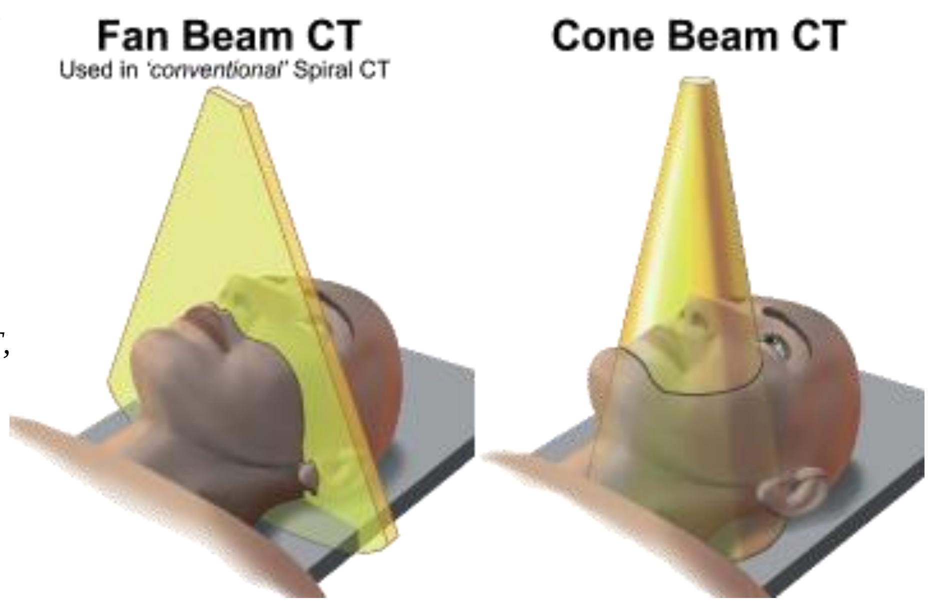


Figure. 6: Cone beam in *MSCT*, in comparison with fan beam shape in *SSCT*.

MS/MD CT has the advantages of:

- Its speed can be used for fast imaging of large volumes of tissue with wide sections. This is particularly useful in studies where patient motion is a limiting factor.
- Their ability to cover large body section in short scan times with thin beams for producing thin, high-detail slice images or 3-D images.
- With conventional single detector array scanners, opening up the collimator increases slice thickness, which is good for utilization of x-ray but reduces spatial resolution in the slice thickness dimension. With the introduction of multiple detector arrays, the slice thickness is determined by the detector size and not by the collimator.
- Overcoming x-ray tube output limitation. One problem quickly encountered with single detector row scanning (SSCT) was excess stress on the x ray tube. That is, the x ray tube would heat to extreme temperatures as very high energy was deposited onto the anode.

Pitch of MS/MD CT

With the introduction of multiple-row detector CT scanner the definition of pitch has changed. Beam pitch needs to be distinguished from detector pitch, which is defined as ***the table rotation per gantry rotation divided by the width of the detector.***

where D is the detector width in millimeters

$$\text{Pitch} = \frac{T}{D}$$

If the x-ray beam is collimated to N active detectors in a multiple-row detector CT scanner, the relationship between beam pitch and collimator pitch is as follows:

$$\text{Beam Pitch} = \frac{\text{Detector Pitch}}{N}$$

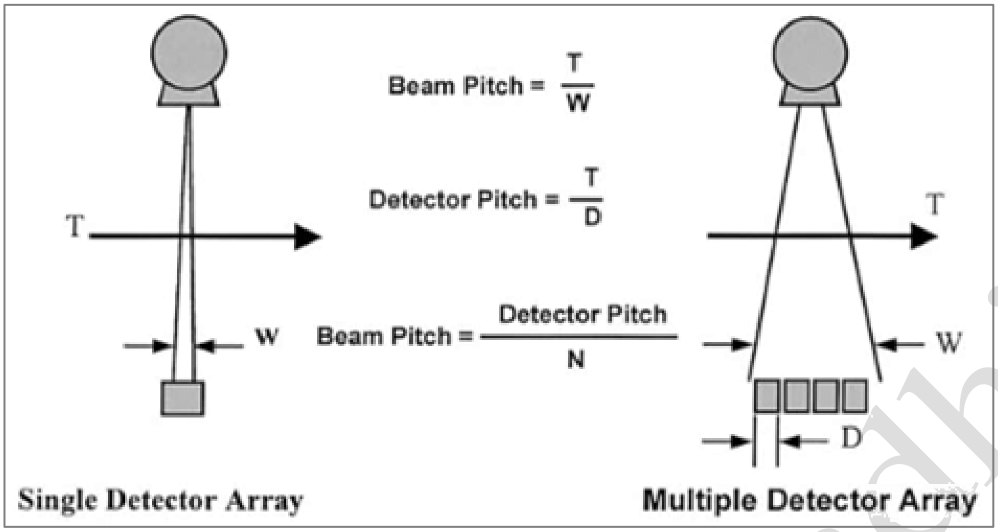


Figure 7. The diagram shows the concepts of beam pitch and detector pitch. Beam pitch is consistent with the previous notion of pitch used in single-row detector helical CT and works well for multiple-row detector CT scanners.

D detector width,

N number of active detectors,

T table travel per gantry rotation,

W beam width.

Comparison of Scanning Time between Single-Row Detector and Multiple-Row Detector CT Scanners

			Scanning Time (sec) by Scanner	
Region Scanned	Distance (cm)	Section Thickness (mm)	Single-Row Detector*	Multiple-Row Detector†
Head	20	8	16.7	2.1
Neck	15	5	20.0	2.5
Chest	30	8	25.0	3.1
Abdomen	20	8	16.7	2.1
Pelvis	20	8	16.7	2.1
All regions	105	. . .	95.1	11.9

*Scanner with 1-second rotation and a pitch of 1.5.
†Scanner with 0.5-second rotation and a pitch equivalent to 1.5.

▶ *Eighth Generation (Dual sources CT)*

The Dual Source CT (DSCT) equipped with two data measurement systems, that it is possible to double the resolution compared with that of a single source CT, and increase the speed of data acquisition.

DSCT include three unique operating modes:

↘ Dual Source mode

Each consisting of one X-ray tube and one corresponding detector array oriented in the gantry with an angular offset of 90 degrees. The two X-ray source/detector systems rotate simultaneously capturing image data in half the time required by conventional technology. With Dual Source CT it is possible to double the resolution compared with that of a single source CT, and increase the speed of acquisition.

Dual Source Single Energy (DSSE)

- In this mode, both X-ray tubes work at the same kVp setting and provide extremely fast volumetric coverage, providing both the power and speed for imaging very obese patients (combining the power of two tubes), whole body trauma and cardiac imaging.

Dual Source Dual Energy (DSDE)

- Utilizes two X-ray tubes and two detectors to obtain simultaneous dual energy acquisition and data processing. X-ray tubes are set at different energies (different kV-settings, e.g., 80 kVp and 140 kV), which is the key to high sensitivity and specificity in imaging, customized for each patient and each acquisition.

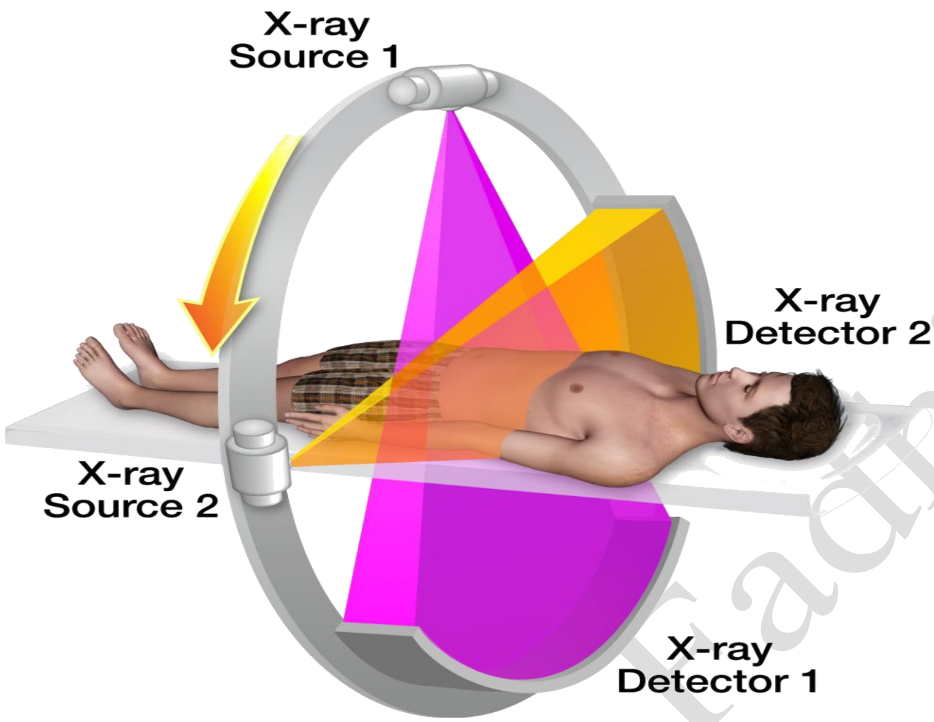


Fig. 8: Dual Source CT equipped with two data measurement systems (tubes and detectors)

Single Source Dual Energy (SSDE)

- Uses single X-ray tube with fast kilovoltage switching (low and high energies) (ie, rapid alternation between high and low kilovoltage settings). It is paired with a detector made of two layers (dual detector layers) that simultaneously detects & registers information from the both energies levels.

Unlike conventional CT, in which one image is acquired per location at a single energy setting (usually 120 or 140 kVp), in DECT two images are acquired per location at two different energies. In general, dual energy spectral data provide added insight over traditional structural only images by making it possible to differentiate not only between fat, soft tissue, and bone, but also between the calcifications and contrast material (iodine) on the basis of their unique energy-dependent attenuation profiles. Furthermore, functional parameters such as iodine concentration in the liver, lung, myocardium or tumors etc. can be acquired.

When using two energies, it is possible to delineate structures based solely on their attenuation differences between, for example 80 kVp and 140 kVp.

The inherent contrast generation of the image dataset depends on differences in photon attenuation of the various materials that constitute the human body (ie, soft tissue, air, calcium, fat). The degree that a material will attenuate the X-ray beam is dependent on: (1) tissue composition and (2) photon energy level and how closely it exceeds the k-edge (ie, inner electron shell binding energy) of the material. Therefore, tissue attenuation can be manipulated by changing photon energy levels.

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4. J. Hsieh, ***"Computed Tomography: Principles, Design, Artifacts, and Recent Advances"***, 2nd ed. Wiley Inter-science, Bellingham, Washington, USA, (2009)

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Radiological Techniques Department **قسم تقنيات الأشعة**

المادة: تقنيات اجهزة التصويرالمقطعي المحوسب

Computed Tomography EquipmentsTechniques

Second stage/ 2nd coarse **المرحلة: الثانية / الكورس الثاني**

Title: **العنوان:**

CT system design: (SSCT & MSCT)

- **X-Ray imaging system (gantry)**
- **X-Ray Tube , X-Ray tubes in MSCT (Straton x-ray tube)**

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Lec. Dr. Lamyaa Fadhil Abdul Hussein

Target population: **الفئة المستهدفة:**

Students of second class طلبة المرحلة الثانية

Scientific Content: المحتوى العلمي:

CT system design

The basic equipment configuration for CT represented by three major systems:

➥ the imaging system,

➥ the computer system,

➥ the image display, recording, storage, and communication system.

The three major systems are housed in separate rooms, as follows:

1. The imaging system is located in the scanner room.
2. The computer system is located in the computer room.
3. The display, recording, and storage system are located in the operator's room.

Today, CT scanners are typically housed in similar physical spaces that contain the three system components identified in Figure 1.

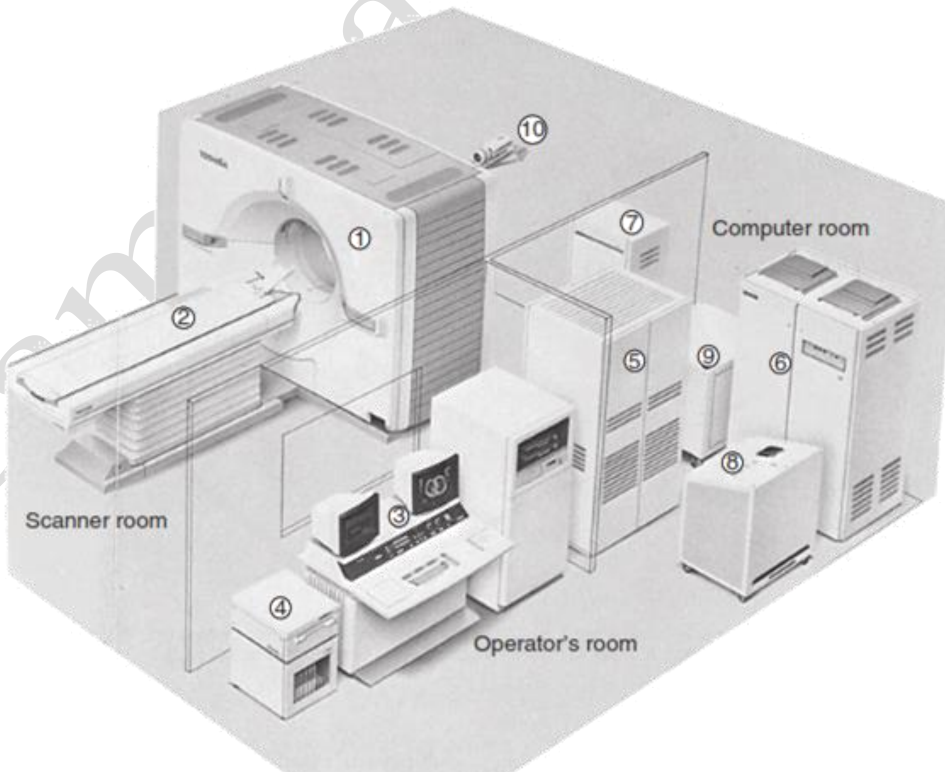

Fig.1 Components of a CT imaging system.
1, Gantry; *2*, patient couch; *3*, integrated console;
4, optical disk system including cassette storage;
5, high-speed processor system; *6*, x-ray high-voltage generator; *7*, couch control unit; *8*, system transformer I; *9*, system transformer II; *10*, patient observation system.

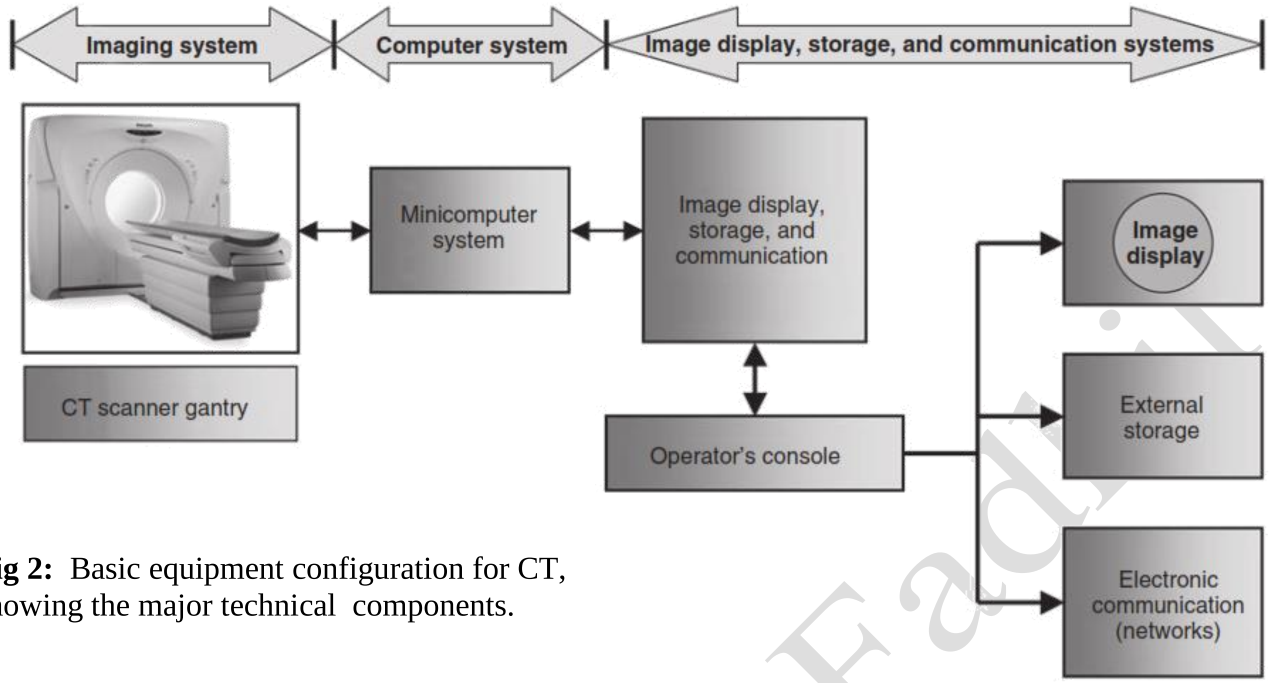

Fig 2: Basic equipment configuration for CT, showing the major technical components.

<u>The imaging system</u>

The purpose of the imaging system is

- to produce x rays, shape and filter the x-ray beam to pass through only a defined cross section of the patient,
- Detect and measure the radiation passing through the cross section,
- Convert the transmitted photons into digital information.

The major component of the imaging system is the **gantry,** that comprises several components housed in it, represented by; the **x-ray tube** and generator, **collimators**, **filter**, **detectors**, and detector electronics. The x-ray tube and generator are responsible for x-ray production. The radiation beam that emanates from the tube is filtered through a specially designed filter that protects the patient from low-energy rays and ensures beam uniformity at the detectors. The collimators help define the slice thickness and restrict the x-ray beam to the cross section of interest. The detectors capture the x-ray photons and convert them into electrical signals (analog information); the detector electronics, or ***data acquisition system (DAS),*** converts this information into digital data.

Gantry

The gantry assembly is the largest of these systems. It is a rotating mounted scan frame that surrounds the patient in a vertical plane.

Two important features of the gantry are the gantry aperture and the gantry tilting range. *The gantry aperture* is the opening in which the patient is positioned during the scanning procedure. The technologist can approach the patient from both the front and back of the gantry. Most scanners have a 70-cm aperture that facilitates patient positioning and helps provide access to patients in emergency situations.

The CT gantry must be capable of tilting (Fig.3) to accommodate all patients and clinical examinations. The degree of tilt varies between systems, but ±12 to ±30 degrees in 0.5-degree increments is somewhat standard.

Fig.3: the gantry tilting

The gantry also includes a set of laser beams to aid patient positioning. There are 3 types of lasers used in the CT Scan which are :

- Internal laser
- Wall-mounted laser
- Overhead laser. When lasers are positioned at zero setting, their intersection point is coincident with the center of the scan plane.

A) Internal laser

- All scanners contain an internal laser to identify the scan plane.
- This internal laser is mounted at the scanner bore of the CT scan machine.
- This type of laser used to mark the patient during the scan & treatment process.

B) Wall-mounted Laser

- These types of laser usually situated at the right and left of the room to align the patient and the couch.

C) Overhead Laser

- These lasers also have the same function as wall-mounted laser which is used to mark the patient. It may represent isocenter, field corners or markers.
- It is projecting at the same fixed distance as lateral laser bot orthogonal to the scan plane.
- These lasers are always capable of lateral movement because the scanner couch may not.

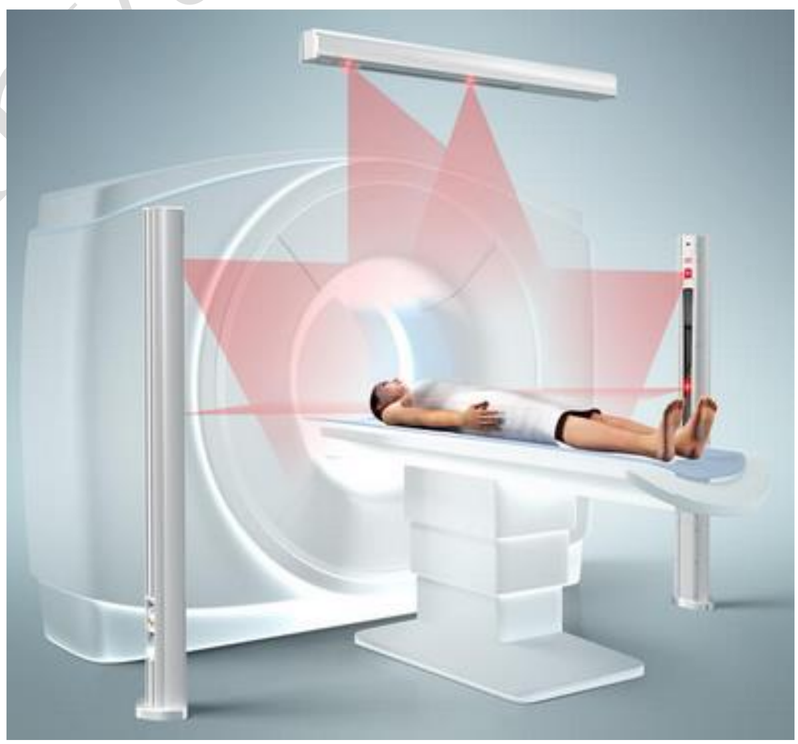

Fig.4: Types of lasers used in the CT Scan.

The essential parts of the gantry include:

- high-voltage generator
- x-ray tube,
- x-ray beam filters
- collimators
- detector array,

x-ray system

- patient support couch,
- slip rings
- DAS
- Mechanical support for each.

These subsystems receive electronic commands from the operating console and transmit data to the computer for image production and postprocessing tasks.

high-voltage Generator

Previously, CT scanners use three-phase power for the efficient production of x rays, but now use **high-frequency generators,** which are small, compact, and more efficient than conventional generators. These generators are located inside the CT gantry. In some scanners, the high-fre quency generator is mounted on the rotating frame with the x-ray tube; in others it is located in a corner of the gantry and does not rotate with the tube.

In a high-frequency generator (Fig.5), the circuit is usually referred to as a *high-frequency inverter circuit*.

The low-voltage, low-frequency current (60Hz) from the main power supply is converted to high- voltage, high-frequency current (500 to 25,000 Hz) as it passes through the components, as shown in Figure 5. Each component changes the low-voltage, low- frequency AC waveform to supply the x-ray tube with high-voltage, high-frequency direct current of almost constant potential. After high-voltage rectification and smoothing, the voltage ripple from a high-frequency generator is less than 1%. This makes the high- frequency generator more efficient at x-ray production than its predecessor.

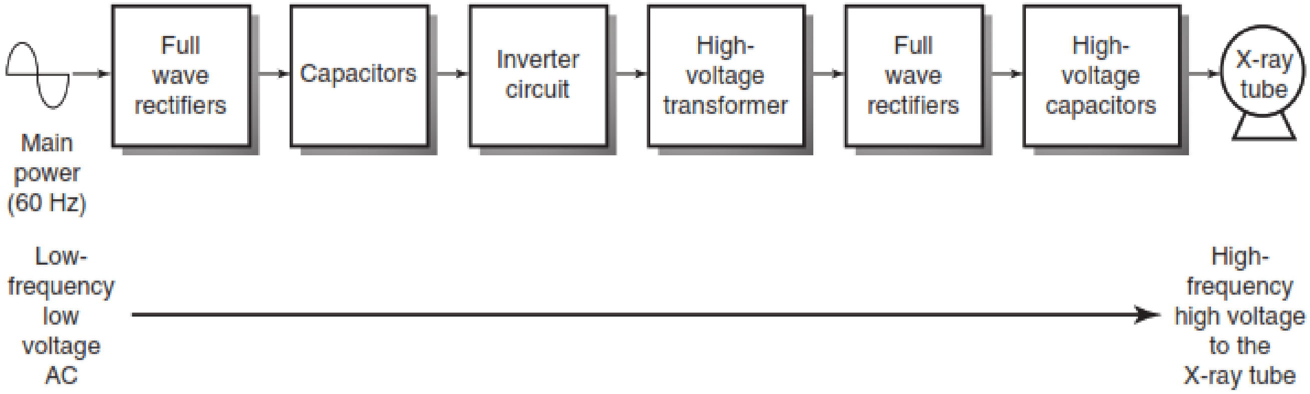

Fig. 5: The basic components of a high-frequency generator used in modern CT scanners.

The x-ray **exposure** technique obtained from these generators depends on the generator power output. The power ratings of CT generators vary and depend on the CT vendor; however, typical ratings can range from 20 to 100 kilowatts. More recently CT manufacturers have generators capable of 120 kW. An output capacity of, say, 60 kW will provide a range of kilovolt and milliampere settings, where 80 and 120 to 140 kV and 20 to 500 milliamperes (mA) are typical.

X-Ray Tubes

The radiation source requirement in CT depends on two factors:

(1) radiation attenuation, which is a function of radiation beam energy, the atomic number and density of the absorber, and the thickness of the object.

(2) The quantity of radiation required for transmission. X-ray tubes satisfy this requirement.

Rotating anode x-ray tubes have become common in CT because of the demand for increased output. These rotating anode tubes, an example of which is shown in Fig. 6, have large-diameter anode disk to facilitate the spatial resolution requirements of the scanner. The disk is usually made of a rhenium, tungsten, and molybdenum (RTM) alloy and other materials with a small target angle (usually 12 degrees) and a rotation speed of 3600 revolutions per minute (rpm) to 10,000 rpm (high-speed rotation). Fig.6, *B*, shows an upgraded tube based on the technology used in the tube shown in Fig.6, *A*.

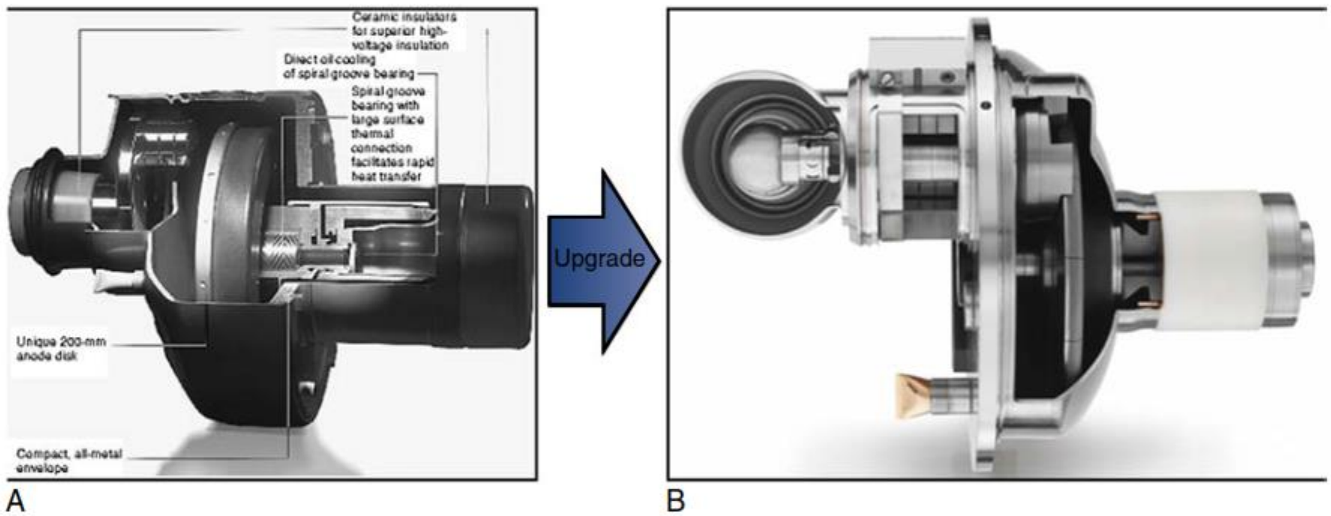

Fig. 6: A modern rotating anode x-ray tube used in CT scanners. The tube shown in **(B)** is an upgraded tube based on the technology used in the tube shown in **(A)**. (**A,** Courtesy Philips Medical Systems, Shelton, Conn; **B,** Courtesy Philips Healthcare.).

The introduction of spiral/helical CT with continuous rotation scanners has placed new demands on x-ray tubes. Because the tube rotates continuously for a longer period, compared with conventional scanners, the tube must be able to sustain higher power levels. Several technical advances in component design have been made to achieve these power levels and deal with the problems of heat generation, heat storage, and heat dissipation. For example, the tube envelope, cathode assembly, anode assembly including anode rotation, and target design have been redesigned.

The glass envelope ensures a vacuum, provides structural support of anode and cathode structures, and provides high-voltage insulation between the anode and cathode. Although the borosilicate glass provides good thermal and electrical insulation, electrical arcing results from tungsten deposits on the glass caused by vaporization. Tubes with metal envelopes, which are now common, solve this problem.

Metal envelope tubes have larger anode disks; for example, the tube shown in Figure (6) has a disk with a 200-mm diameter compared with the 120- to 160-mm diameter typical of conventional tubes. This feature allows the technologist to use higher tube currents. Heat-storage capacity is also increased with an improvement in heat dissipation rates.

The cathode assembly consists of one or more tungsten filaments positioned in a focusing cup.

The anode assembly consists of the disk, rotor , hub, and bearing assembly. The large anode disk is thicker than conventional disks; the three basic designs are (Fig.7):

- The conventional all-metal disk,
- The brazed graphite disk,
- The chemical vapor deposition (CVD) graphite disk.

In conventional tubes, the all-metal disk (Fig.7, *A*) consists of a base body made of titanium, zirconium, and molybdenum (TZM) with a focal track layer of 10% rhenium and 90% tungsten. It can transfer heat from the focal track very quickly. Unfortunately, tubes with this all-metal design cannot meet the needs of spiral/helical CT imaging because of their weight.

The brazed graphite anode disk (Fig. 7, *B*) consists of a tungsten-rhenium focal track brazed to a graphite base body. Graphite increases the heat- storage capacity because of its high thermal capacity, which is about 10 times that of tungsten. The material used in *the brazing process influences the operating temperature of the tube, and the higher temperatures result in higher heat-storage capacities and faster cooling of the anode. Tubes for spiral/helical CT scanning are based mostly on this type of design.

∗ *Is a metal-joining process in which two or more metal items are joined together by melting and flowing a filler metal into the joint, with the filler metal having a lower melting point than the adjoining metal.*

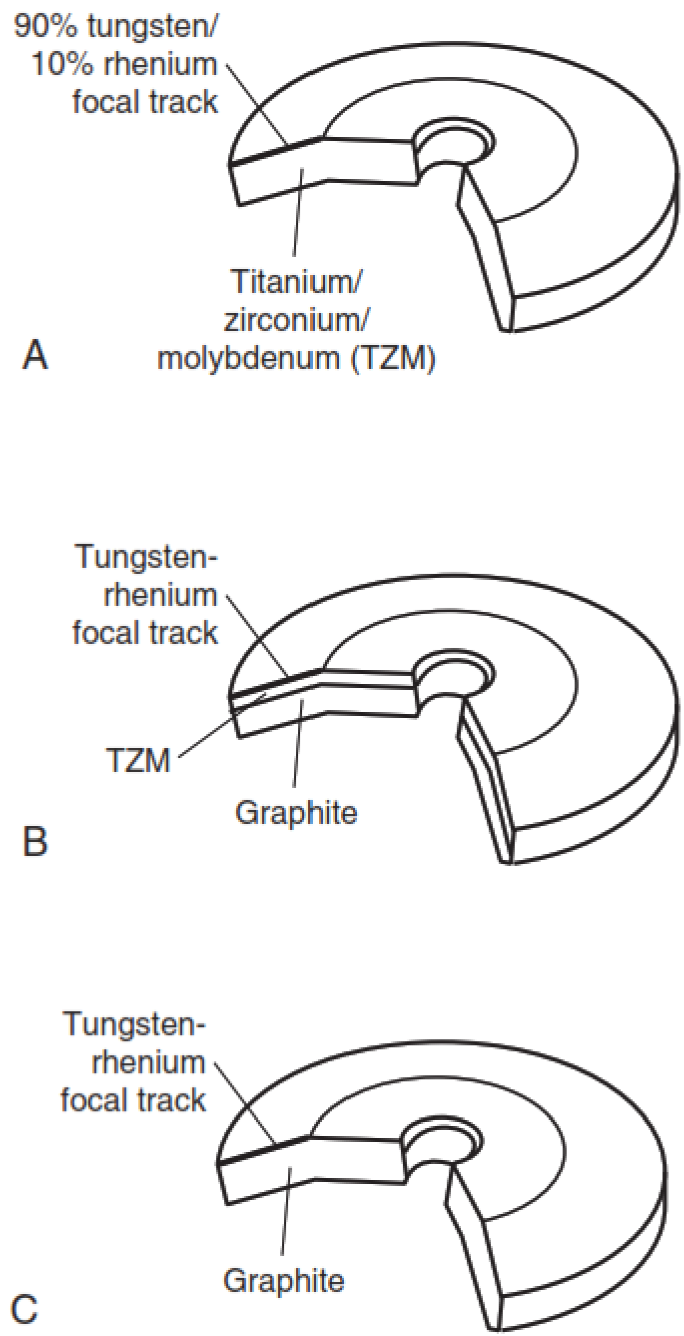

Fig. 7: Three types of disk designs for modern x-ray tubes used in CT scanners: **(A)** conventional all-metal disk; **(B)** brazed graphite anode disk; and **(C)** CVD graphite anode disk.

The final type of anode design (Fig. 7, *C*) is also intended for use in spiral/helical CT x-ray tubes. The disk consists of a graphite base body with a tungsten-rhenium layer deposited on the focal track by a chemical vapor process. This design can accommodate large, lightweight disks with large heat-storage capacity and fast cooling rates.

The purpose of the bearing assembly is to provide and ensure smooth rotation of the anode disk. In modern CT, smooth rotation of the disk is improved by using a liquid-bearing method (Fig. 8). The stationary shaft of the anode assembly consists of grooves that contain gallium-based liquid metal alloy. During anode rotation, the liquid is forced into the grooves and results in a hydroplaning effect between the anode sleeve and liquid.

The purpose of this bearing technology is to conduct heat away from the x-ray tube more efficiently than conventional ball bearings with improved tube cooling. Additionally, the liquid-bearing technology is free of vibrations and noise.

The working life of the tubes can range from about 10,000 to 40,000 hours, compared with 1000 hours, which is typical of conventional tubes.

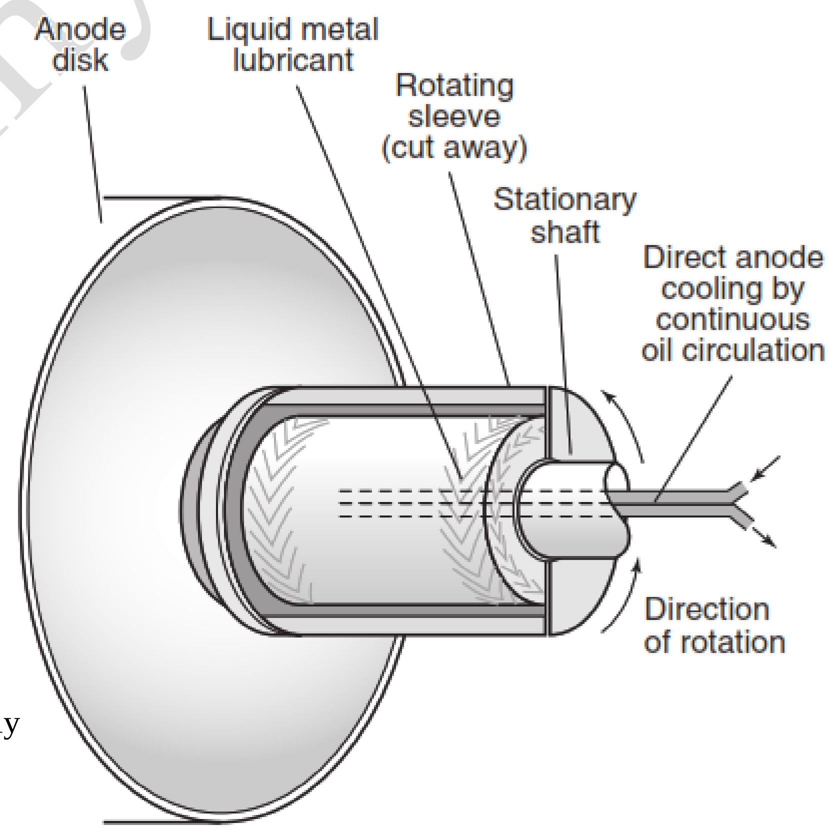

Fig. 8: The anode assembly of a modern x-ray tube used in CT imaging.
The main parts of the assembly are the disk, rotor, and bearing assembly that contains liquid metal lubricant.

Straton X-Ray Tube: A New X-Ray Tube for MSCT Scanning

As it is known, the fundamental problem with conventional x-ray tubes is that of heat dissipation and slow cooling rates. Efforts have been made to deal with these problems *by introducing various designs, such as large anode disks and the introduction of the compound anode design (RTM disk), which has higher heat-storage capacities and cooling rates.* Additionally, as gantry rotation times increase, higher milliampere values are needed to provide the same milliamperes per rotation. As the electrical load (milliamperes and kilovolts) increases, faster anode cooling rates are needed.

Despite these efforts, the problems of heat transfer and slow cooling rates still persist with MSCT scanners. To overcome these problems, a new type of x-ray tube called *Straton x-ray tube* has been introduced for use with MSCT scanners. This unique and revolutionary tube was designed by Siemens Medical Solutions (Siemens AG Medical Solutions, Erlangen, Germany).

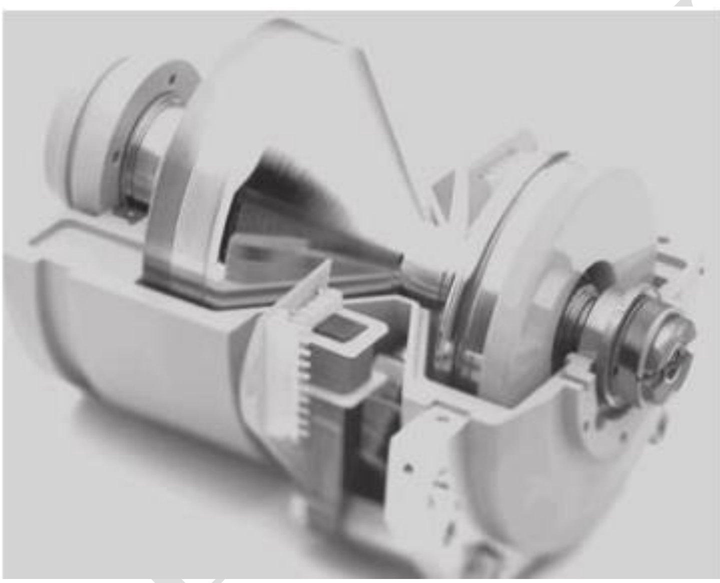

Fig.9: The Straton x-ray tube, a new x-ray tube for MSCT scans.

A photograph of the Straton x-ray tube is shown in Fig.9. As can be seen,. The tube is compact in design to much smaller than conventional x-ray tubes described earlier. This size ensures a fast gantry rotation of 0.37 seconds.

In the Straton x-ray tube, it is encased in a protective housing that contains oil for cooling, and the motor provides a rotation of the entire tube (which is immersed in oil). It is important to note that the anode is in direct contact with the oil (directly cooled tube). Because the anode is in direct contact with the circulating oil, very high cooling rates will result in, as illustrated in Fig.10.

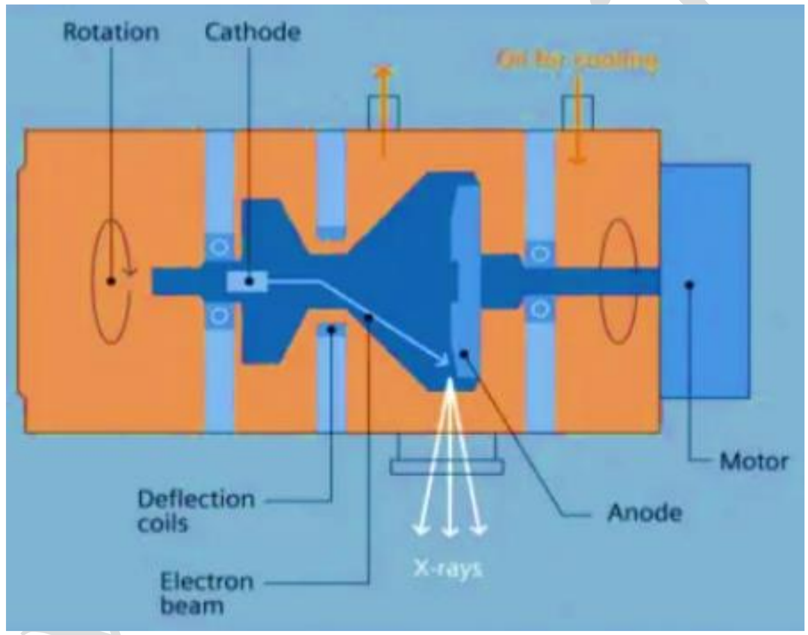

Fig. 10: The Straton x-ray tube. This diagram shows the anode and cathode structures, deflection coils, an electron beam, and a motor.

Another important feature of the Straton x-ray tube relates to the electron beam from the cathode. This beam is deflected to strike the anode at two precisely located focal spots (Fig.11) that vary in size. 1.1 mm, and 0.7 mm. The electron beam alternates at about 4640 times per second to create two separate x-ray beams that pass through the patient and fall on the detectors.

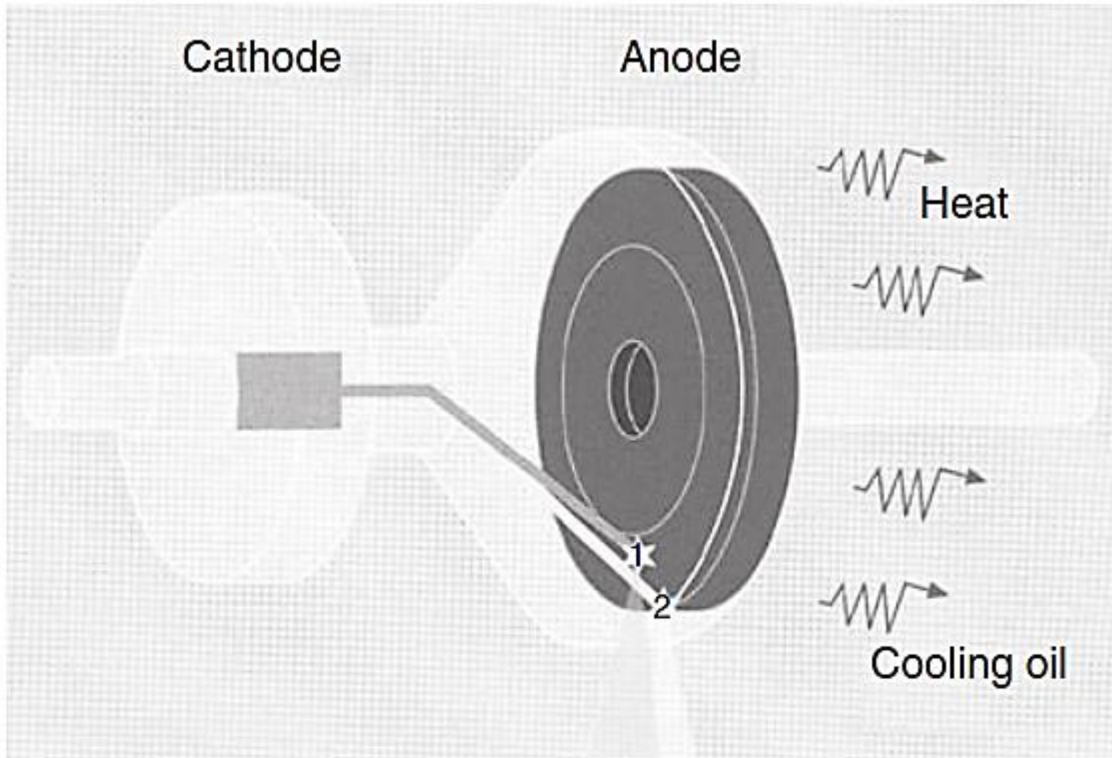

Fig. 11: An important feature of the Straton x- ray tube is that the electron beam from the cathode is deflected to strike the anode at two precisely located focal spots that vary in size.

The Advantages of Straton x-ray tubes

♣ Better Heat Dissipation

♣ Various size multiple focal spot

♣ Longer tube life

♣ high-speed volume scanning is possible

♣ Can be used in high KV and high mA technique for prolonged Duration. *Ie; (High mAs & long exposure times for increasing lengths of anatomic coverage)*.

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Radiological Techniques Department قسم تقنيات الأشعة

المادة: تقنيات اجهزة التصويرالمقطعي المحوسب

Computed Tomography EquipmentsTechniques

Second stage/ 2nd coarse المرحلة: الثانية / الكورس الثاني

Title: **العنوان:**

Collimation,

Filtration

Detector: Detector Characteristics & types

Name of the instructor: **اسم المحاضر:**

م.د.لمياء فاضل عبدالحسين

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Target population: **الفئة المستهدفة:**

Students of second class طلبة المرحلة الثانية

Scientific Content: المحتوى العلمي:

Filtration

In clinical CT, a special filter must be used, it serves a dual purpose:

1. Filtration removes long-wavelength x rays because they do not play a role in CT image formation; instead they contribute to patient dose. As a result of filtration, the mean energy of the beam increases and the beam becomes "harder".

Recall that the total filtration is equal to the sum of the inherent filtration and the added filtration. In CT the inherent filtration has a thickness of about 3 mm Al-equivalent. The added filtration, on the other hand, consists of filters that are flat or shaped filters made of copper sheets, for example, the thickness of which can range from 0.1 to 0.4 mm.

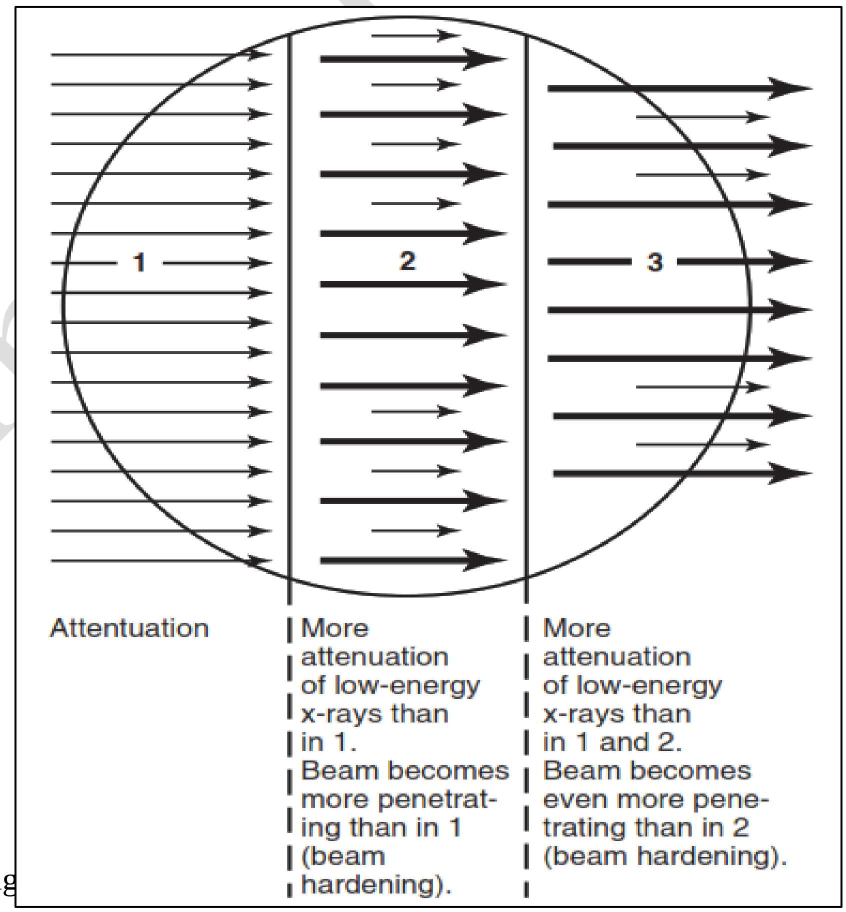

Fig. 1: Attenuation of radiation through a circular object. The beam becomes more penetrating (harder) in section 3 because of differences in attenuation in sections 1 and 2. The heavier arrows indicate less attenuation and more penetrating rays.

2. Filtration shapes the energy distribution across the radiation beam to produce uniform beam hardening when x rays pass through the filter and the object.

In Figure (1), the attenuation differs in sections 1, 2, and 3 and the penetration increases in sections 2 and 3. This results from the absorption of the soft radiation in sections 1 and 2, which is referred to as *hardening of the beam*. Because the detector system does not respond to beam-hardening effects for the circular object shown, "the problem can be solved by introducing additional filtration into the beam".

Special shaped filters conform to the shape of the object and are positioned between the x-ray tube and the patient (Fig. 2). These filters are called shaped filters, such as the "bowtie" filter, and are usually made of Teflon, a material that has a low atomic number and high density. "The term 'bowtie' applies to a class of filter shapes featuring bilateral symmetry with a thickness that increases with the distance from the center. Bowtie filters compensate for the difference in beam path length through the axial plane of the object such that a more uniform fluence can be delivered to the detector.

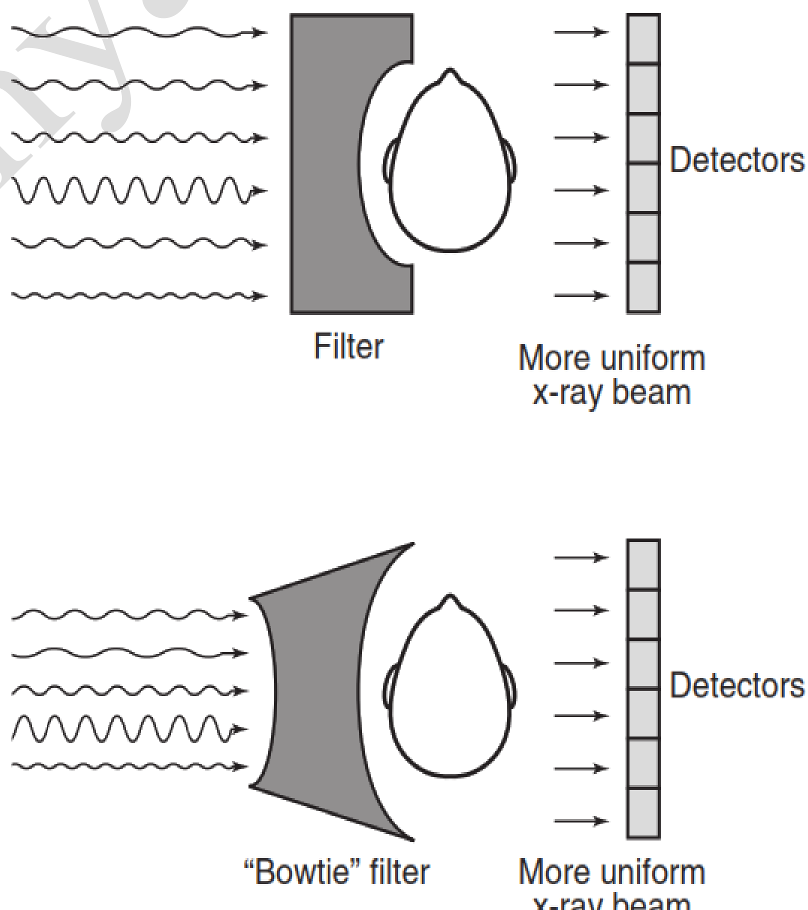

Fig.2: Two types of beam-shaping filters for use in CT. These filters attenuate the beam so that a more uniform (monochromatic) beam falls onto the detectors.

Collimation

Collimation is required during CT imaging for precisely the same reasons as in conventional radiography. Proper collimation reduces patient radiation dose by restricting the beam to the anatomy of interest only. The basic collimation scheme in CT is shown in Fig. 3, with adjustable prepatient and postpatient (predetector collimators).

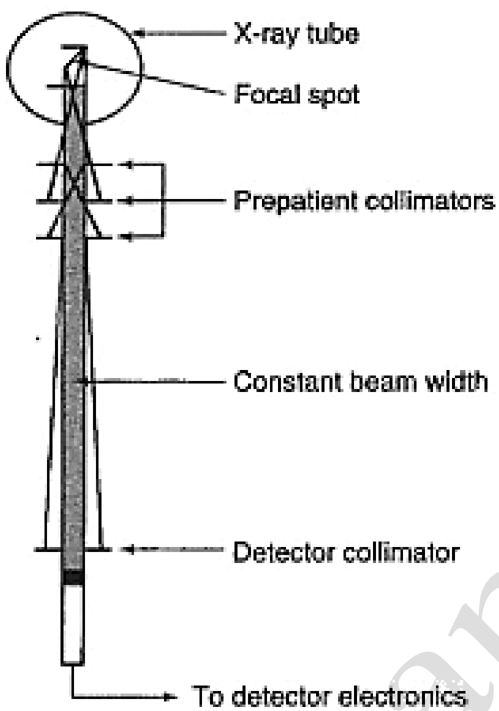

Fig.3: Collimation scheme typical of CT scanning

In general, a set of collimator sections is carefully arranged to shape the beam. Both proximal and distal collimators are arranged to ensure a constant beam width at the detector. Detector collimators can shape the beam and remove scattered radiation. Moreover, the collimator section at the distal end of the collimator assembly also helps define the thickness of the slice to be imaged. Various slice thicknesses are available depending on the type of scanner.

Adaptive Section Collimation

The introduction of MSCT scanners posed some challenges with the design of the collimation scheme especially as the detectors become wider. The problems are related to what has been referred to as overscanning and overbeaming.

- Overbeaming "relates to x-ray beams being slightly wider than the detector which means that patients are exposed over a small area without the signal being detected".
- Overscanning "refers to exposure of the patient outside the imaged range which occurs for spiral CT with multi-row detectors at the start and the end of the scan".

These two problems result in increased dose to the patient. For example, overscanning may result in an increase of 5% to 30% in dose keeping the length of the scan in mind.

The problems of overscanning and overbeaming can be solved using a technique called ***adaptive section collimation***. With adaptive section collimation (see Fig. 4) "parts of the x-ray beam exposing tissue outside of the volume to be imaged are blocked in the z-direction by dynamically adjusted collimators at the beginning and at the end of the CT scan".

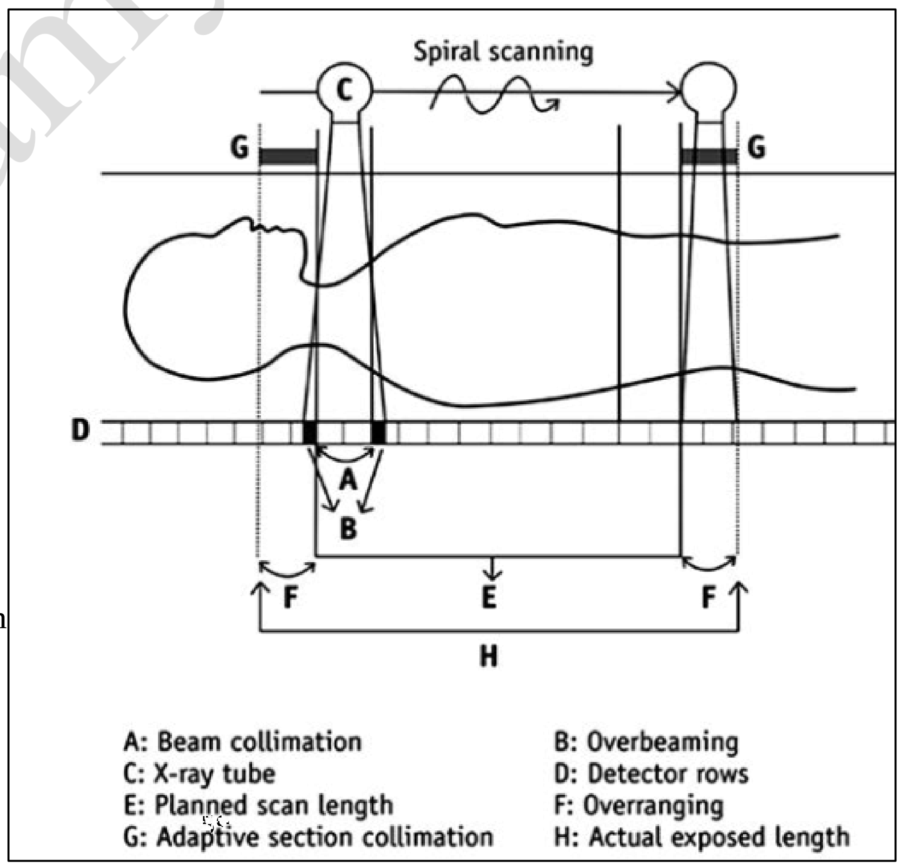

FIG.4: Graphic illustration of the concepts of overbeaming, overranging,and adaptive section collimation technology during spiral scanning.

Detectors

CT detectors capture the radiation beam from the patient and convert it into electrical signals, which are subsequently converted into binary coded information.

Detector Characteristics

Detectors exhibit several characteristics essential for CT image production affecting good image quality, such as: ✦ efficiency & ✦ Response time (speed of response).

Efficiency refers to the ability to capture, absorb, and convert x-ray photons to electrical signals. CT detectors must possess high capture efficiency, absorption efficiency, and conversion efficiency. Three important factors contributing to the detector efficiency are:

✦***Capture efficiency*** refers to the efficiency with which the detectors can obtain photons transmitted from the patient; the size of the detector area facing the beam and distance between two detectors determines capture efficiency.

✦***Absorption efficiency*** refers to the number of photons absorbed by the detector and depends on the atomic number, physical density, size, and thickness of the detector face.

✦***Conversion efficiency*** refers to the ability to accurately convert the absorbed x-ray signal into an electrical signal.

The Overall efficiency is the product of the three, and it generally lies between 0.45 and 0.85. A value of less than 1 indicates a nonideal detector system and results in a required increase in patient radiation dose if image quality is to be maintained. The term *dose efficiency* sometimes has been used to indicate overall efficiency.

*The **response time*** of the detector refers to the speed with which the detector can detect an x-ray event and recover to detect another event. Response times should be very short (i.e., microseconds).

X-ray detectors used in CT systems must (a) have a high overall efficiency to minimize the patient radiation dose, have a large dynamic range, (b) be very stable with time, and (c) be insensitive to temperature variations within the gantry.

Types

The conversion of x-rays to electrical energy in a detector is based on two fundamental principles (Fig. 5). Scintillation detectors (luminescent materials) convert x-ray energy into light, after which the light is converted into electrical energy by a photodetector (Fig.5, *A*). Gas- ionization detectors, on the other hand, convert x-ray energy directly to electrical energy, (Fig. 5, *B)*.

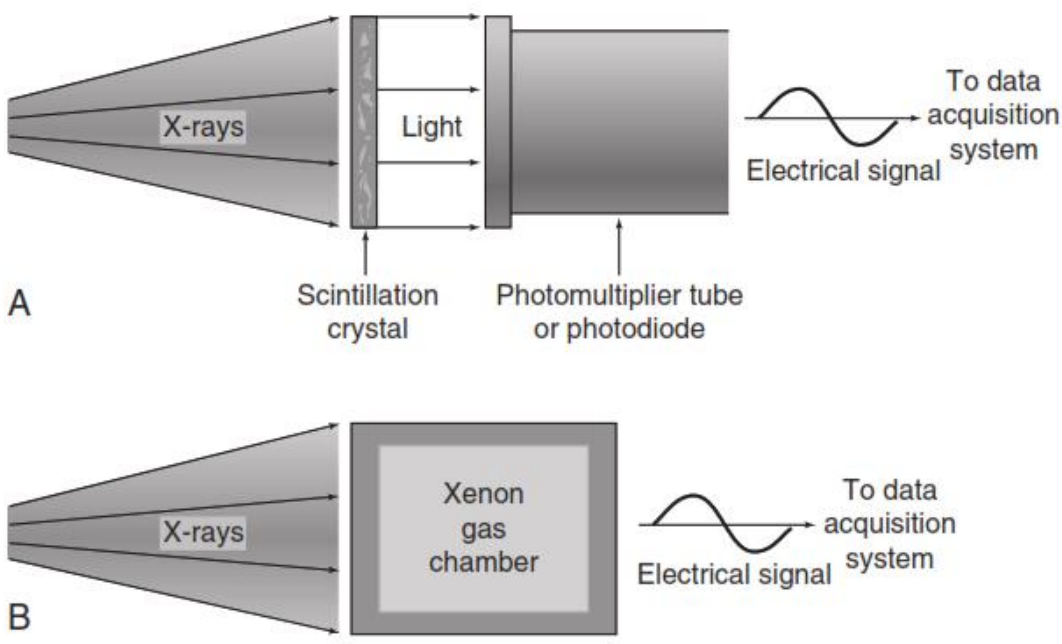

Fig. 5: Two methods for converting x-ray photons into electrical energy. **A,** Scintillation crystal detection and conversion scheme. **B,** Conversion of x rays into electrical energy through gas ionization.

1. Scintillation detectors

Are solid-state detectors that consist of a scintillation crystal coupled to a photo-diode tube. When x rays fall onto the crystal, flashes of light, or scintillations, are produced. The light is then directed to the photomultiplier, or PM tube. As illustrated in Fig.6, the light from the crystal strikes the photocathode of the PM tube, which then releases electrons. These electrons cascade through a series of dynodes that are carefully arranged and maintained at different potentials to result in a small output signal.

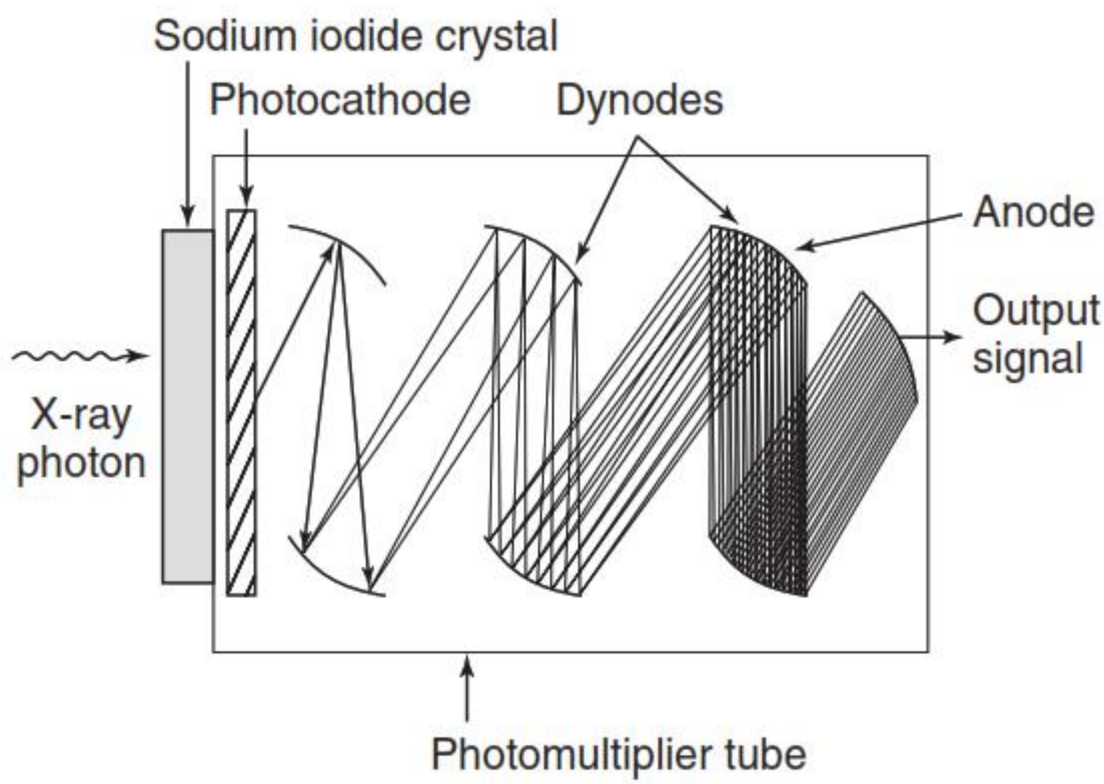

Fig. 6: Schematic representation of a scintillation detector based on the photomultiplier tube.

In CT, scintillation detectors must exhibit a high light output, short primary decay time (up to tens of μs), low afterglow, radiation damage resistance, light-output stability (time, temperature), compact packaging, and easy machining.

As a result, single crystals and polycrystalline ceramics have become popular scintillators for use in CT imaging. In the past, early scanners used *sodium iodide crystals* coupled to PM tubes. Because of afterglow problems and the limited dynamic range of sodium iodide, other crystals such as *calcium fluoride* and *bismuth germanate* were used in later scanners.

Scintillation materials currently used with photo- diodes are *cadmium tungstate* ($CdWO_4$) and a *ceramic material* made of high-purity, rare earth oxides based on doped rare earth compounds.

2. Gas-Ionization Detectors

Gas-ionization detectors, which are based on the principle of ionization, were introduced in third- generation scanners. The basic configuration of a gas- ionization detector consists of a series of individual gas chambers, usually separated by tungsten plates carefully positioned to act as electron collection plates (Fig. 7). When x rays fall on the individual chambers, ionization of the gas (usually xenon) results and produces positive and negative ions. The positive ions migrate to the negatively charged plate, whereas the negative ions are attracted to the positively charged plate. This migration of ions causes a small signal current that varies directly with the number of photons absorbed.

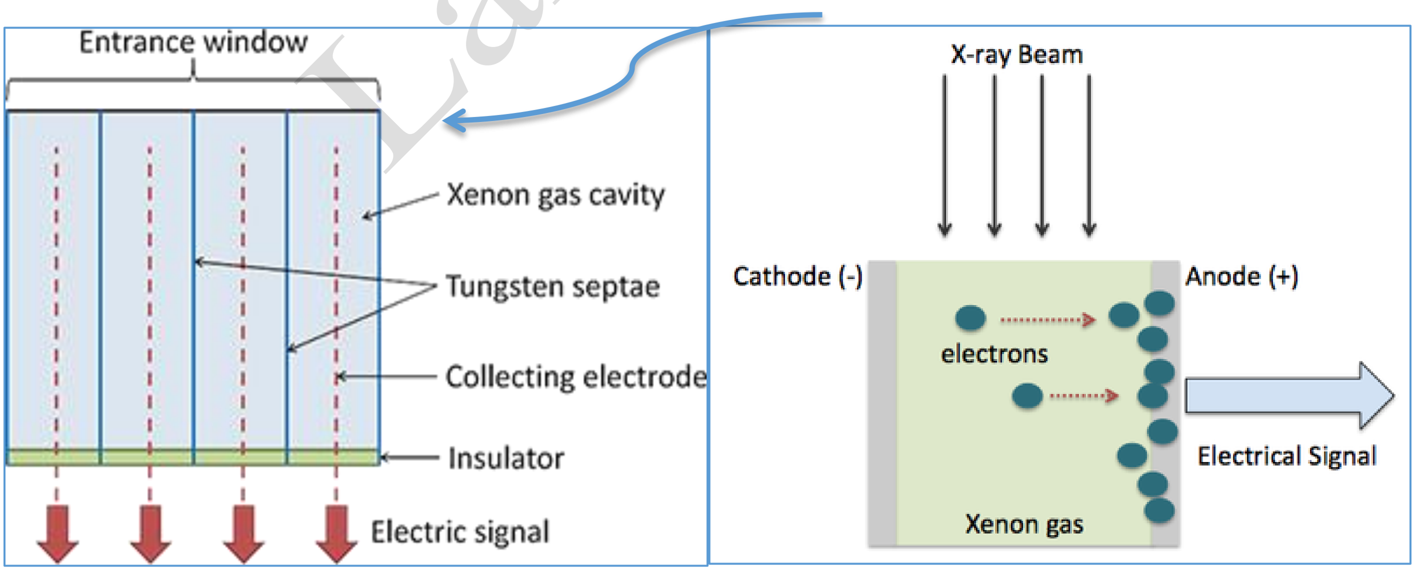

Fig. 7: Schematic representation of a Gas-ionization detectors.

The gas chambers are enclosed by a relatively thick ceramic substrate material because the xenon gas is pressurized to about 30 atmospheres to increase the number of gas molecules available for ionization. Xenon detectors have excellent stability and fast response times. However, their quantum detection efficiency (QDE) is less than that of solid-state detectors. As reported in the past, the QDE is 95% to 100% for crystal solid-state scintillation detectors and 94% to 98% for ceramic solid-state detectors, and it is only
50% to 60% for xenon gas detectors.

∗ It is important to note that with the introduction of MSCT scanners with their characteristic multirow detector arrays, gas-ionization detectors are not used anymore. MSCT scanners are all based on the use of solid-state detector arrays.

Detector electronics (DAS)

The DAS refers to the detector electronics positioned between the detector array and the computer (Fig. 8). Because the DAS is located between the detectors and the computer, it performs three major functions: (1) measuring the transmitted radiation beam, (2) encoding these measurements into binary data, and (3) transmitting the binary data to the computer.

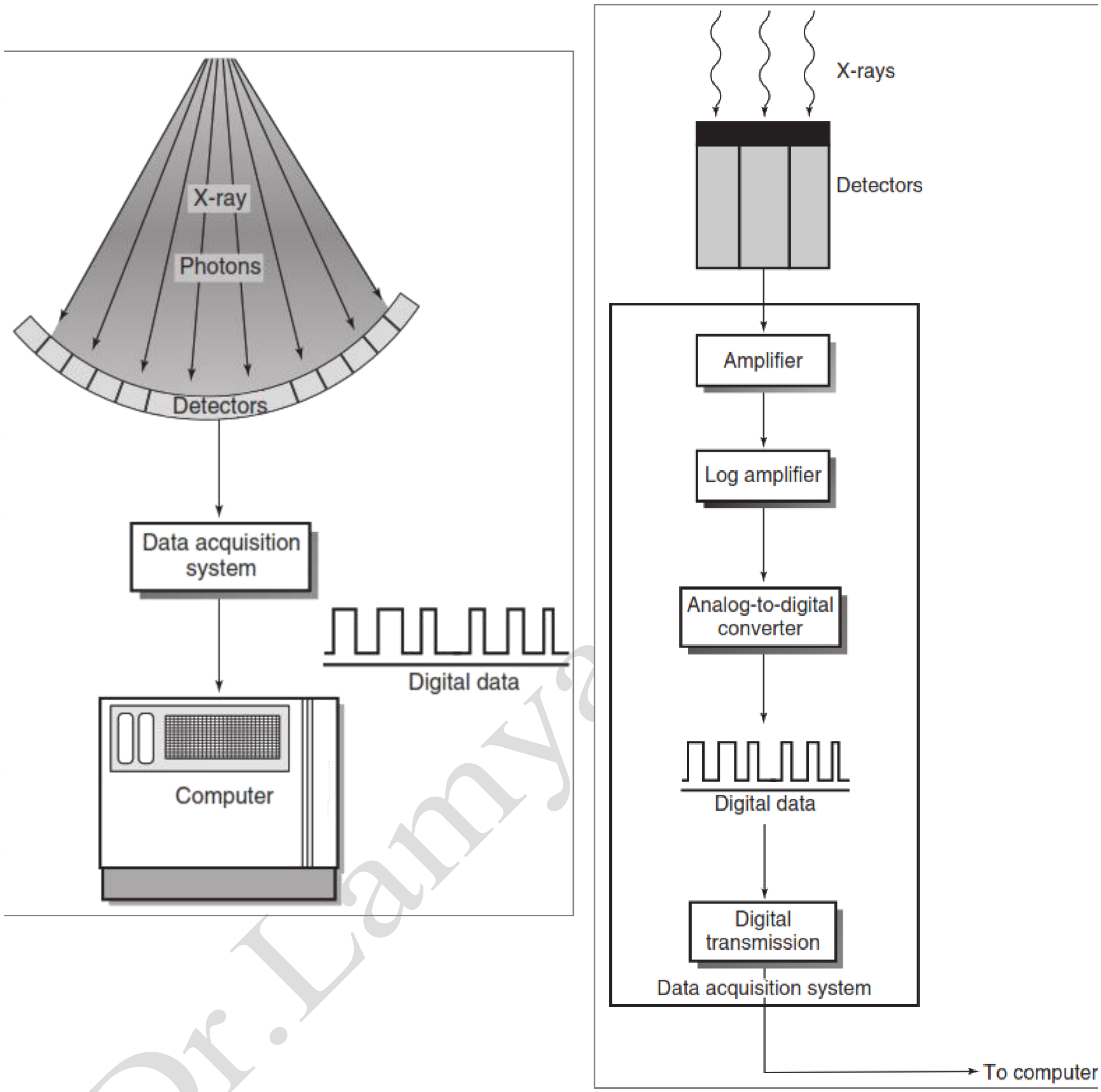

Fig. 8: (A) Position of the data acquisition system in CT.
(B) Essential components of the data acquisition system in CT.

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المادة: تقنيات اجهزة التصويرالمقطعي المحوسب

Computed Tomography EquipmentsTechniques

Second stage/ 2nd coarse **المرحلة: الثانية / الكورس الثاني**

Title: **العنوان:**

- **Control Console**
- **Patient Table or Couch**
- **Computer system: image display, recording, storage, and communication system**

Name of the instructor: **اسم المحاضر:**

م.د.لمياء فاضل عبدالحسين

Lec. Dr. Lamyaa Fadhil Abdul Hussein

Target population: **الفئة المستهدفة:**

Students of second class طلبة المرحلة الثانية

Scientific Content: المحتوى العلمي:

The Operating Console

Computed tomography imaging systems can be equipped with two or three consoles.

- One console is used by the CT radiologic technologist to operate the imaging system.
- Another console may be available for a technologist to post process images to annotate patient data on the image (e.g., hospital identification, name, patient number, age, gender) and to provide identification for each image (e.g., number, technique, couch position). The second monitor also allows the operator to view the resulting image before transferring it to the physician's viewing console.
- A third console may be available for the physician to view the images and manipulate image contrast, size, and general visual appearance. This is in addition to several remote imaging stations available to the radiologist and other physicians.

The operating console contains meters and controls for selection of proper imaging technique factors, for proper mechanical movement of the gantry and the patient couch, and for the use of computer commands that allow image reconstruction and transfer.

The physician's viewing console accepts the reconstructed image from the operator's console and displays it for viewing and diagnosis.

A typical operating console contains controls and monitors for the various technique factors. Operation is usually in excess of 120 kVp.The maximum mA is usually 400mA and is modulated (varied) during imaging according to patient thickness to minimize the patient radiation dose.

Patient Table or Couch

The patient table is an important and highly integrated component of the CT scanner. The CT computer controls table motion using precision motors with telemetric feedback, for patient positioning and CT scanning. This is critically important in helical scanning where the coordination of tube rotation and table movement is essential. The patient table (or couch) can be retracted from the bore of the CT gantry and lowered to sitting height to allow the patient to comfortably get on the table, usually in the supine position as shown in Figure 1. Under the CT technologist's control, the system then moves the table upward and inserts the table with the patient into the bore of the scanner. A series of laser lights provide references in multiple directions to allow the patient to be centered in the bore (both laterally and in terms of table height) and to adjust the patient longitudinally, as mentioned in the previous lecture.

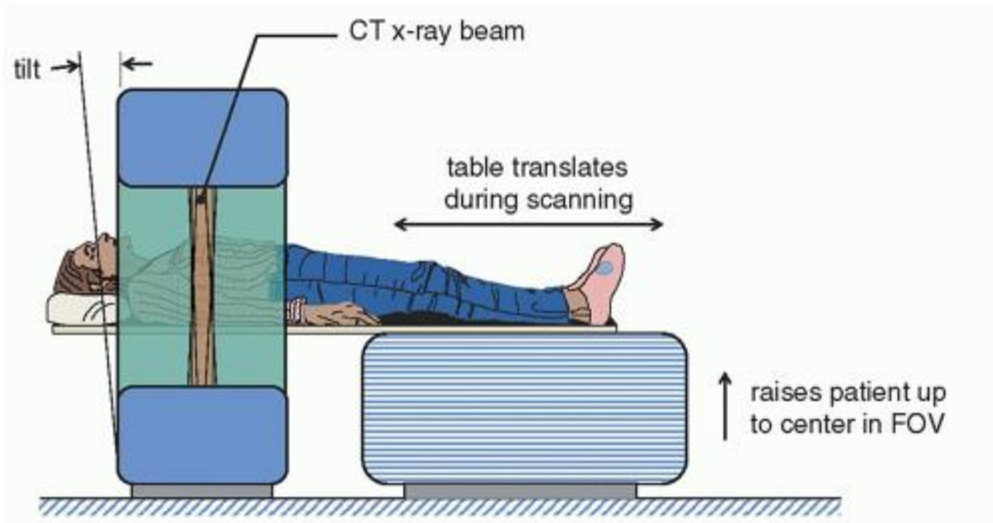

Fig. (1): The patient table is a perfunctory but surprisingly high-tech component of a CT scanner. The patient table lowers to sitting height to allow patients—including the elderly and physically impaired—to sit on the table and reposition to a prone or supine position, with help from the attending technologist.

<u>Image Display, Storage, and Communication</u>

The third and final step in the CT process involves image display, storage, and communication. After the CT image has been reconstructed, it exits the computer in digital form. This must be converted to a form that is suitable for viewing and meaningful to the observer.

Display device. The grayscale image is displayed on a television monitor (Cathode ray tube [CRT]) or liquid crystal display, which is an essential component of the control or viewing console. In the display and manipulation of grayscale images for diagnosis, it is important to optimize image fidelity (i.e., the faithfulness with which the device can display the image). *This is influenced by* physical characteristics such as luminance, resolution, noise, and dynamic range.

Resolution, however, is an important physical parameter of the grayscale display monitor and is related to the size of the pixel matrix**,** or matrix size. The display matrix can range from 64 × 64 to 1024 ×1024, but high-performance monitors can display an image with a 2048 × 2048 matrix.

Windowing

Because the human eye cannot distinguish all of the possible 5,000–10,000 shades of gray, the grayscale of the CT image is limited to be composed of a range of (+1000 to −1000) that represent varying shades of gray, that appreciated to the human eye.

The process of limiting the number of shades of gray presented on the CT image to optimize viewing by the human eye is called ***windowing***. In other words, windowing is a term used to refer to the fine adjustments made in the image on the computer screen to enhance viewing and emphasize various tissues of interest. Windowing is performed by the reader at the workstation (are located on the control console) after the image is obtained, reconstructed and transferred to the viewing workstation.

The specific number of shades of grey chosen for presentation the CT image's on the compressed scale is called the ***window width (WW)*** and artificially defines the number of shades of grey presented to the viewer; (*controls the CT image contrast)*.

While ***the window level (WL)*** is defined as the center or midpoint of the CT number range that composes a CT image's gray scale; (*controls the CT image brightness*).

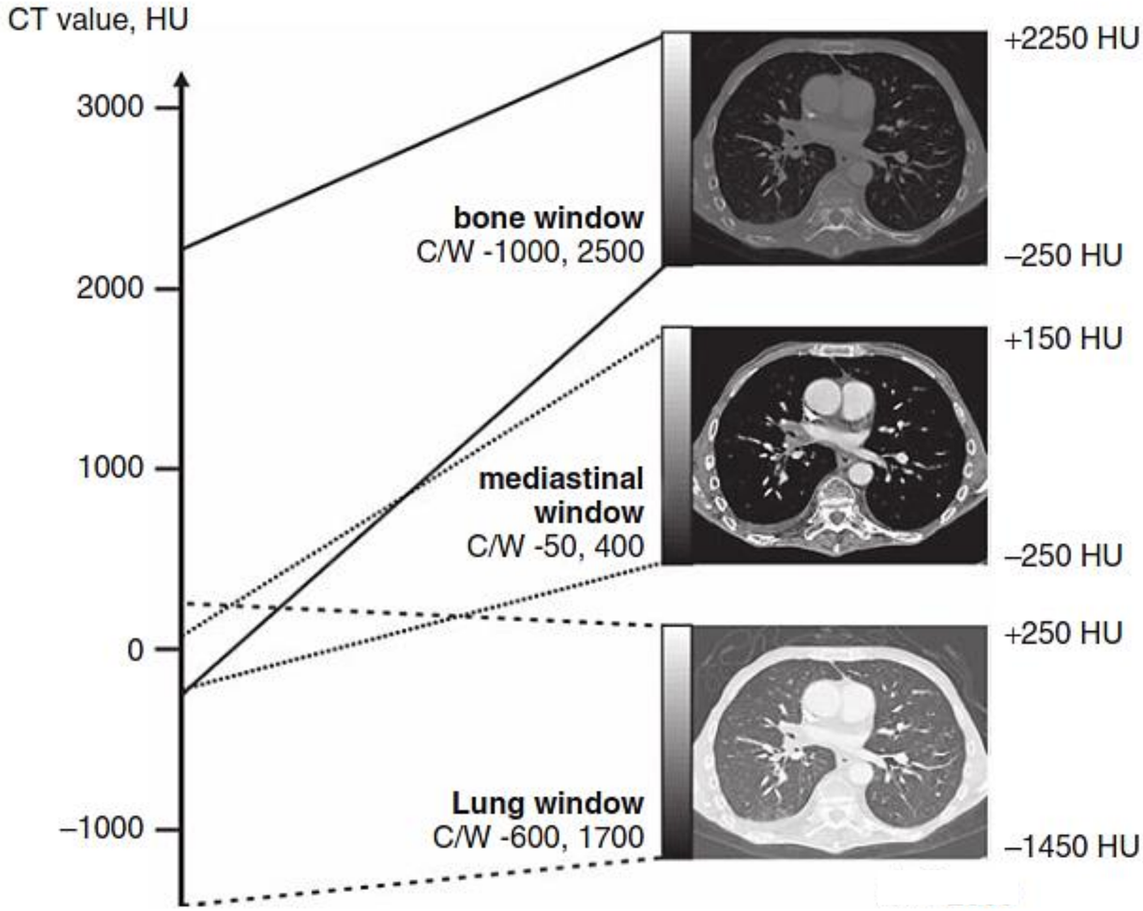

Fig. 2: Windowing is a digital image postprocessing operation intended to alter the image contrast (a function of the WW) and the image brightness (a function of the window center, C, or WL, as it is often referred).

The image contrast is optimized for the anatomy under study; therefore, specified values of WW and WL or C must be used during the initial scanning of the patient. Note that in (Figure 2), three windows are shown: the bone window (optimized for imaging bone), the **mediastinal** window (optimized for imaging the mediastinal structures), and the lung window (optimized for imaging the lungs).

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Computed Tomography EquipmentsTechniques

Second stage/ 2nd coarse **المرحلة: الثانية / الكورس الثاني**

Title: العنوان:

Reconstruction methods:

- **Backprojection reconstruction**
- **Filtered Backprojection**

Name of the instructor: اسم المحاضر:

م.د.لمياء فاضل عبدالحسين

Lec. Dr. Lamyaa Fadhil Abdul Hussein

Target population: الفئة المستهدفة:

Students of second class طلبة المرحلة الثانية

◣ Image Reconstruction

The image obtained in CT is different from that obtained in conventional radiography, Fig.1, in which rays form an image directly on the image receptor. While with CT imaging systems, it is created from data received and represent a depictions of relative attenuation of x rays as they pass through the body. The x-rays from a stored electronic image that is displayed as a matrix of intensities.

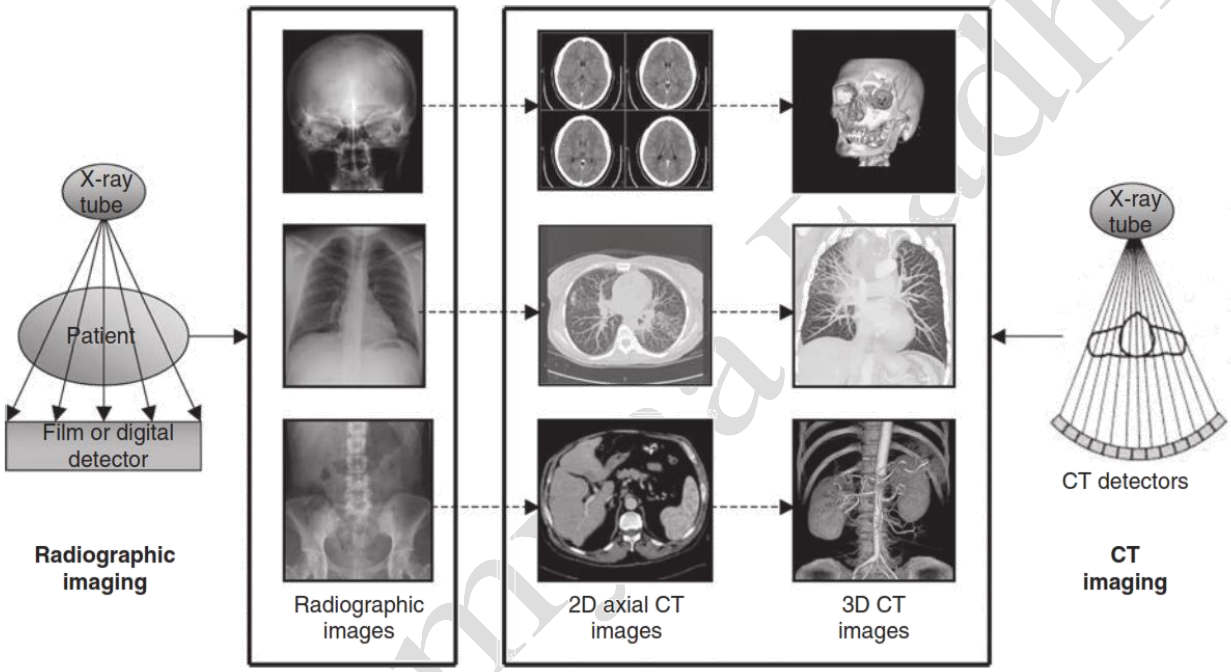


Fig. (1): The most conspicuous difference between conventional radiographic imaging and CT imaging.

A tissue's attenuating ability is related to its density and represents the likelihood that an x ray photon will pass through the tissue to be recorded by the detectors rather than interacting with tissue's atoms (absorption of the x rays into the tissue) which prevents the photon from reaching the detector at all. A particular tissue's x-ray attenuating ability is expressed by its **attenuation coefficient**, **μ** (explained earlier). The higher the μ value, the lower number of photons that reach the detector when passing through that tissue type. The μ value is directly related to the tissue's density. That is, the higher the tissue density, the higher its μ value.

However, the attenuation coefficient of a tissue is not constant and may be altered by the tissue thickness and the energy of the x ray photon (KeV).

◣ Image Reconstruction Techniques

Image reconstruction is a mathematical process that generates tomographic images from x-ray projection data acquired at many different angles around the patient. The reconstruction process is based on the use of an algorithm that uses the attenuation data measured by detectors to systematically build up the image for viewing and interpretation.

Image reconstruction involves several algorithms to calculate all the μ from a set of projection data. The algorithms applicable to CT include back-projection, iterative methods, and analytic methods.

Currently, there are currently two forms of image reconstruction:

- Filtered back-projection (FBP) and
- Iterative reconstruction (IR).

<u>Back-Projection</u>

Back-projection is a simple procedure that does not require much understanding of mathematics. Back-projection, also called the “summation method” or “linear superposition method. Back- projection can be best explained with a graphical or numerical approach.

Consider four beams of x rays that pass through an unknown object to produce four **projection profiles** P_1, P_2, P_3, and P_4 (Fig. 2). The problem involves the use of these profiles to reconstruct an image of the unknown object (black dot) in the box.

The projected datasets are back-projected to form the corresponding images *BP1*, *BP2*, *BP3*, and *BP4*. The reconstruction involves summing these back-projected images to form an image of the object.

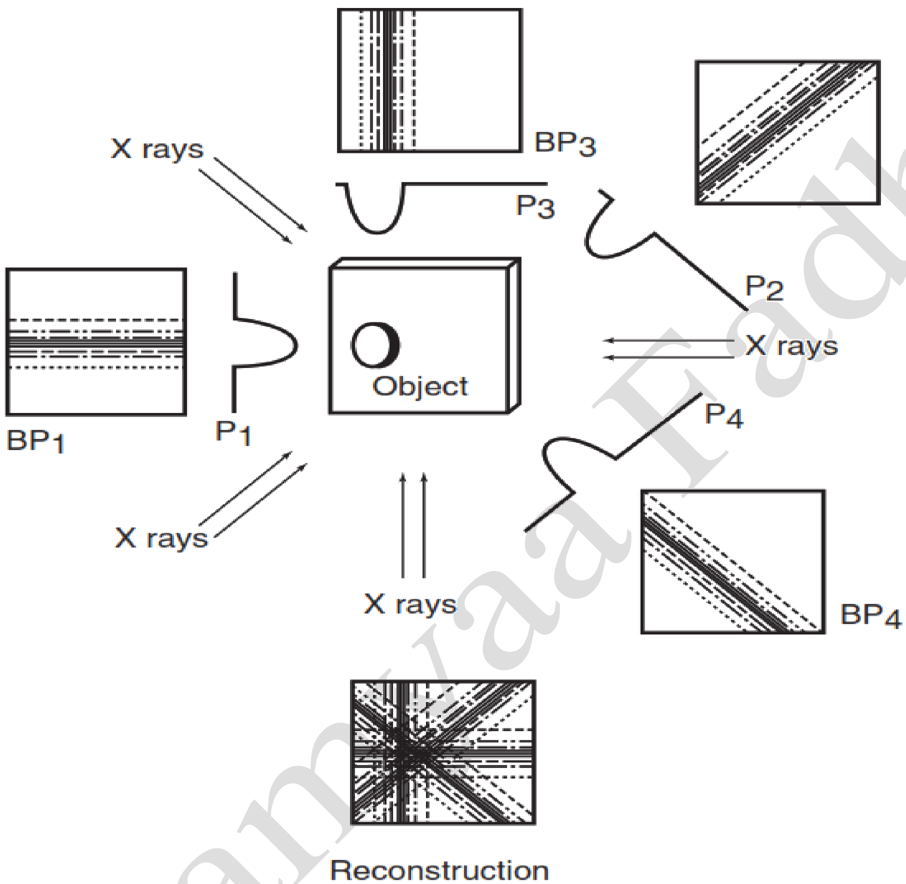


Fig.2: Graphic representation of the back-projection reconstruction technique.

BP involves summing the data from hundreds of projection angles to reconstruct the image. Since the data from a projection angle of 0° is identical to the data from a projection angle of 180°, only the data from a 180° gantry rotation is necessary to reconstruct the full CT image. The displayed CT image is composed of the CT number data (Hounsfield unit data) from the summed projection information.

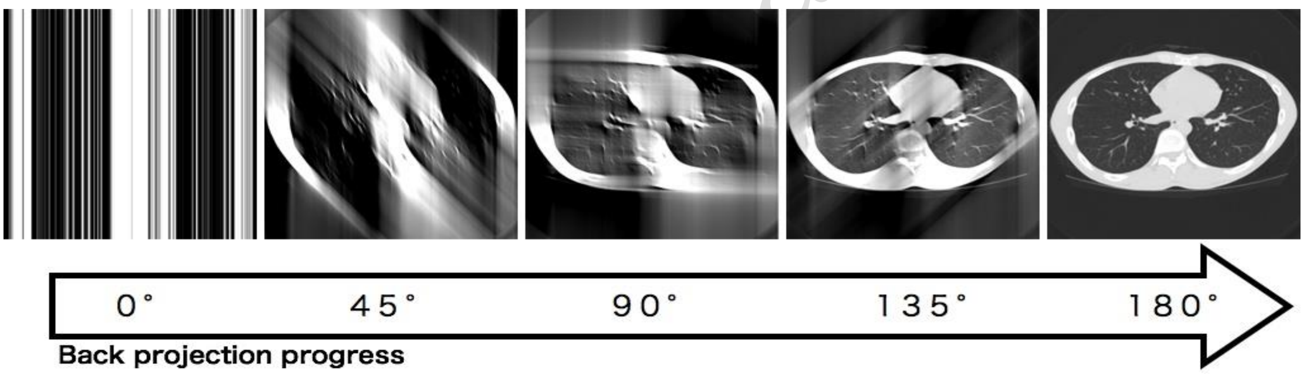
*Back-projection can also be explained with the following 2 × 2 matrix***:**

$$\begin{array}{cccc} I_0 \rightarrow & \boxed{\mu_1} & \boxed{\mu_2} & x \rightarrow I_1 \\ I_0 \rightarrow & \boxed{\mu_3} & \boxed{\mu_4} & x \rightarrow I_2 \\ & \leftarrow X \rightarrow & \leftarrow X \rightarrow & \\ & \downarrow & \downarrow & \\ & I_3 & I_4 & \end{array}$$

Four separate equations can be generated for the four unknowns, μ_1, μ_2, μ_3, and μ_4:

$$I_1 = I_0 e^{-(\mu_1+\mu_2)x}$$
$$I_2 = I_0 e^{-(\mu_3+\mu_4)x}$$
$$I_3 = I_0 e^{-(\mu_1+\mu_3)x}$$
$$I_4 = I_0 e^{-(\mu_2+\mu_4)x}$$

A computer can solve these equations very quickly.

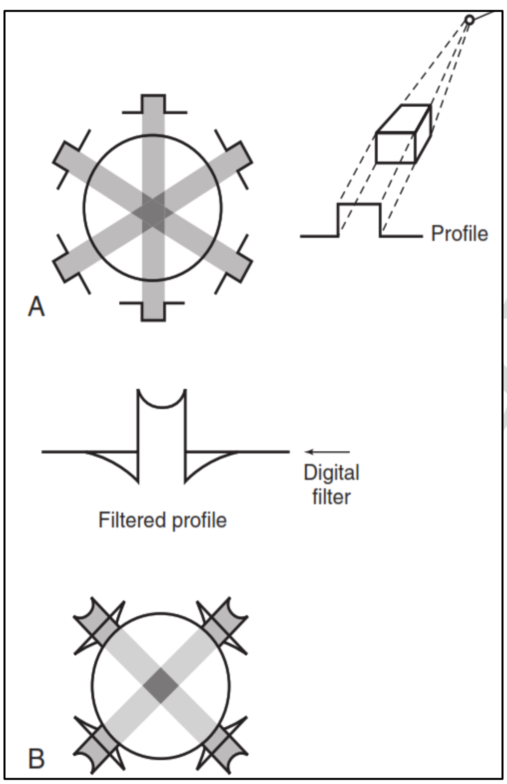


BP advantages include its relatively short time for complete reconstruction (≤30–40 slices per second). *The problem with the back-projection technique* is that it does not produce a sharp image of the object and therefore is not used in clinical CT. The most striking artifact of back-projection is the typical star pattern that occurs because points outside a high- density object receive some of the back-projected intensity of that object.

Filtered Back-Projection

Filtered back-projection is also referred to as the ***convolution method*** (Fig. 3). The projection profile is filtered or convolved to remove the typical starlike blurring that is characteristic of the simple back-projection technique. The steps in the filtered back-projection method (Fig. 3, *B*) are as follows:

1. All projection profiles are obtained.
2. The logarithm of the data is obtained.
3. The logarithmic values are multiplied by a digital filter, or convolution filter, to generate a set of filtered profiles.
4. The filtered profiles are then back-projected.
5. The filtered projections are summed and the negative and positive components are therefore canceled, which produces an image free of blurring.



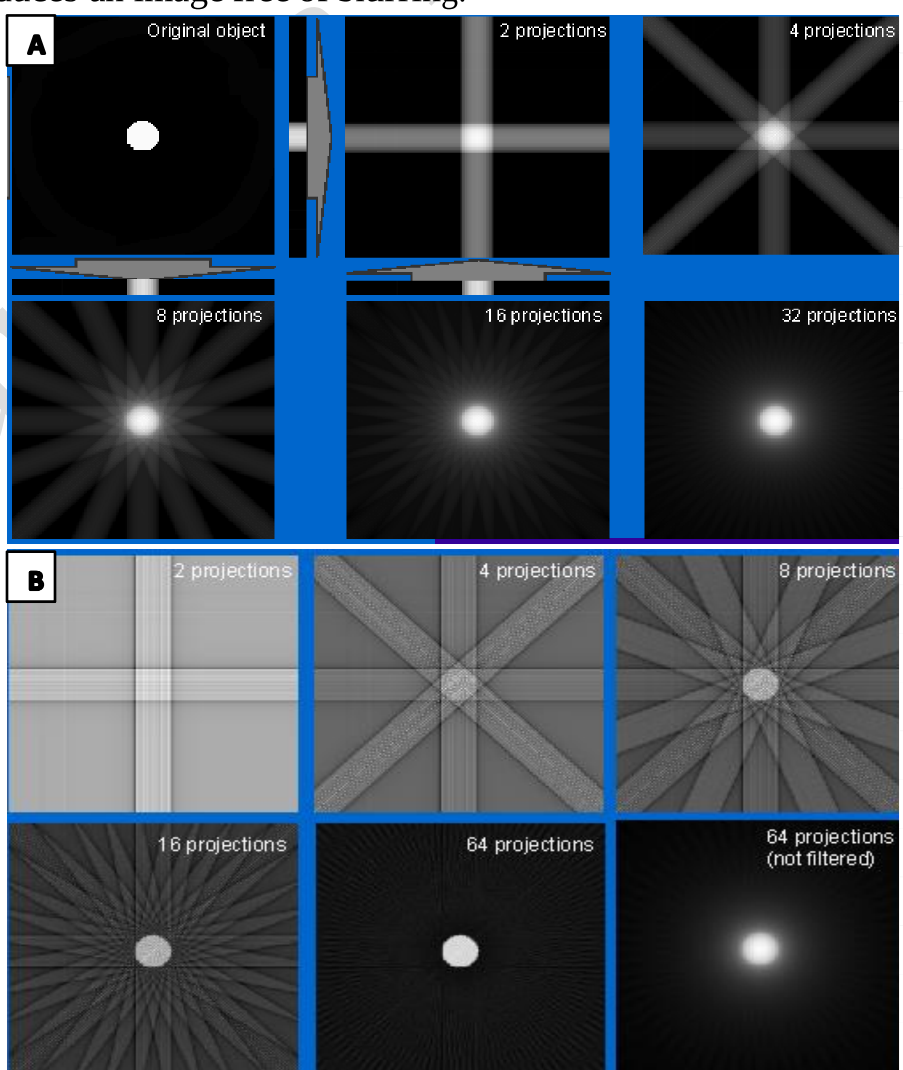


Fig. 3: Back- and filtered back techniques used in CT.
A, Back-projection results in an unsharp image.**B,** Filtered back-projection uses a digital filter to remove blurring, which produces a sharp image.

The image quality is acceptable, but not optimal and thus, its major disadvantage is its limitations in image quality due to the necessary filtering used with this technique. These filtering techniques accentuate noise and mandate the need for higher radiation doses to permit adequate image quality. The excess image noise using FBP results from the inaccuracy of several assumptions used in this technique that limit spatial resolution and lead to increased streak artifact and relatively poor low contrast detectability. FBP tends to falter in larger patients due to increased tissue attenuation and in intentional low radiation dose scanning, which is becoming more important as understanding and awareness of the effects of cumulative radiation dose are realized. However, the advantages and acceptability of FBP have traditionally limited the incentive to change reconstruction methods. However, with the increased numbers of CT scans and the advanced applications such as cardiac CT angiography, the importance of more radiation efficient reconstruction methods has been emphasized, mandating the onset of IR.

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Title: العنوان:

- **Iterative reconstruction**

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Lec. Dr. Lamyaa Fadhil Abdul Hussein

Target population: الفئة المستهدفة:

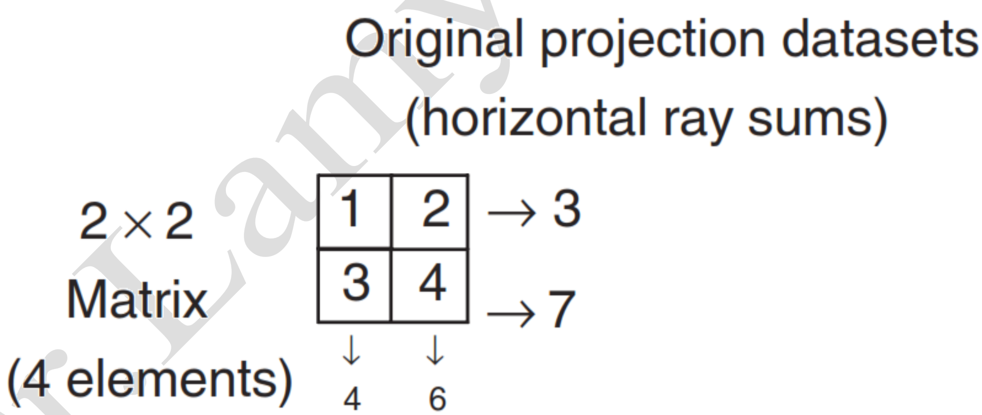
Students of second class طلبة المرحلة الثانية

- **Iterative reconstruction**

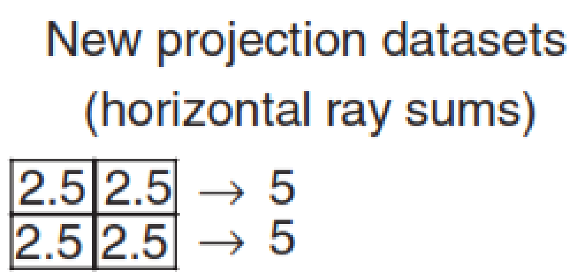
Another approach to image reconstruction is based on iterative techniques. Iteration is defined as a procedure in which repetition of a sequence of operations results in values successively closer to a desired result. Said another way, iteration is a computational, mathematical procedure in which a cycle of operations is repeated, often, to approximate the desired result more closel .

"An iterative reconstrution starts with an assumption (for example, that all points in the matrix have the same value) and compares this assumption with measured values, makes corrections to bring the two into agreement, and then repeats this process over and over until the assumed and measured values are the same or within accept- able limits".

<u>Consider the following numeric illustration:</u>



1. Initial estimate: Compute the average of four elements and assign it to each pixel, that is, 1 + 2 + 3 + 4 = 10; 10/4 = 2.5



2. First correction for error (original horizontal ray sums minus the new horizontal ray sums divided by 2) = (3 − 5)/2 and (7 − 5)/2 = −2/2 and 2/2 = −1.0 and 1.0

3. Second estimate:

(2.5 – 1) 1.5	(2.5 – 1) 1.5
(2.5 – 1) 3.5	(2.5 – 1) 3.5

4. The second correction for error (original vertical ray sums minus new vertical ray sums divided by 2) = (4 − 5)/2 and (6 − 5)/2 = −1.0/2 and +1.0/2 = −0.5 and +0.5:

(1.5 – 0.5) 1	(1.5 – 0.5) 2
(3.5 – 0.5) 3	(3.5 – 0.5) 4

1.5	1.5
3.5	3.5
↓	↓
5	5

New projection dataset
(vertical ray sums)

The final matrix solution is thus →

1	2
3	4

In this technique, repeated estimations of the x ray photon counts that would be acquired in each projection are calculated, and compares them with the actual measured counts (*forward projection*) acquired by the scanner's detector array. At each step, the ratio of estimated to actual x ray counts is used to formulate a correction factor that is used to create the next estimate (*back projecting the ratio*).

This process is repeated over and over again resulting in movement of the estimated x ray photon count distribution ever closer to the actual, measured photon count distribution, Fig. 1.

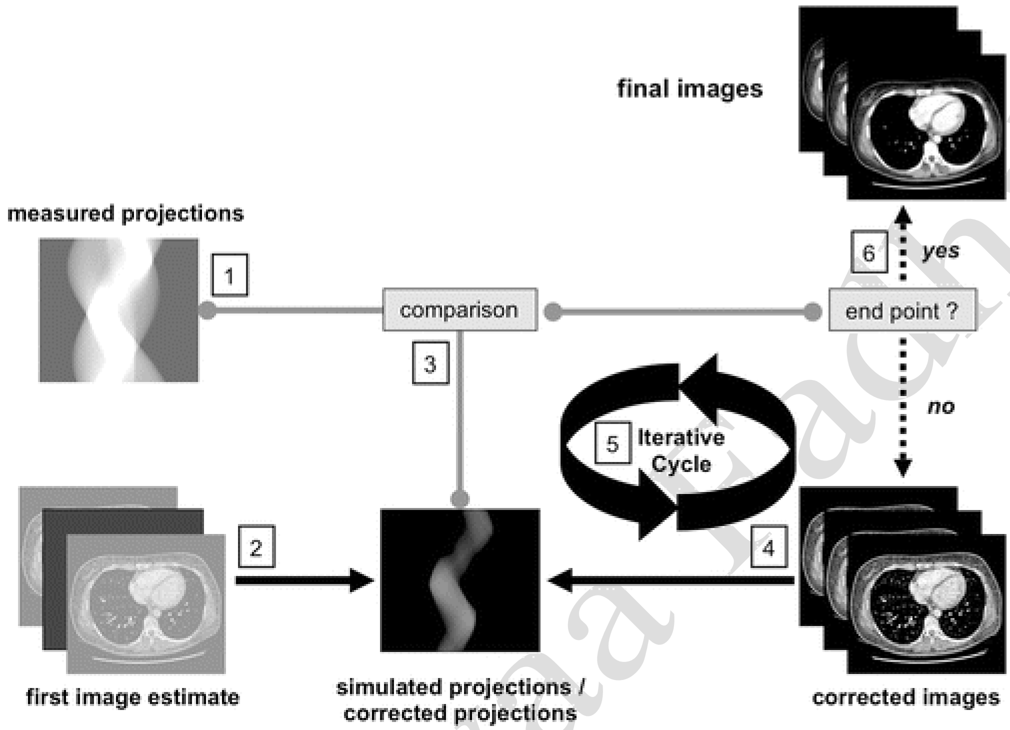


Fig. 1: Iterative reconstruction techniques used in CT.

Today, iterative reconstruction algorithms have resurfaced because of the availability of high-speed computing. The primary advantages of iterative image reconstruction algorithms are to reduce image noise and minimize the higher radiation dose inherent in the filtered back-projection algorithm.

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Radiological Techniques Department	قسم تقنيات الأشعة

المادة: تقنيات اجهزة التصويرالمقطعي المحوسب

Computed Tomography EquipmentsTechniques

Second stage/ 2nd coarse — **المرحلة: الثانية / الكورس الثاني**

Title: العنوان:

CT image quality

- **Image contrast,**
- **Spatial resolution,**

Name of the instructor: **اسم المحاضر:**

م.د.لمياء فاضل عبدالحسين

Lec. Dr. Lamyaa Fadhil Abdul Hussein

Target population: الفئة المستهدفة:

Students of second class — طلبة المرحلة الثانية

CT image quality

Because CT images are composed of discrete pixel values, image quality is somewhat easier to characterize and quantitate.

A number of methods are available for CT image quality is dependent on:

- ❖ Image contrast,
- ❖ Spatial resolution,
- ❖ Noise
- ❖ Artifacts

Depending on the diagnostic task, these factors interact to determine sensitivity; the ability to perceive low & high -contrast structures to yield a diagnostic CT image and the visibility of details.

➥ ***Image contrast***

The ability to distinguish one soft tissue from another without regard for size or shape is called **contrast resolution**. This is an area in which multislice helical CT excels. CT image contrast depends on subject contrast and display contrast display contrast. Display contrast is arbitrary and based on the windowing parameters (window level & window width selected).

As in radiography, CT subject contrast is determined by differential attenuation: that is, differences in x-ray attenuation by absorption or scattering in different types of tissue and thus resulting in differences in the intensity of the x-rays ultimately reaching the detectors. Because of the high peak kilovoltage and relatively high beam filtration (beam hardness) used in CT, most of the radiological events in CT are Compton scatter events which differ in intensity based on differences in tissue electron density (electrons/ cm^3), which in turn are due primarily to differences in physical density.

Thus, subject soft-tissue contrast in CT comes mainly from differences in physical density. That the small differences in soft-tissue density can be visualized on CT is due to the nature of the image (a 2-dimensional image of a 2-dimensional slice). However, these differences may be mapped to large differences in grey levels (grey scale) through windowing which makes CT visualization of various tissues possible.

For more explainaton:

The absorption of x-rays in tissue is characterized by the x-ray linear attenuation coefficient. The absorption of x-rays in tissue is characterized by the x-ray linear attenuation coefficient. This coefficient, as we have seen, is a function of x-ray energy and the atomic number of the tissue. In CT, absorption of x-rays by the patient is determined also by the mass density of the body part.

Consider the situation outlined in Figure (1), a fat–muscle–bone structure. Not only are the atomic numbers somewhat different (Z = 6.8, 7.4, and 13.8, respectively), but the mass densities are different (r = 0.91, 1.0, and 1.85 g/cm3, respectively). Although these differences are measurable, they are not imaged well on conventional radiography.

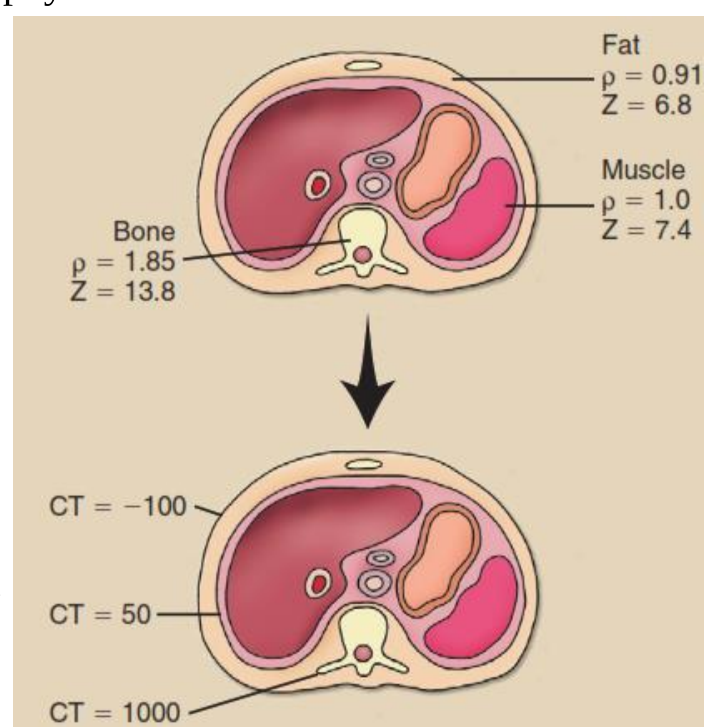

Fig. (1): No large differences are noted in mass density and effective atomic number among tissues, but the differences are greatly amplified by computed tomography imaging.

The CT imaging system is able to amplify these differences in subject contrast so the image contrast is high. The range of CT numbers for these tissues is approximately −100, 50, and 1000, respectively. This amplified contrast scale allows CT to better resolve adjacent struc- tures that are similar in composition.

The contrast resolution provided by CT is considerably better than that available in radiography princi- pally because of the scatter radiation rejection of the prepatient and predetector collimators. The ability to image low-contrast objects with CT is limited by the size and uniformity of the object and by the ***noise of the system.***

Factors influencing contrast:

- **Noise:** a higher noise will obscure any contrast between objects
- **Tube current:** a higher tube current reduces the noise in the image
- **Inherent tissue properties:** the difference in the linear attenuation coefficient of adjacent imaged objects will determine the contrast between those objects
- **Beam kilovoltage:** a higher beam energy will generally reduce the contrast between objects
- **Use of contrast media**

➥ ***CT Spatial Resolution***

Spatial resolution in CT, as in other modalities, is the ability to distinguish small, closely spaced objects on an image.

Characteristics of the CT imaging system that contribute to such image degradation include collimation, detector size , mechanical and electrical gantry control, and the reconstruction algorithm.

Factors affecting spatial resolusion

If one images a regular geometric structure that has a sharp interface, the image at the interface will be some- what blurred. The degree of blurring is a measure of the spatial resolution of the system and is controlled by a number of factors.

- ***Focal spot***

Spatial resolution is determined by x ray tube focal-spot size as well as blurring occurring in the image detector. focal-spot size is a contributor to spatial resolution (smaller focal spot size equals better spatial resolution).

- ***Detectors size***

The size of the detector measurements (referred to as aperture size and represents sampling size) and the detector spacing (spacing of measurements) are the predominant factors that determining a CT scanner's spatial resolution.

The smaller the detector measurement capability and the closer the detector spacing, the better the spatial resolution. This concept is known as ***sampling***.

Detectors must be the same size or smaller than the imaged object in order to resolve it. In addition, detectors must be close together to resolve objects that are near to each other. Further, the detectors must be properly aligned. Improper alignment may result in less resolving power than would be predicted by detector size and spacing alone.

- ***Pixel size***

Spatial resolution is a function of pixel size: The smaller the pixel size, the better is the spatial resolution. CT imaging systems allow reconstruction of images after imaging followed by postprocessing tasks; this is a powerful way to affect spatial resolution.

➤ ***Voxel size***

The displayed spatial resolution may also be affected by the image reconstruction or by the voxel size on the computer screen. For example, it is possible that the voxel size on the computer matrix is too large to resolve an object that is theoretically resolvable based on the sampling characteristics. This limitation may be overcome by reducing the scanned field of view which will have the effect of yielding smaller voxels. For example, if the matrix size is 512 voxels by 512 voxels and the scanned field of view is a 50 cm diameter, the voxel size will be 50 cm/512 pixels or approximately 0.1 cm by 0.1 cm by 0.1 cm. However, if the scanned field of view is reduced to 25 cm then the resulting voxel size will be 0.5 cm by 0.5 cm by 0.5 cm.

➤ ***Number of projections***

Larger number of projections gives finer resolution (up to a point).

➤ ***Detector slice thickness***

The above discussion applies mainly to x and y axis spatial resolution. Z axis spatial resolution (head to toe) depends on the image thickness which in turn depends on the length of the individual detector in the z axis. In addition, z axis resolution depends on the reconstruction interval (degree of overlap of z axis image slices) **Overlapping samples** .Acquiring the data using overlapping slices can improve Z-sensitivity. This is achieved by using a low spiral pitch e.g. pitch <1.

Thinner slice thicknesses also allow better spatial resolution. Anatomy that does not lie totally within a slice thickness may not be resolved, an artifact called *partial volume.*

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Computed Tomography EquipmentsTechniques

Second stage/ 2nd coarse المرحلة: الثانية / الكورس الثاني

Title: العنوان:

CT image quality

❖ **Image noise**

Name of the instructor: اسم المحاضر:

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Image noise

Noise

Noise in computed tomography is an unwanted change in pixel values in an otherwise homogeneous image. Often noise is defined loosely as the grainy appearance on cross-sectional imaging.

Noise in CT is measured via *the signal to noise ratio (SNR)*; comparing the level of desired signal (photons) to the level of background noise (pixels deviating from normal). The higher the ratio, the less noise is present in the image.

Noise in a cross-sectional image will equal a decrease in the picture quality and inadvertently will hinder the contrast resolution.

CT numbers of a particular substance such as water are not uniform but rather fluctuate. For water, the CT numbers will fluctuate around an average of 0 (by convention). These random fluctuations in the CT number of a uniform material appear as graininess on a CT image. The degree of random fluctuations depends on the number of x ray photons that contribute to the formation of a CT image. CT noise, therefore, is associated with the number of x rays contributing to each detector measurement. The more x rays used to generate an image, the smaller the amount of image noise. Thus, to understand how each CT technique factor affects image noise, one must imagine the affect of this technique on the number of x rays reaching detector to form the image.

Fators affect image noise

In radiography, image noise is related to the numbers of x-ray photons contributing to each small area of the image (e.g., to each pixel of a digital radiograph). In CT, x-rays contribute to detector measurements and not to individual pixels. CT image noise is thus associated with the number of x-rays contributing to each detector measurement.

To understand how CT technique affects noise, one should imagine how each factor in the technique affects the number of detected x-rays. Examples are as follows and its illustrates in table (1):

- tube current,
- scan time,
- slice thickness,
- tube voltage
- patient size.

Tube current in mA is directly proportional to the number of x rays reaching the detector. Therefore, increasing mA will decrease image noise.

Scan time is also directly proportional to the x ray number and thus as scan time increases, image noise decreases.

Scan time and tube current are considered together and measured as mAs (milli- amperes-sec).

Table (1) :The affect of various CT conditions on image noise

Factor	Affect on CT image noise
Increasing Tube current (mA)	Decrease
Increasing scan time	Decrease
Increasing slice thickness	Decrease
Increasing tube voltage (KeV)	Decrease
Increasing patient size	Increase

Slice thickness changes the beam width entering the detectors. Thus increasing slice thickness results in increasing the beam width which in turn increases the number of x rays proportionately. Increased slice thickness decreases image noise.

Increasing tube voltage (keV) increases the energy of the generated x rays and thus, more x rays will penetrate the patient and reach the detectors.

Increasing KeV decreases image noise. Decreasing KeV will increase image noise (less penetrated x rays) and will brighten the contrast.

Patient factors may also contribute to image noise. That is, the larger the patient, the more attenuation of x rays and thus, less of these x rays will reach the detectors. Increasing patient size results in more image noise due to less x rays reaching the detectors to form the CT image.

Figure (1) shows examples of noise in scans of uniform phantoms using standard and higher-resolution (bone) filters and with standard and very low values for mAs.

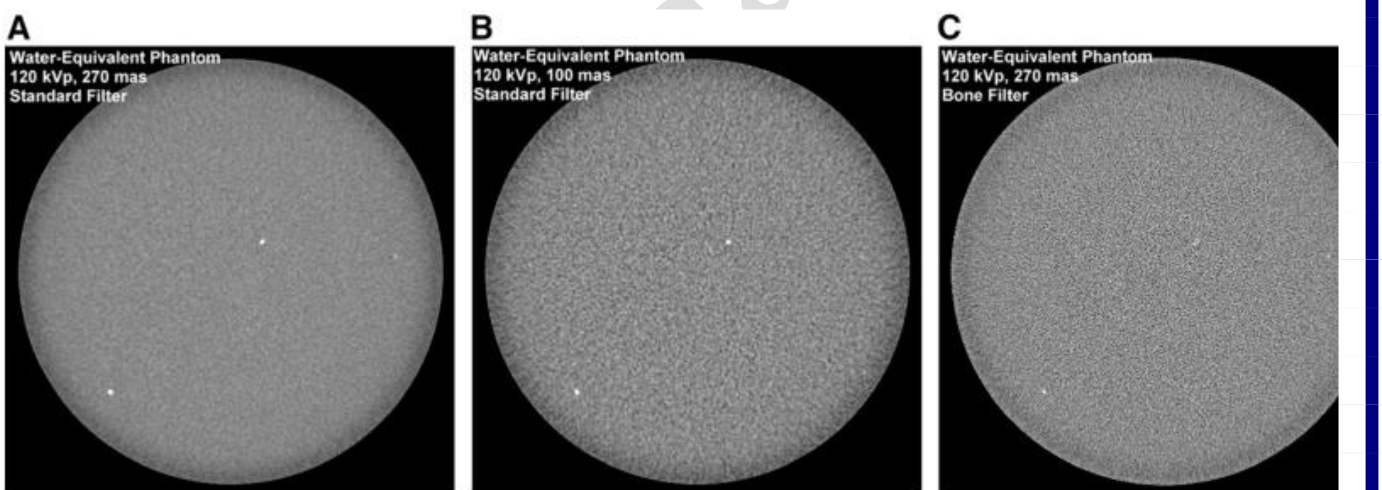

Figure (1): CT image noise depends on number of x-ray photons contributing to image. (A and B) Comparison of noise from scans using 270 mAs (typical clinical value) and 100 mAs. (C) Appearance of image noise is strongly affected by reconstruction filter; sharp filter such as bone also sharpens (enhances) appearance of noise.

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Computed Tomography EquipmentsTechniques

Second stage/ 2nd coarse | **المرحلة: الثانية / الكورس الثاني**

Title: العنوان:

CT image quality

- ❖ **Image artifacts**
- ❖ **Types , causes & correction techniques**

Name of the instructor: **اسم المحاضر:**

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Target population: الفئة المستهدفة:

Students of second class | طلبة المرحلة الثانية

CT image quality

Image artifacts

Artifacts may be defined as any structure that is seen on an image but is not representative of the actual anatomy.

Artifacts can degrade image quality, affect the perceptibility of detail, or even lead to misdiagnosis. They can cause serious problems for the radiologist who has to provide a diagnosis from images obtained by the CT scanner. Therefore it is mandatory that the technologist understand the nature of artifacts in CT.

In general, an **artifact** is "a distortion or error in an image that is unrelated to the subject being studied". Specifically, a *CT image artifact* is defined as *"any discrepancy between the reconstructed CT numbers in the image and the true attenuation coefficients of the object"*.

Because CT numbers represent gray shades in the image, incorrect measurements will produce incorrect CT numbers that do not represent the attenuation coefficients of the object. These errors result in various artifacts that affect the appearance of the CT image.

Most types of CT artifacts fall into 3 categories:

- Streak artifacts
- Shading artifacts
- Ring artifacts

Streak artifacts may appear as intense straight lines across an image, which may be caused by improper sampling of the data, motion, metal, beam hardening, noise, spiral/ helical scanning, and mechanical failure or imperfections.

These discrepancies are enhanced by the convolution process and manifested into lines during the back-projection, as shown in Figure (1,A).

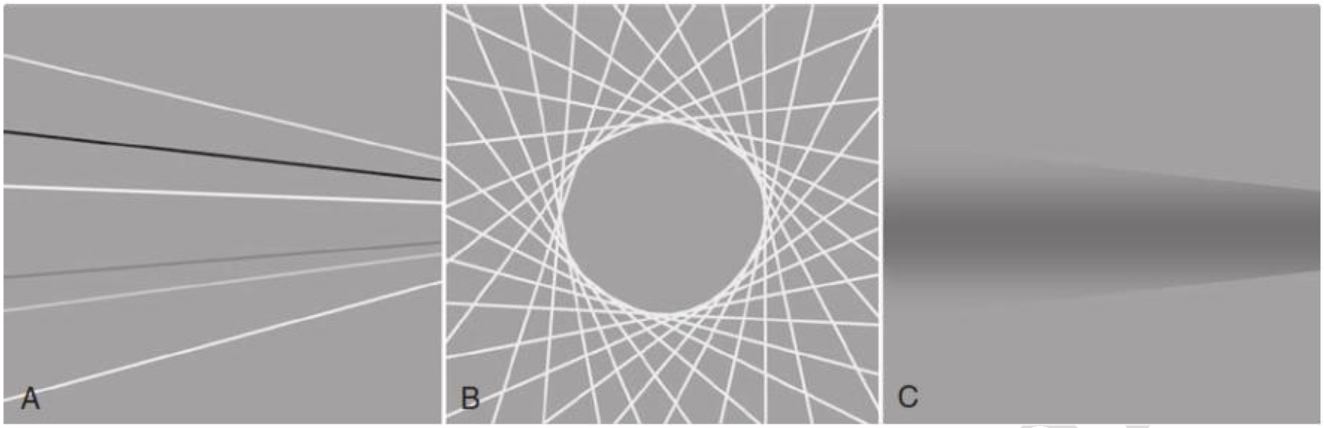

Fig. (1): Different appearances of artifacts. **A,** Streak. **B,** Ring. **C,** shading.

Streak artifacts may occur in all scanners. Although arising for many reasons, most are due to inconsistent or bad detector measurements. Factors causing inconsistencies include motion (anatomy in different locations during different parts of the scan), insufficient x-ray intensity (leading to high random errors), and malfunctions (tube arcing or system misalignment).

An inconsistency due to partial-volume effects is illustrated in Figure **(2)**. During a 360 axial scan, the same ray (or nearly the same ray) is sampled twice, but with x-rays traveling in opposite directions. Because of beam divergence, however, the cone-shaped x-ray beam samples slightly different volumes in each direction. A small structure, such as the edge of a bone, may partially extend into the volume so as to attenuate the beam traveling in one direction (say, downward when the tube is above) but may be missed when the beam is coming from the opposite direction (upward, when the tube is underneath). The two measurements of the same ray path are thus inconsistent and will lead to an image streak.

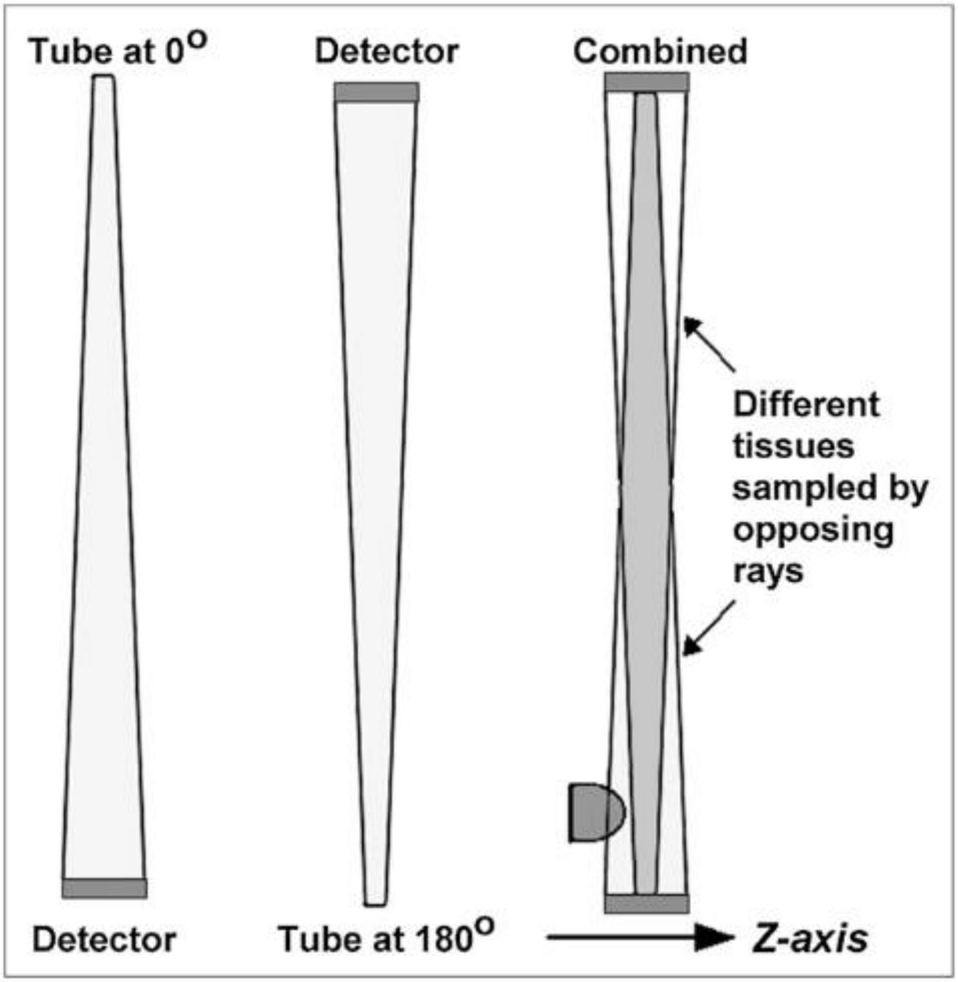

Figure (2): Partial-volume streaks are caused by opposing x-ray beams, which nominally pass through the same voxels but actually sample slightly different cone-shaped tissue volumes as a result of beam divergence. Small structure, such as piece of bone, is detected by beam from one direction but is missed by opposing beam. Resulting inconsistency leads to streak artifact.

Shading artifacts often appear near objects of high densities and can be caused by beam hardening, partial volume averaging, spiral/helical scanning, scatter radiation, off-focal radiation, and incomplete projections, as shown in Figure 2, C.

The most common type of ***shading artifact*** is beam- hardening effects. Beam-hardening artifacts are actually present on all CT images to some extent and are due to imperfect beam-hardening correction. They appear as non- uniformities in the CT numbers of a uniform material, such as CT numbers that are lower at the center of a uniform phantom than at the periphery. Such nonuniformities are generally quite small (< 5 HU) and not apparent unless one is viewing a scan of a uniform phantom with a very narrow window. Occasionally, however, a larger amount of hardening occurs when the scan is passing through thick regions of bone orthrough contrast medium. In that case, regions of hypointensity (i.e., CT numbers that are lower than expected) may appear downstream along the paths of rays that have been overly hardened . Scatter can also cause shading artifacts, although these are uncommon in most modern scanners.

Ring or partial ring (arc) artifacts are produced when the projection readings of a single channel or a group of channels consistently deviate from the truth. They can be the result of defective detector cells or DAS channels, deficiencies in system calibration, or a sub- optimal image-generation process. This is predominately a third-generation CT scanner phenomenon. Because a detector channel reading is always mapped to a straight line that is at a fixed distance to the iso- center of the system, any such inaccuracies in measurements from a particular detector occurring during a scan (or part of a scan) are backprojected along the ray paths measured by that detector. These inaccuracies contribute only slightly to pixels over most of the image (because several hundred backprojected values contribute to each pixel) but tend to reinforce along a ring of radius d, where several such rays intersect.a defective reading forms a ring pattern during the back-projection process, as illustrated in Figure 2, B.

Ring artifacts are usually readily recognizable by software ring- correction algorithms and thus can be removed from the image. Small-radius rings (i.e., near the center of rotation) or arcs of small angular extent may not be recognized as artifacts and thus wind up in the image. In practice, third- generation scanners are sensitive to detector inaccuracies, which, without corrective algorithms, would be visible on most or all CT images.

Causes of artifacts

CT artifacts are common and can occur for various reasons. Knowledge of these artifacts is important because they can mimic pathology (e.g. partial volume artifact) or can degrade image quality to non-diagnostic levels.

CT artifacts can be classified according to the underlying cause of the artifact.

Patient-based artifacts

- ✓ motion artifact
- ✓ transient interruption of contrast
- ✓ clothing artifact

Physics-based artifacts

- beam hardening
    - ✓ cupping artifact
    - ✓ streak and dark bands
    - ✓ metal artifact / high-density foreign material artifact
- partial volume averaging
- photon starvation
- aliasing
- truncation artifact

Hardware-based artifacts

- ring artifact
- tube arcing
- out of field artifact
- air bubble artifact
- helical and multichannel artifact
    - ✓ windmill artifact
    - ✓ cone beam effect

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Computed Tomography EquipmentsTechniques

Second stage/ 2nd coarse **المرحلة: الثانية / الكورس الثاني**

Title: العنوان:

Positron Emission Tomography/ PET-CT

Single-photon emission computed tomography (SPECT)

Name of the instructor: **اسم المحاضر:**

م.د.لمياء فاضل عبدالحسين

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Target population: الفئة المستهدفة:

Students of second class طلبة المرحلة الثانية

Positron Emission Tomography/ PET-CT

Positron emission tomography, also called PET imaging or a PET scan, is a type of nuclear medicine imaging, of dual-modality imaging that utilizes the advantages of both positron emission tomography (PET) and computed tomography (CT).

Nuclear medicine uses small amounts of radioactive material called *radiotracers*, to diagnose, evaluate, and treat various diseases. Radiotracers are molecules linked to, or "labeled" with, a small amount of radioactive material. They accumulate in tumors or regions of inflammation. They can also bind to specific proteins in the body. The most common radiotracer is *2-[F-18]fluoro-2-deoxy-D-glucose (FDG)*, a molecule similar to glucose. Fluorine-18 is an unstable radioisotope and has a half-life of approximately 110 minutes.

Cancer cells are more metabolically active and may absorb glucose at a higher rate. It accumulates in the area under examination. A special camera detects gamma ray emissions from the radiotracer. The camera and a computer produce pictures and supply molecular information.CT imaging uses special x-ray equipment to produce multiple images of the inside of the body. A radiologist views and interprets these images on a computer monitor. CT imaging provides excellent anatomic information.

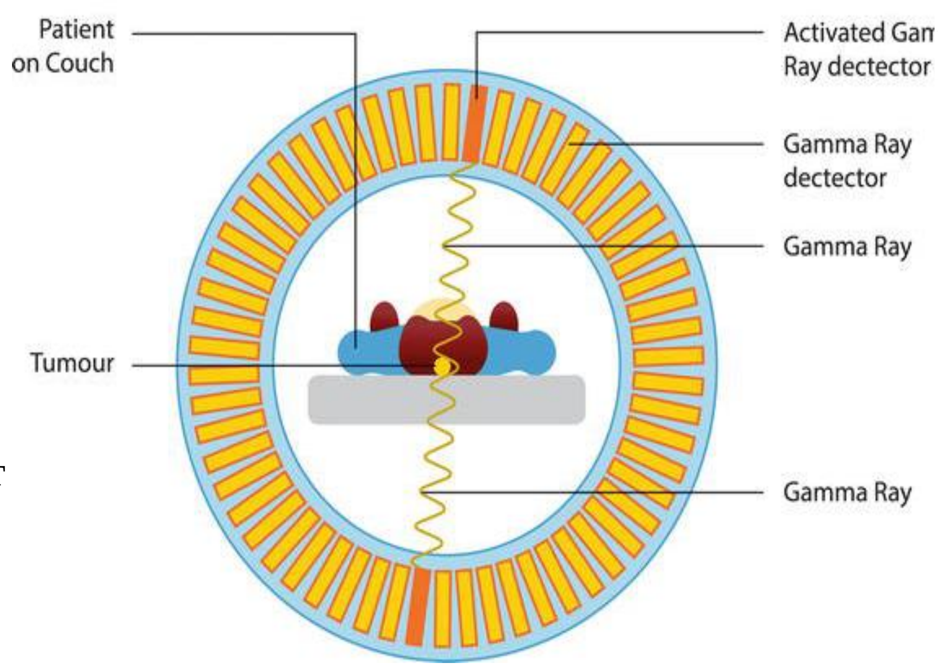

Figure1: the principle of PET-CT

The common uses of PET-CT

- ❖ detect cancer and/or make a diagnosis.
- ❖ determine whether a cancer has spread in the body.
- ❖ staging of cancer which potentially can be treated radically
- ❖ establish baseline staging before commencing treatment
- ❖ determine if a cancer has returned after treatment.
- ❖ evaluate prognosis.
- ❖ assess tissue metabolism and viability.
- ❖ map normal human brain and heart function.
- ❖ assessing response to therapy
- ❖ evaluation of suspected disease recurrence, relapse and/or residual disease

The procedure of PET-CT

Ordinary x-ray exams pass x-rays through the body to create an image. The radioactive materials (F-18 fluorodeoxyglucose) injected the bloodstream intravenously, or may swallow it or inhale it as a gas.

The material accumulates in the area under examination (tumor cells, that have a high metabolic rate), where it gives off gamma rays. Special cameras detect this energy and, with the help of a computer, create pictures that detail how the organs and tissues look and function.

Unlike other imaging techniques, nuclear medicine focuses on processes within the body. These include rates of metabolism or levels of various other chemical activities. Areas of greater intensity are called “hot spots.” These may show large concentrations of the radiotracer and where there is a high level of chemical or metabolic activity. Less intense areas, or “cold spots,” indicate a smaller concentration of radiotracer and less activity.

The radioactive materials detection

The positron-emitting isotope administered to the patient undergoes β^+ decay in the body, with a proton being converted to a neutron, a positron (the antiparticle of the electron, sometimes referred to as a β^+ particle), and a neutrino. The positron travels a short distance and annihilates with an electron. The annihilation reaction results in the formation of two high energy photons which travel in diametrically opposite directions. Each photon has an energy of 511 keV. Two detectors at opposite ends facing each other detect these two photons traveling in opposite directions, and the radioactivity is localized somewhere along a line between the two detectors. This is referred to as the line of response, fig.(2).

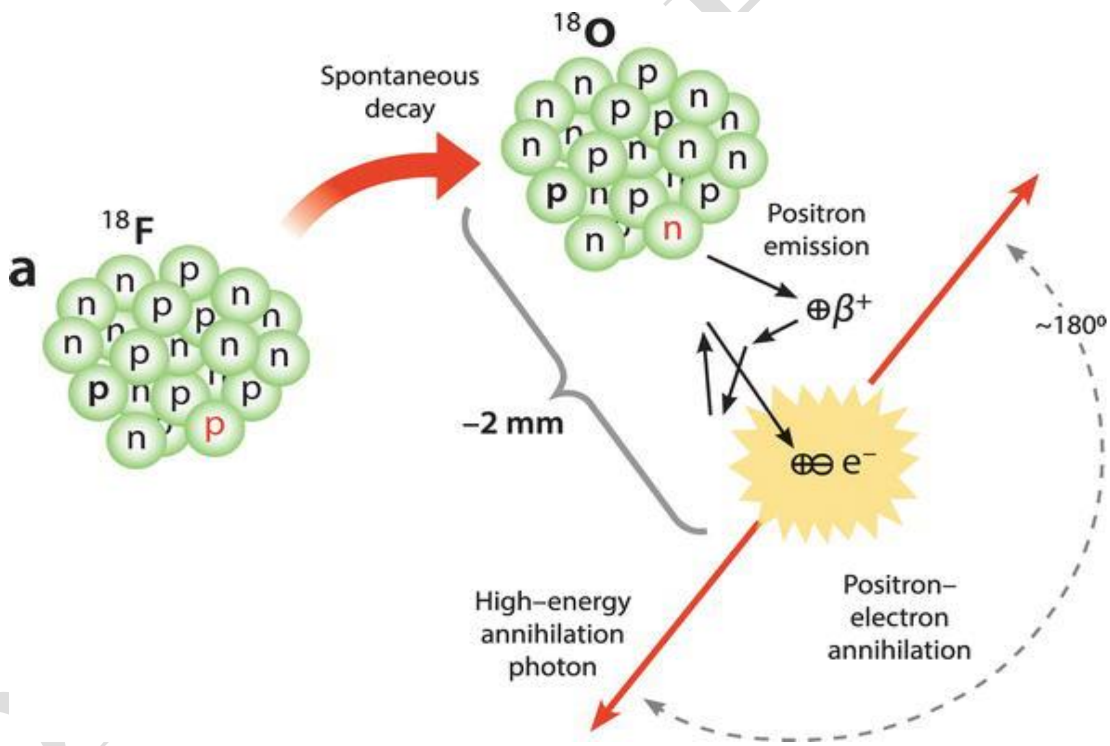

Figure 2. Beta decay causing the positron electron annihilation at 180 degree.

PET-scanner

These scanners are made up of the various many small detectors which are usually placed in adjacent rings around the patient. The clinical PET system was having a ring diameter of 70–100 cm with the extent of 10–25 cm and made up to 25,000 detectors. The single PET detector is made up of the very high-density scintillator crystal which are capable of converting the photons striking on the detector into light. The crystals used in the construction of PET scanner detectors are called scintillators. Photons interact in the crystal, resulting in the emission of light, which is collected by an array of photomultiplier tubes (PMT), where the light will be converted into an amplified electric signal (Figure 3).

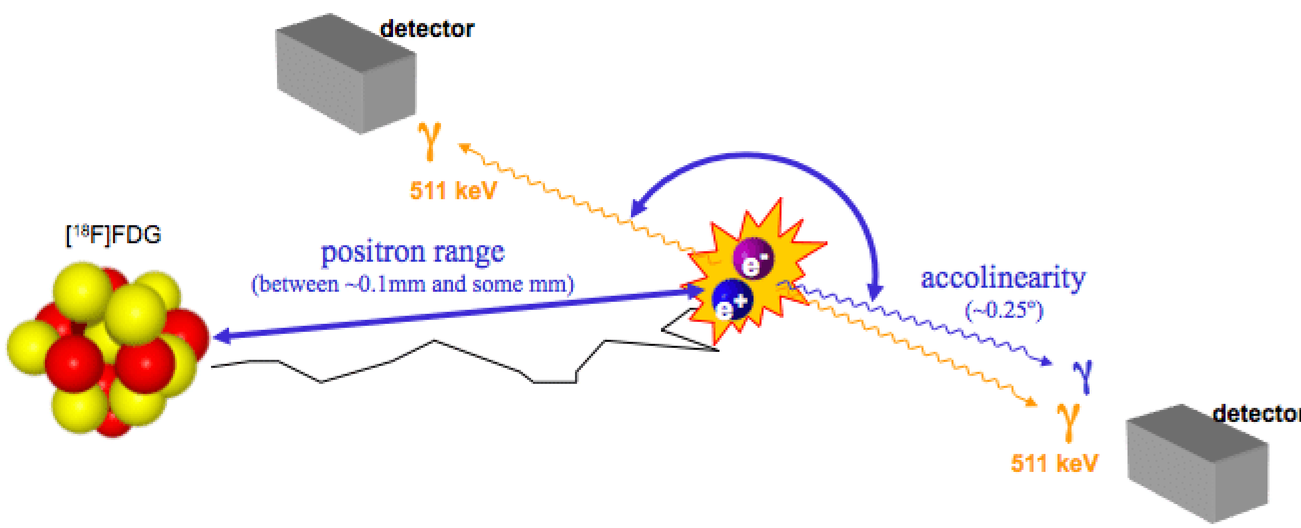

Figure 3: the principle of PET-CT

Single photon emission computed tomography (**SPECT**) is a three-dimensional nuclear medicine imaging technique combining the information gained from scintigraphy with that of computed tomography. This allows the distribution of the radionuclide to be displayed in a three-dimensional manner offering better detail, contrast and spatial information than planar nuclear imaging alone. It is used to help diagnose seizures, strokes, stress fractures, infections, and tumors in the spine. shows how blood flows into and within tissues and organs.

Design

SPECT machines combine an array of gamma cameras (ranging from one to four cameras) which rotate around the patient on a gantry. SPECT may be also combined with a separate CT machine in a form of hybrid imaging; single photon emission computed tomography-computerized tomography (SPECT-CT) mainly for the purposes of attenuation correction and anatomical localization .

Principle

Gamma cameras rotate around the patient providing spatial information on the distribution of the radionuclide within tissues. The use of multiple gamma cameras increases detector efficiency and spatial resolution. The projection data obtained from the cameras are then reconstructed into three-dimensional images usually in axial slices. When SPECT-CT is used, attenuation correction and higher resolution anatomical localization can be achieved

SPECT vs PET

Single photon emission computed tomography (SPECT) and positron emission tomography (PET) are nuclear medicine imaging techniques which provide metabolic and functional information unlike CT and MRI.

They have been combined with CT and MRI to provide detailed anatomical and metabolic information.

Positron emission tomography (PET):

- uses positron emitting radioisotope (tracer)
    - F-18 fluorodeoxyglucose (FDG)
- gives better contrast and spatial resolution (cf. SPECT)
- has a ring of multiple detectors

Single-photon emission computed tomography (SPECT):

- uses gamma emitting radioisotope (tracer):
    - technetium-99m
    - iodine-123
    - iodine-131
- gives poorer contrast and spatial resolution (cf. PET)
- usually one large crystal based detector

Middle Technical University (MTU) الجامعة التقنية الوسطى
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Radiological Techniques Department قسم تقنيات الأشعة

المادة: تقنيات اجهزة التصويرالمقطعي المحوسب

Computed Tomography EquipmentsTechniques

Second stage/ 2nd coarse **المرحلة: الثانية / الكورس الثاني**

Title: العنوان:

Advanced technical CT applications

- **CT Angiography**
- **Cardiac CT imaging**
- **CT flouroscopy**

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Target population: الفئة المستهدفة:

Students of second class طلبة المرحلة الثانية

CT Angiography

CT angiography is defined as CT imaging of blood vessels opacified by **contrast media**. During contrast injection, the entire area of interest is scanned with spiral/helical CT and images are recorded when vessels are fully opacified to show arterial or venous phases of enhancement.

CT angiography uses 3D imaging principles to display images of the vasculature through intravenous injection of contrast media compared with those of intra-arterial angiograms. Four essential elements are patient preparation**;** selection of acquisition parameters (total spiral/helical scan time, slice thickness, table speed) to optimize the imaging process; contrast media injection; and postprocessing techniques and visualization tools such as algorithms to display 3D images, multiplanar reconstruction, maximum intensity projection.

Cardiac CT Imaging

To image the beating heart with the goal of reducing motion artifacts and a loss of both spatial and contrast resolution, fast CT scanners such as the EBCT scanner were introduced to overcome these problems and produce good diagnostic images of the heart. Alternatively, the patient's electrocardiogram (ECG) is used to provide prospective imaging, after it is recorded at the same time with the scanning with stop and go scanners. Subsequently, retrospective imaging has been developed where the ECG is correlated with image reconstruction in spiral/helical CT scanning.

The recent technical developments in MSCT scanners and the introduction of the DSCT scanner open up a whole new avenue for successful imaging of the heart with excellent image quality based on meeting several technical requirements. These include **low-contrast resolution** to visualize small differences in tissue contrast, high-contrast resolution (spatial resolution) to visualize small structures in the anatomy scanned, to image fast-moving objects to reduce motion artifacts. These have all been made possible by fast data acquisition and dedicated reconstruction algorithms, such as the segmented (multiple) algorithms that “allow for merging synchronized transmission data from successive heart cycles.

So, *cardiac CT* is routinely performed to gain knowledge about cardiac or coronary anatomy, to detect or diagnose coronary artery disease (CAD), to evaluate patency of coronary artery bypass grafts or implanted coronary stents.

Artifacts

Several artifacts can potentially occur and include:

- motion artifact
    - cardiac motion
    - respiratory motion
    - gross patient motion
- partial volume averaging
- beam hardening
- metal or streak artifact
- slab or banding artifacts
- poor contrast enhancement
- artifacts from overlapping structures

CT Fluoroscopy

CT fluoroscopy, or continuous imaging, depends on spiral/helical data acquisition methods, high-speed processing, and a fast image-processing algorithm for image reconstruction.

In conventional CT, the time lag between data acquisition and image reconstruction makes realtime display of images impossible. CT fluoroscopy allows for the reconstruction and display of images in real time with variable frame rates.

CT fluoroscopy is based on three advances in CT technology:

(1) fast, continuous scanning made possible by spiral/helical scanning principles.

(2) fast image reconstruction made possible by spe- cial **hardware** performing quick calculations and a new image reconstruction algorithm.

(3) continuous image display by use of cine mode at frame rates of two to eight images per second.

Other support tools were developed to facilitate **interventional procedures** in CT fluoroscopy. One such tool, the Fluoro Assisted Computed Tomography System, uses a unique flat-panel amorphous silicon digital detector coupled with an x-ray tube by a C-arm, which is a part of the CT gantry.

References

1. Stewart Carlyle Bushong, ***"Radiologic Science for Technologists Physics, Biology, and Protection"*** Elsevier, Inc. , 7th edition, 2017.

2. Chris Guy & Dominic ffytche, ***"An Introduction to The Principles of Medical Imaging"*** , Imperial College Press, 2005.

3. Perry Sprawls, ***"Physical principles of medical imaging"***, 2nd Edition 1996.

4. J. Hsieh, ***"Computed Tomography: Principles, Design, Artifacts, and Recent Advances"***, 2nd ed. Wiley Inter-science, Bellingham, Washington, USA, (2009)

5. Lee W. Goldman, "***Principles of CT and CT Technology"*** , Journal of Nuclear Medicine Technology September 2007, 35 (3) 115-128.