▸ ***Seventh Generation (MS/MD CT)***

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

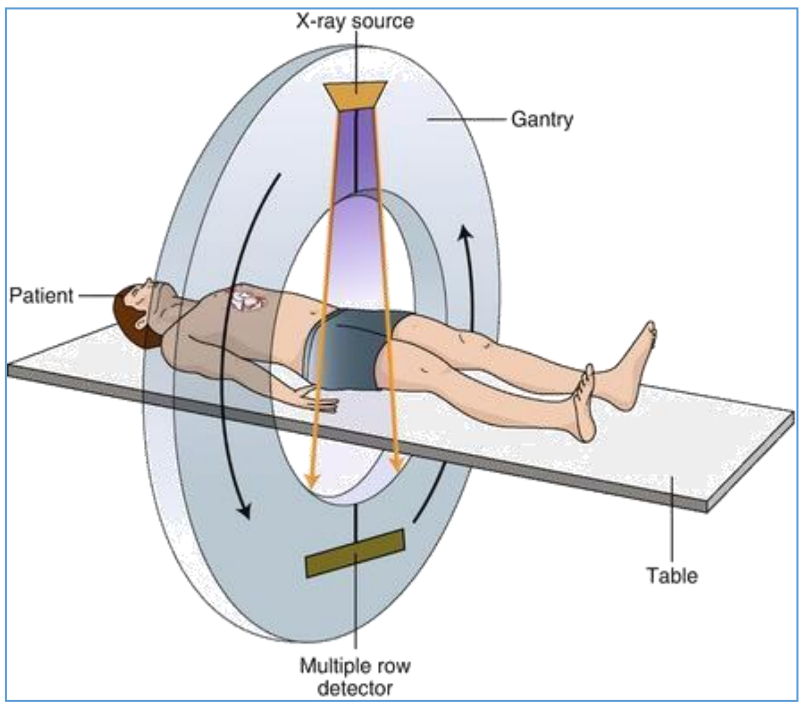


Figure (1): A cartoon depiction of a typical MDCT.

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.

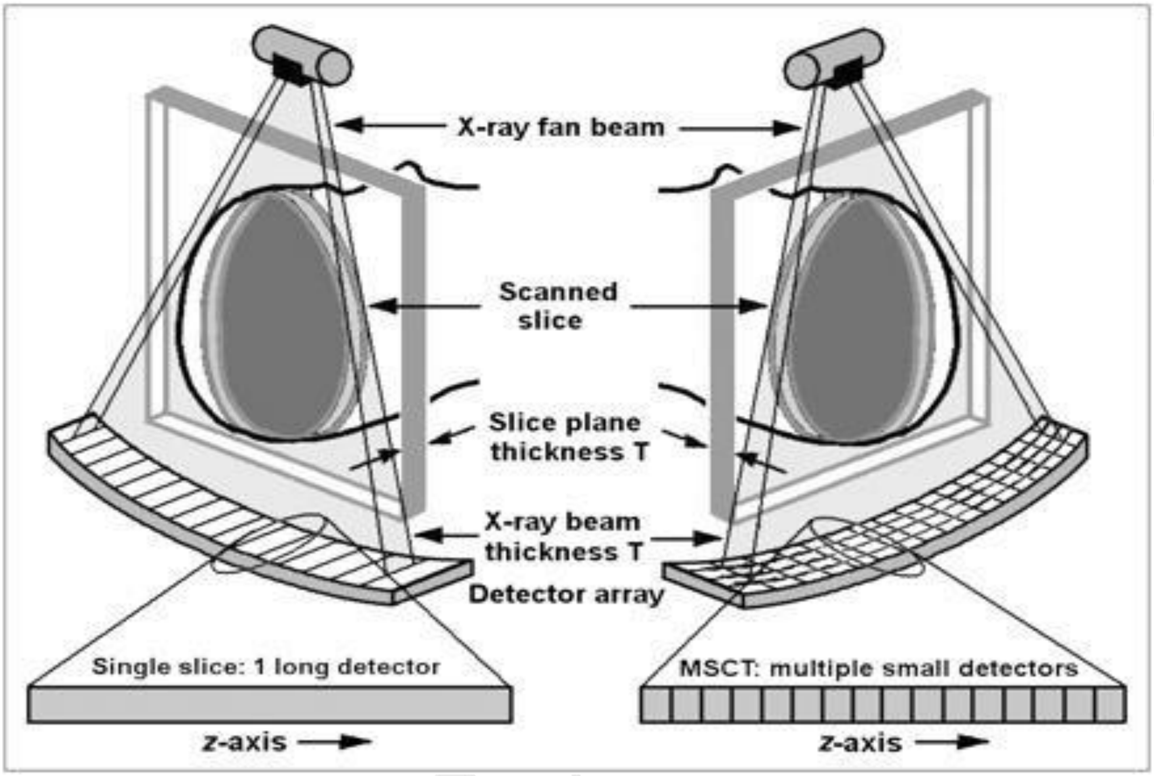


Figure. 2. A cartoon depicting a single slice scanner and a multislice scanner

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.

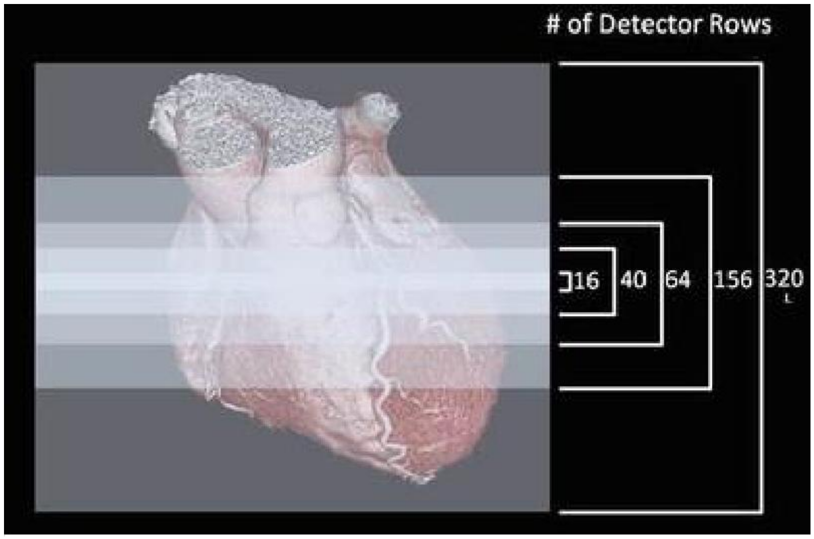


Figure (3): An artificial representation of the meaning of slab thickness or z axis coverage. Z axis coverage is directly proportional to detector row number

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.

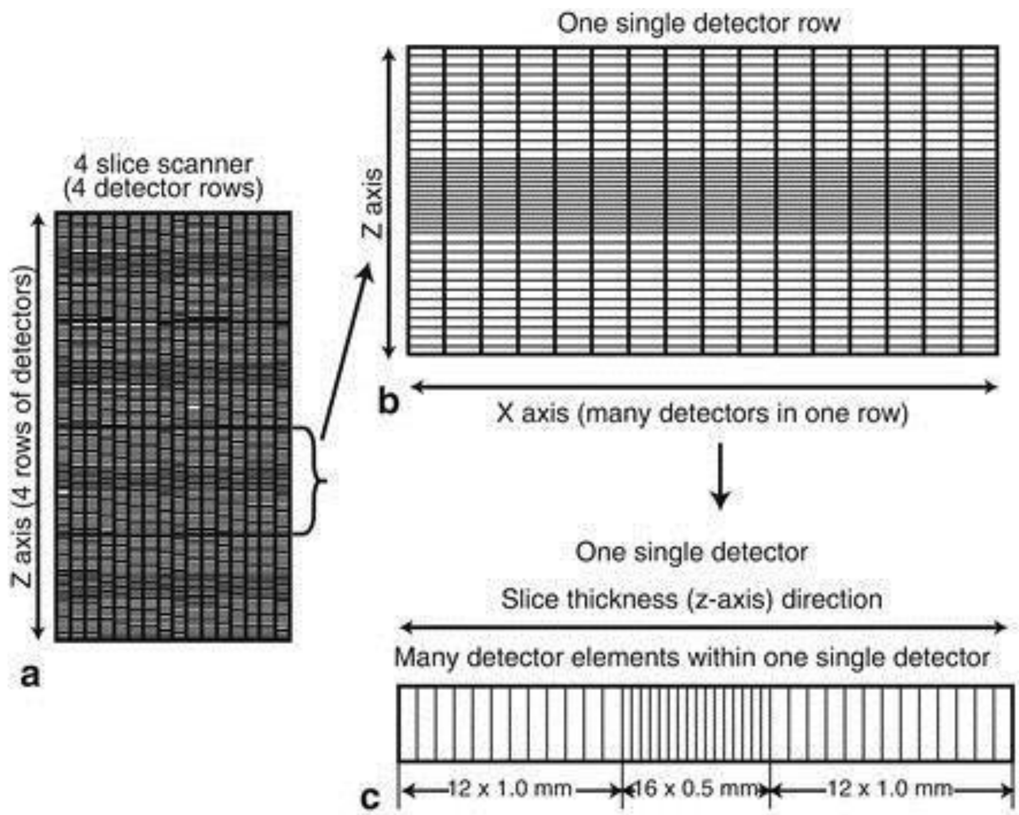


Figure. 4 Cartoon depicting a particular detector array configuration. Panel (**a**) depicts a four slice scanner (four detector rows). Panel (**b**) illustrates one single detector row. Within each row, there are multiple single detectors. There may be as man as 800 detectors per row. Panel (**c**) 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).

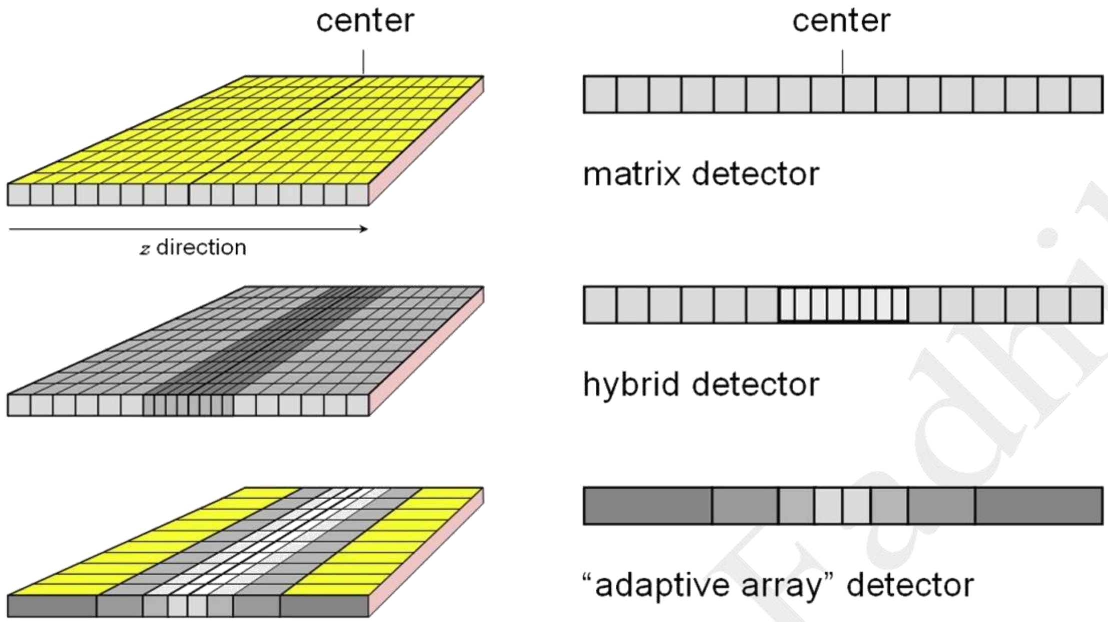


Figure. 5: types of detector arrays in MSCT

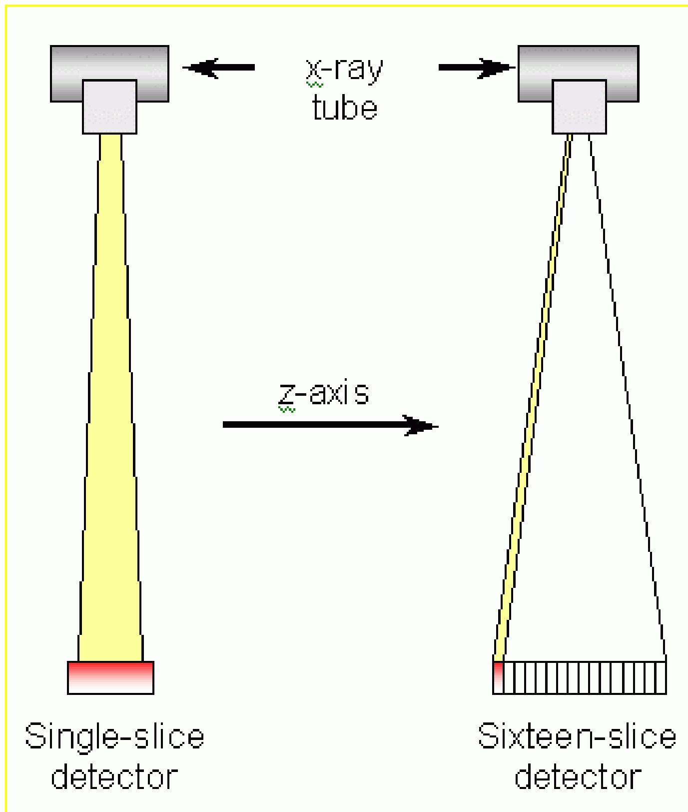
Two significant other differences exist between SSCT and MSCT:

- 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.



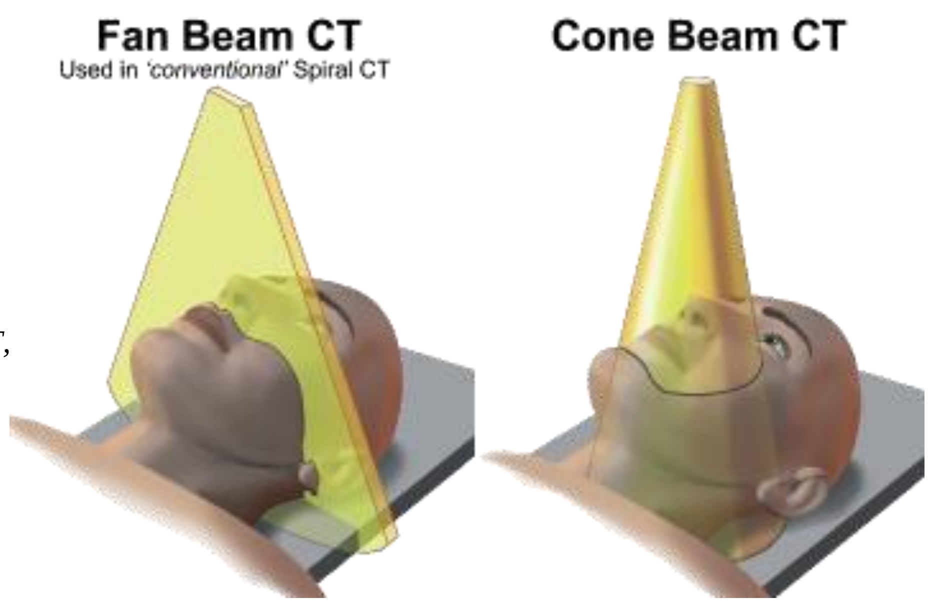


Figure. 6: Cone beam in *MSCT*, in comparison with fan beam shape in *SSCT*.

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:

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

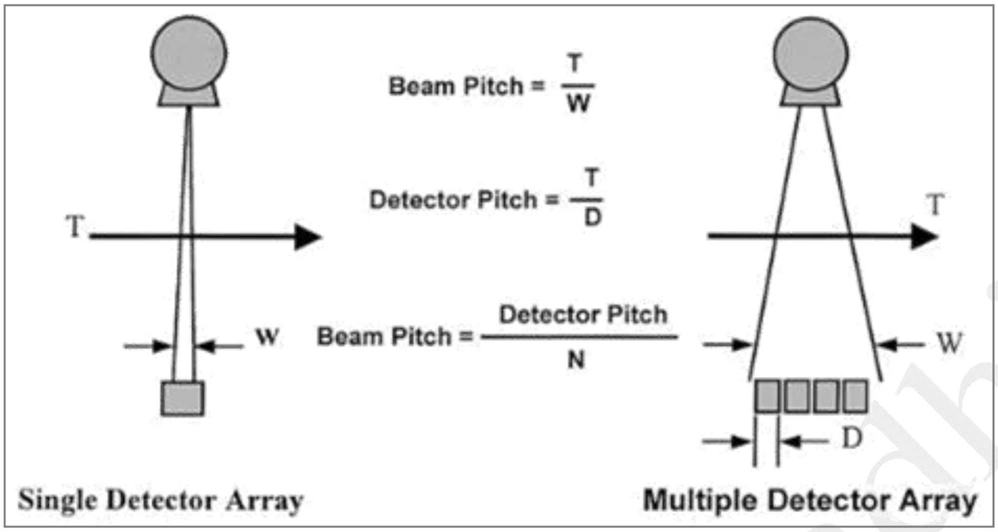


Figure 7. The diagram shows the concepts of beam pitch and detector pitch. Beam pitch is consistent with the previous notion of pitch used in single-row detector helical CT and works well for multiple-row detector CT scanners.

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

Region Scanned	Distance (cm)	Section Thickness (mm)	Scanning Time (sec) by Scanner: 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.

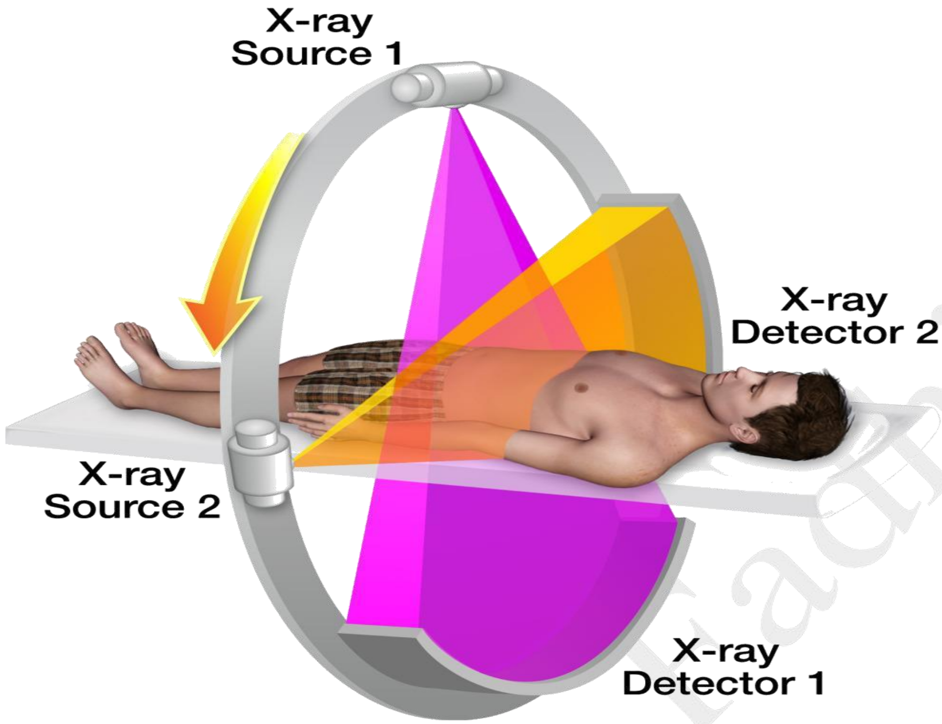


Fig. 8: Dual Source CT equipped with two data measurement systems (tubes and detectors)

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.