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College of Engineering & Technology

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Lecture Title: Data Acquisition, part 2.

Lecture Four

Data Acquisition, part 2

Scanner Generation

The configuration of the x-ray tube to the detectors determines scanner generation. The first system produced by the now defunct EMI medical division had a design that is referred to as first generation. A thin x-ray beam passed linearly over the patient, and a single detector followed on the opposite side of the patient. The tube and detector were then rotated slightly, and the process was repeated until a 180° arc was covered. Scan times were very long. This design is no longer in use.

As new developments in scanning occurred, each new tube-detector design was referred to by a consecutive generation number. The second-generation design is one in which the x-ray beam also passed linearly across the patient before rotating. However, a fan-shaped x-ray beam was used, rather than the thin beam used with first-generation designs. Only part of the field of view could be covered with this fan beam. A detector array was also incorporated in the second-generation design. Although scan times were shorter than that of the original design, they were still very long. This type of design is also no longer used.

The next advance in CT technology brought the ***third-generation*** design. This design consists of a detector array and an x-ray tube that produces a fan-shaped beam that covered the entire field of view and a detector array (Fig. 2-7).

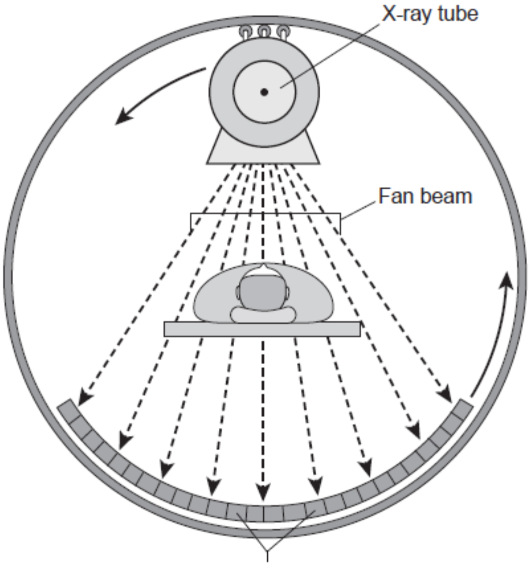


FIGURE 2-7 A third-generation scanner design is one in which the x-ray tube is placed opposite the detector array. Both the tube and the detector move in a circle within the gantry.

Reference detectors are typically located at either end of the detector array to measure the unattenuated x-ray beam. The third-generation design made it no longer necessary to translate the beam and detector as both could move in a circle within the gantry. The rotating detector design allows all of the readings that make up a view to be recorded instantaneously and simultaneously. This greatly reduced scan times and helped to reduce artifact resulting from patient motion. An advantage of the third-generation system is that the tube is directly focused on the detector array (Fig. 2-8).

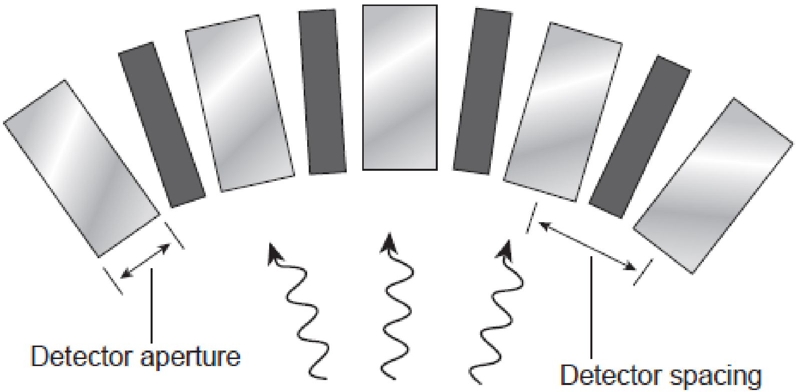


FIGURE 2-8 Third-generation systems allow the x-rays to be focused directly on the detector bank, which reduces the amount of scatter that reaches the detectors.

The fixed relationship between the x-ray source and the detectors allow the beam to be highly collimated, which greatly reduces scatter radiation, thereby improving image quality. A disadvantage of the third-generation design (as compared with the fourth-generation design) is the more frequent occurrence of ring artifacts. Because the same bank of detectors is used repeatedly, even a very small misalignment of a single detector will result in visible ring artifact. Third-generation systems are sometimes referred to as rotate-rotate scanners. The third-generation design is the most widely used configuration in the industry today.

Fourth-generation scanners use a detector array that is fixed in a 360° circle within the gantry. The tube rotates within the fixed detector array and produces a fan-shaped beam (Fig. 2-9).

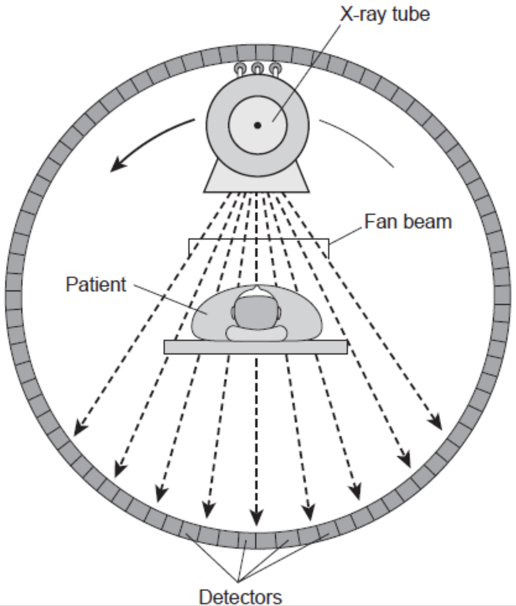


FIGURE 2-9 A fourth-generation scanner design uses a detector array that is fixed in a 360° circle within the gantry. The tube rotates within the gantry.

Although many more detector elements are included in this design, the number of detectors in use at any one time is controlled by the width of the beam. In the stationary detector design, the readings that make up a view are recorded consecutively during approximately one-fifth of the scan time. Because the emerging beam does not strike the detectors at exactly the same time, motion artifacts are more of a problem.

Fourth generation systems often use overscans to address this problem. An overscan is a tube arc greater than 360°. The use of an overscan technique will increase the radiation dose to the patient. In addition, because the tube is closer to the patient, the same milliampere-seconds (mAs) and kilovolt-peak (kVp) setting will produce a higher dose when a fourth- generation system is used (as compared with the same settings used in a third-generation system).

However, because the x-ray source is closer to the patient, techniques necessary to produce an adequate image are generally somewhat lower than that used on a third-generation system. Fourth-generation scanners may also be called rotate-only systems. Many variations of these basic designs have been introduced and then abandoned. The only other design currently in use is called electron beam imaging, also referred to as EBCT or ultrafast CT. It differs from conventional CT in a number of ways. This system, which was originally produced by Imatron, uses a large electron gun as its x-ray beam source. A massive anode target is placed in a semicircular ring around the patient. Neither the x-ray beam source nor the detectors move, and the scan can be acquired in a short time (Fig. 2-10). Invented in the 1980s, its superior speed compared with traditional CT scanners of the time made it particularly suited to cardiac imaging. However, shortfalls in spatial resolution kept EBCT from use in routine imaging, dramatically limiting the technology's clinical versatility. Additional drawbacks were high cost and difficulties obtaining insurance reimbursement.

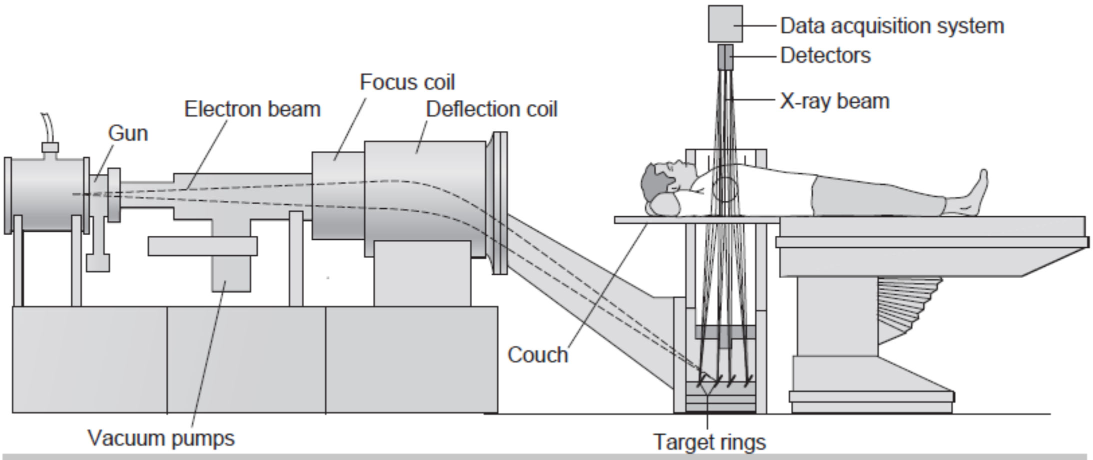
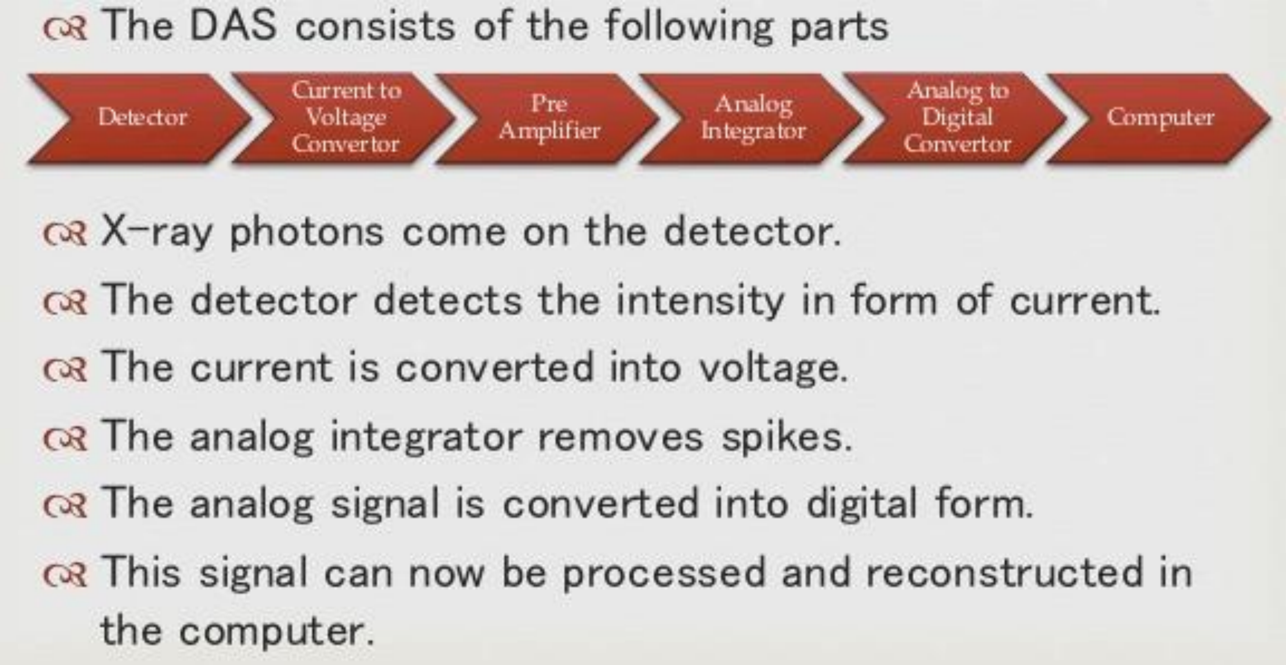


FIGURE 2-10 The design of EBCT scanners is fundamentally different from the design of other CT systems.

Detector Electronics

X-ray photons that strike the detector must be measured, converted to a digital signal, and sent to the computer. This is accomplished by the data-acquisition system (DAS), which is positioned within the gantry near the detectors. Signals emitted from the detectors are analog (electric), whereas computers require digital signals. Therefore, one of the tasks of the DAS is to convert the analog signal to a digital format. This is accomplished with analog-to-digital converter or ADC.

To measure the x-ray photons that have penetrated the patient, the detectors are sampled many times, as many as 1,000 times per second by the DAS. The number of samples taken per second from the continuous signal emitted from the detector is known as the sampling rate, sample rate, or sampling frequency. Artifacts, such as streaking, can appear on the image if the number of samples is insufficient.



- The DAS consists of the following parts

Detector → Current to Voltage Convertor → Pre Amplifier → Analog Integrator → Analog to Digital Convertor → Computer

- X-ray photons come on the detector.
- The detector detects the intensity in form of current.
- The current is converted into voltage.
- The analog integrator removes spikes.
- The analog signal is converted into digital form.
- This signal can now be processed and reconstructed in the computer.

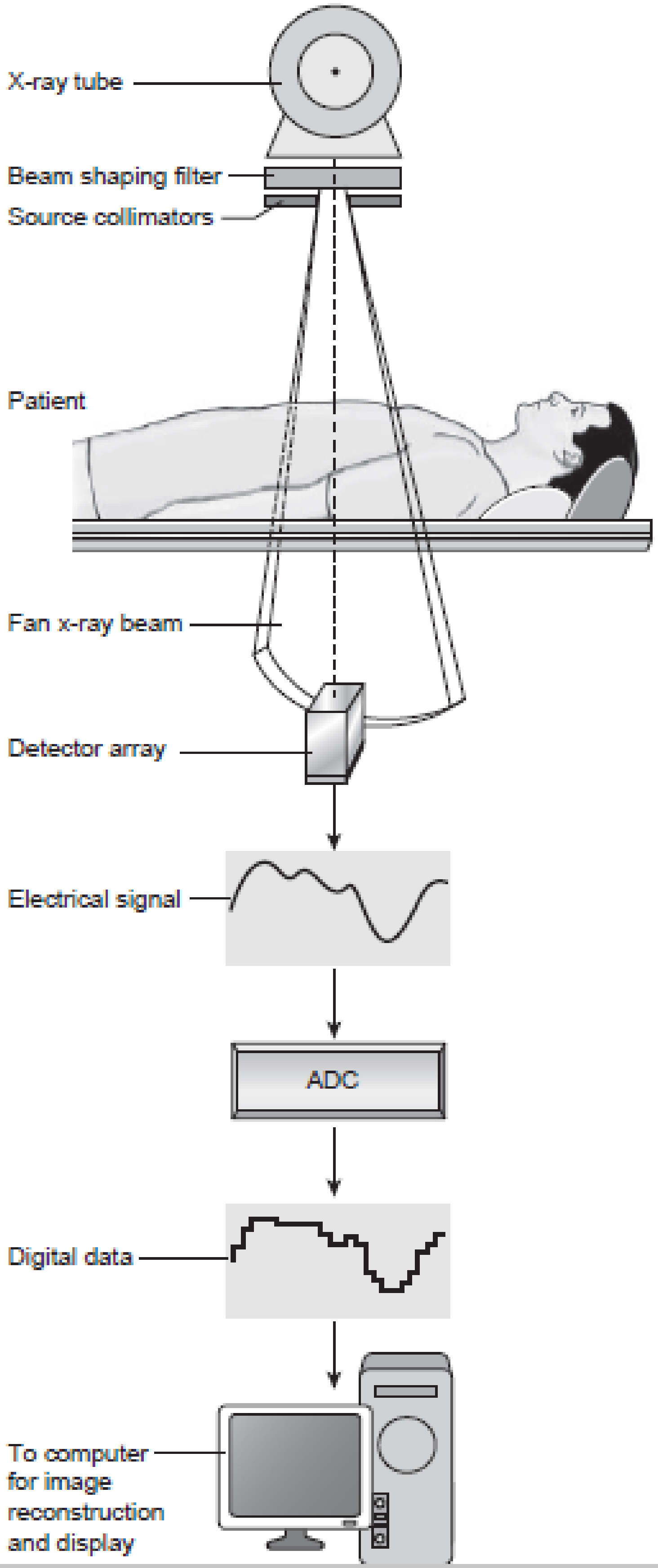


FIGURE 2-11 A schematic of data acquisition in CT.

Patient table

The patient lies on the table (or couch, as it is referred to by some manufacturers) and is moved within the gantry for scanning. The process of moving the table by a specified measure is most commonly called incrementation, but is also referred to as feed, step, or index. Helical CT table incrementation is quantified in millimeters per second because the table continues to move throughout the scan. The degree to which a table can move horizontally is called the scannable range, and will determine the extent a patient can be scanned without repositioning.

A numeric readout of the table location relative to the gantry is displayed. When the patient is placed within the gantry, an anatomic landmark, such as the xiphoid or the iliac crest, is adjusted so that it lies at the scan point. At this level, the table is referenced, which means that the table position is manually set at zero by the technologist. Accurate table referencing helps to maintain consistency between examinations. For example, if a lesion is seen on an image that is 50 mm inferior to the xiphoid landmark (zero point), the patient is removed from the gantry, and a ruler is used to measure 50 mm inferior from the xiphoid. This point provides an approximation of the location of the lesion. This system is also helpful if the scan will be repeated at a later date, exclusively through the area of interest determined on the earlier scan. For this reason, the setting of landmarks must be consistent among CT staff.

The specifications of tables vary, but all have certain weight restrictions. If the patient's weight exceeds the specified limits, scanning is often still possible. However, the table increments may not be as accurate. This problem affects small table increments more than those 5 mm or larger. On most scanners, it is possible to place the patient either head first or feet first, supine or prone. Patient position within the gantry depends on the examination being performed.

Various attachments are available for specific types of scanning procedures. For example, attachments for direct coronal scanning of the head and for therapy planning are common.

CT GANTRY CONTROL PANEL

1. Gantry Tilt (+/-30 degrees).
2. Laser Alignment Lights on/off.
3. Couch in/out.
4. Free (manual) Couch Movement.
5. Zero Couch Position.
6. Couch up/down.
7. Home Button (couch out & down).

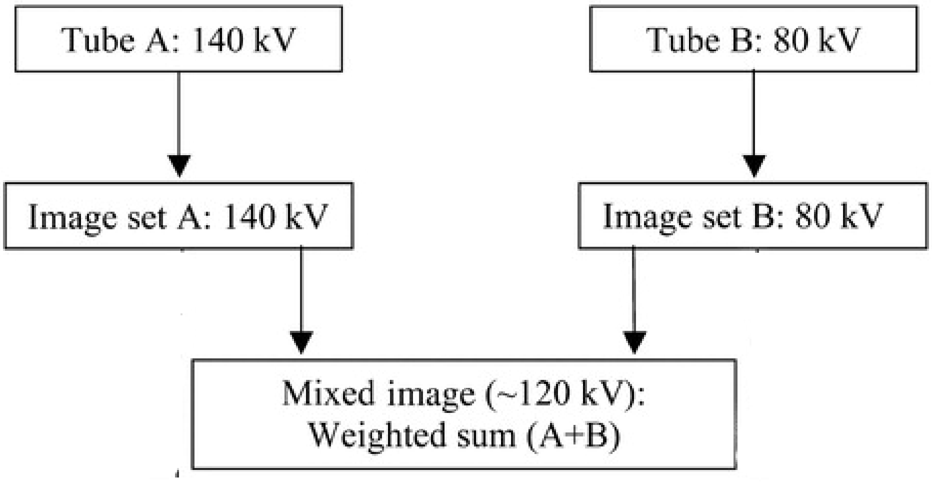
Key concepts:

- The gantry houses many of the components necessary to produce and detect x-rays. Components are mounted on a rotating scan frame.
- Slip rings permit the gantry frame to rotate continuously, making helical scan modes possible.
- High kV is used to increase the intensity of the beam, increasing its penetrating ability and thereby reducing patient dose. High kV settings also help to reduce the heat load on the x-ray tube by allowing a lower mA setting.
- Filtering the x-ray beam helps to reduce the radiation dose to the patient and improves image quality.
- The configuration of the x-ray tube to the detector determines scanner generation. The data-acquisition system, or DAS, measures the number of photons that strikes the detector, converts the information to a digital signal, and sends the signal to the computer.

Dual Energy CT (DECT)

Dual energy CT, also known as spectral CT, is a computed tomography technique that uses two separate x-ray photon energy spectra, allowing the interrogation of materials that have different attenuation properties at different energies. Whereas conventional single energy CT produces a single image set, dual energy data (attenuation values at two energy spectra) can be used to reconstruct numerous image types:

- weighted average images (simulating single energy spectra).
- virtual monoenergetic images (attenuation at a single photon energy rather than a spectrum).



- material decomposition images (is a method for differentiation and quantification of materials in a sample and it utilizes the energy dependence of the linear attenuation coefficient, i.e. mapping or removing substances of known attenuation characteristics, such as iodine, calcium, or uric acid):
    - virtual non-contrast images (iodine removed).
    - iodine concentration (iodine maps): Dual-energy CT iodine maps are used to detect pulmonary embolism (PE) with CT angiography.
    - calcium suppression (calcium removed): Calcium-suppressed (CaSupp) technique involving spectral-based images has been used to observe bone marrow edema by removing calcium components from the image.

- uric acid suppression (uric acid removed): is an imaging modality used for the diagnosis of gout. It is a good noninvasive alternative to synovial fluid aspiration. DECT is increasingly useful in diagnosing cases of gout where synovial fluid fails to demonstrate monosodium urate crystals.

The doctor might refer a patient for a dual energy CT scan for:

- It can selectively increase or decrease the effects of some chemical substances in the body, making some abnormalities clearer on the images taken; for example, iodine is a commonly used substance in X-ray contrast agents, and dual energy CT can selectively increase its effects to produce better images of blood vessels (CT angiography);
- Images with and without contrast agents can be obtained using a single examination instead of two separate examinations;
- It can detect particular substances in the body that can be useful for patients with kidney stones to see what type of stone is present and assist in deciding the type of treatment required;
- It can significantly improve image quality if patients have metal in the area being scanned (e.g. joint replacements).

Acquisition technique

There are different DECT acquisition technologies available from different vendors. These can be broadly classified as techniques that occur before the patient is scanned (prospective) which need to be pre-selected and those that occur after the patient is scanned (retrospective) which do not need to be pre-selected:

Prospective techniques

- dual-source:
    - two x-ray tubes producing different voltages (kVp) offset at approximately 90°
    - reconstructed in the image space
    - limited field of view (FOV) as both detectors cannot be the same size
    - excellent temporal resolution as both datasets acquired at the same time

- single-source consecutive
    - two helical scans are consecutively acquired at different tube potentials followed by coregistration for postprocessing
    - reconstructed in the image space
    - full field of view
    - poor temporal resolution as the patient is scanned twice (therefore increased dose)

- single-source twin-beam
    - two-material filter splits the x-ray beam into high-energy and low-energy spectra on the z-axis before it reaches the patient

- single-source sequential ("rotate-rotate")
    - each x-ray tube rotation is performed at high- and low- tube potential reconstructed in the image space
    - full field of view
    - poor temporal resolution as the patient is scanned twice (therefore increased dose)

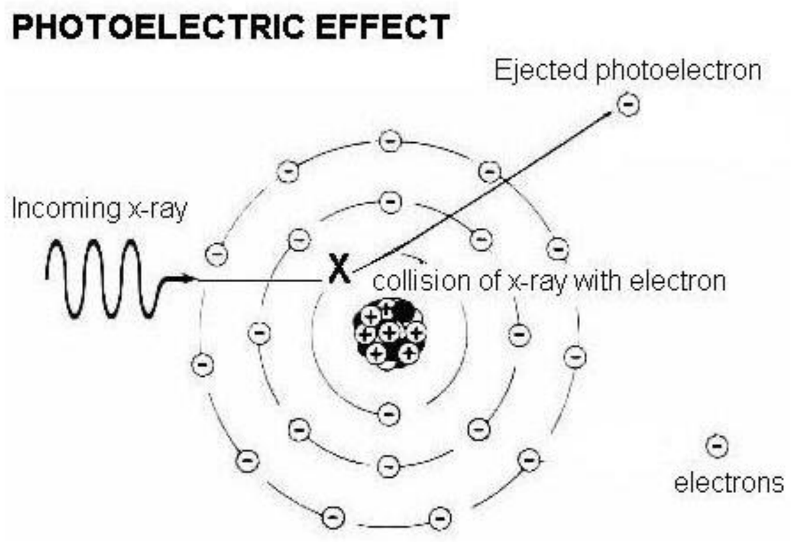
- single-source rapid kilovoltage switching (fast kVp-switch)
  - the x-ray tube switches between high- and low- tube potential multiple times within the same rotation
  - reconstructed in the projection space
  - full field of view
  - slight reduction in temporal resolution due to tube rotation

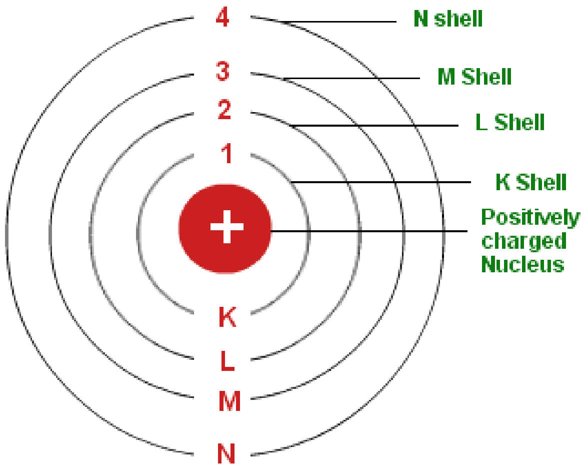
Retrospective techniques

- dual-layer DECT ("sandwich")
  - the top (innermost) layer of the detector absorbs low-energy photons while high-energy photons pass through to the bottom (outermost) layer
  - reconstructed in the projection space
  - full field of view
  - excellent temporal resolution as both datasets acquired at the same time

Basic principles

X-ray photons primarily interact with matter via the photoelectric effect, and Compton scattering producing the diagnostic images used in medicine today. When an atom undergoes the photoelectric effect (the X-Ray photon falling on some matter is absorbed by the matter and its energy is transferred to an electron of the matter.), the electron from that respective K-shell (the innermost shell of electrons surrounding an atomic nucleus and constituting the lowest available energy level for the electrons) is ejected via the incident photon. As that electron is excited, vacant space is 'filled' by a neighboring electron, releasing energy as a photoelectron.





In short, when a photon has sufficient energy to overcome the electron's binding energy in the K-shell, that atom undergoes the photoelectric effect.

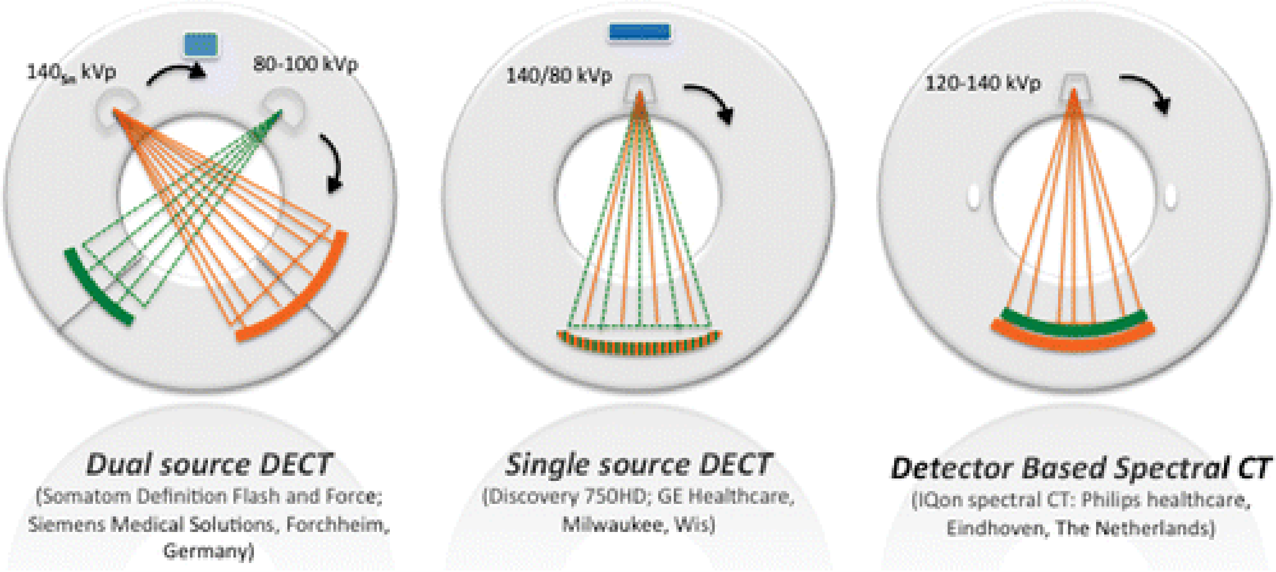
Each substance owns a unique K-shell binding energy; known as the K-edge. There is a significant spike in attenuation that results just beyond the energy of the K-edge, this peak is unique to every material and holds valuable information about the substance's composition.

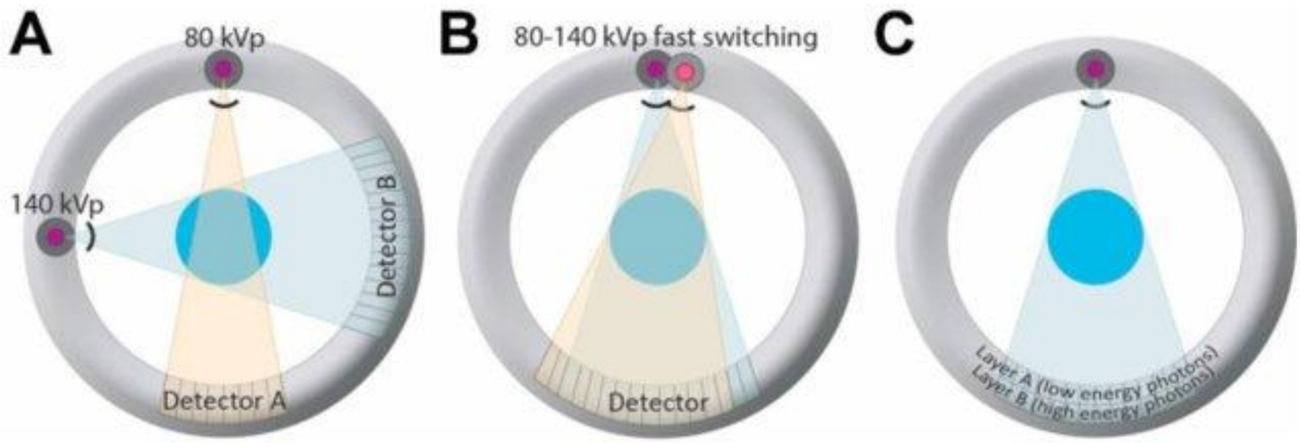
The different photoelectric energies and K-edges are the bread and butter of dual-energy CT. Although most elements in the human body have very low K-edges (0.01-0.53 keV), elements like iodine and calcium have higher K-edges of 33.2 keV and 4.0 keV respectively, making them sufficiently larger than surrounding structures and are particularly important in the clinical setting.

For instance, at 80 kVp a structure that contains no (introduced) iodine, such as the liver, has an attenuation based on its K-edge of x, yet when iodine (33.2 keV) is introduced into that same structure, it has a higher attenuation of y bringing it closer to 80 kVp.

As 80 kVp is closer to 33.2 keV than 140 kVp, the structures containing iodine will retain less attenuation as the kVp progressed beyond the K-edge of iodine. Therefore, when using two energies, it is possible to delineate structures based solely on their attenuation differences between 80 kVp and 140 kVp.

A dual x-ray source, tube A (140 kVp) and tube B (80 kVp or 100 kVp) with an angular offset of 90 degrees are preferred offsets for a dual source scanner.





Different dual-energy CT (DECT) scanners currently in clinical use. (A) Illustration of a dual source DECT, consisting of two source x-ray tubes with corresponding detectors; (B) Illustration of a single source DECT with rapid kVp switching. With this type of scanner, the tube voltage follows a pulsed curve, and projection data are collected twice for every projection, one at high and one at low tube voltage, during rapid kVp switching; (C) Illustration of a dual layer DECT, consisting of a single source and single (but layered) detector. The detector is composed of two scintillation layers enabling separation of high and low energy spectra produced by a single source.