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Lecture Title: Data Acquisition, Part 1.

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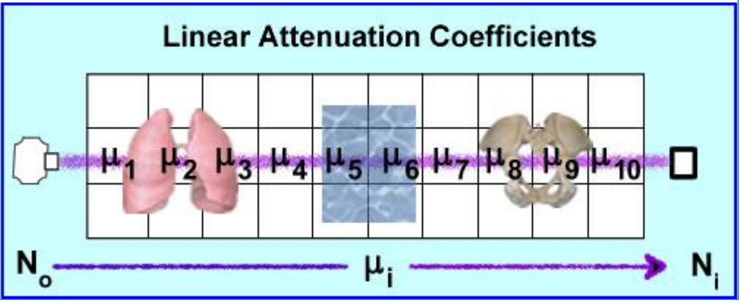
- No is the production of the x-ray that travels through 3D space (the example below shows 10 voxels)
- Depending of the amount of density encounters (air, liquid, and/or bone) there will be a variation in the ray's attenuation. These values of attenuation is referred to as u_i where the beam may encounter different degrees of attenuation within a certain volume of voxels. The end point of the attenuated ray is then referred to Ni were the detector records the amount of attenuation. End results - linear attenuation coefficient , μ, that occurs from that specific x-ray.



- The chart shows attenuation coefficients for a 100 kVp x-ray beam. While a CT tube uses a poly-energetic beam within 100 kVp (1% variation), its output can be adjusted between 70 and 140 kVp.
- The attenuated sum (X_n) occurs after the initial ray N_o passes through a set of voxels, which in turn becomes N_i and can written as $X_i = -\ln(N_i/N_o)$.
- When a narrow beam of mono-energetic photons with an incident intensity I_0, penetrating a layer of material with mass thickness x and density ρ, emerges with intensity I given by the exponential attenuation law as:

$$I = I_o exp\left[-\left(\frac{\mu}{\rho}\right)x\right]$$

Linear Attenuation Coefficients (cm^{-1})

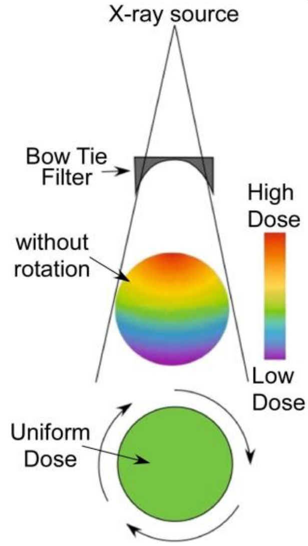
Bone	0.528
Muscle	0.237
Brain White	0.213
Brain Gray	0.212
Blood	0.208
Water	0.206
Fat	0.185
Lung	0.093
Air	0.0004

Beam Hardening: A Polychromatic Challenge

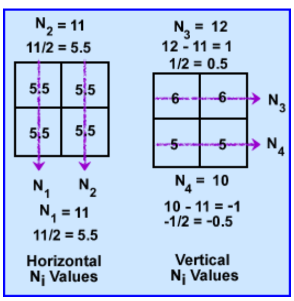
- The standard attenuation formula assumes a mono-energetic beam. However, actual CT x-ray tubes produce a poly-energetic beam (e.g., varying within 100 kVp).
- As the beam passes through the patient, lower-energy (soft) X-rays are absorbed first by the tissue.
- This increases the average energy of the remaining beam as it travels, making it harder.
- **The Problem:** This causes nonlinear attenuation, leading to cupping artifacts in which the center of a uniform object falsely appears less dense than its edges.
- **The Engineering Solution:** Compensating filters (such as bow-tie filters) pre-harden the beam, removing soft x-rays to make the beam's intensity more uniform and reduce this artifact by reducing the intensity at the periphery. They ensure a more uniform X-ray signal reaches the detector.

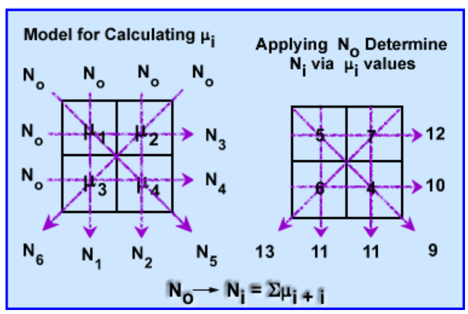
Beam Hardening: A Polychromatic Challenge

- The bow-tie filter used in CT is a specialized beam-shaping component designed to compensate for the unequal x-ray attenuation caused by the patient's roughly oval or cylindrical body shape. Being thicker at the edges and thinner in the center, it attenuates the beam more at the periphery, where the patient is thinner, resulting in a more uniform x-ray flux reaching the detector.
- They are typically mounted near the collimator at the point where the X-ray beam exits the tube housing. Most modern CT scanners feature a motorized assembly (often a "filter wheel" or slide) that automatically swaps filters based on the selected clinical protocol.



- How CT actually determines the µ value in a voxel?
- Consider look at a 2x2 matrix using an iterative algorithmic approach, Algebraic Reconstruction Technique (ART).
- The attenuation in these 4 pixels are hypothetically unknown, however, when applying an x-ray beam to them, N_o, at six different angles, 6 N_i, the values are detected ($N_{1\text{-}6}$). This is demonstrated below





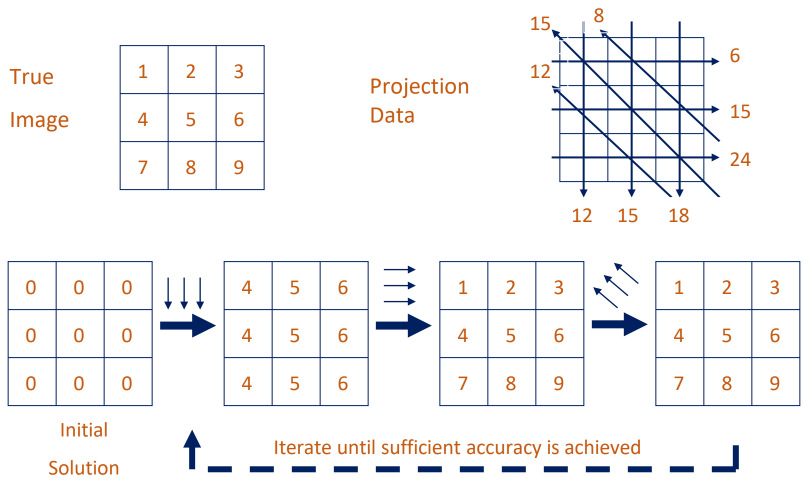
Algebraic Reconstruction Technique (ART)

- A low-complexity iterative solver to the algebraic reconstruction problem
- Starts with an initial estimate and tries to push the estimate closer to the true solution
  - Instead of back-projecting the average ray value, the error between the projection computed from current estimate and the true is used

$$Update = (P_\theta(\rho) - \sum_{x,y} \alpha_\rho^\theta(x,y) \cdot \hat{I}(x,y)) / N$$

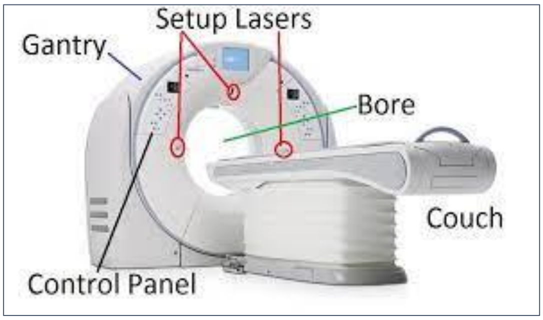
Update = (True Projection – Estimated Projection) divided over all points in projection path

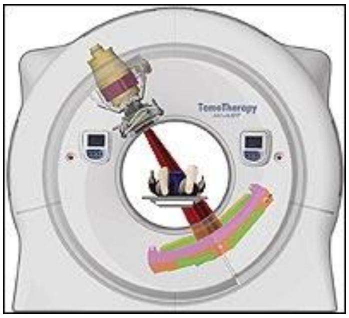
ART Example

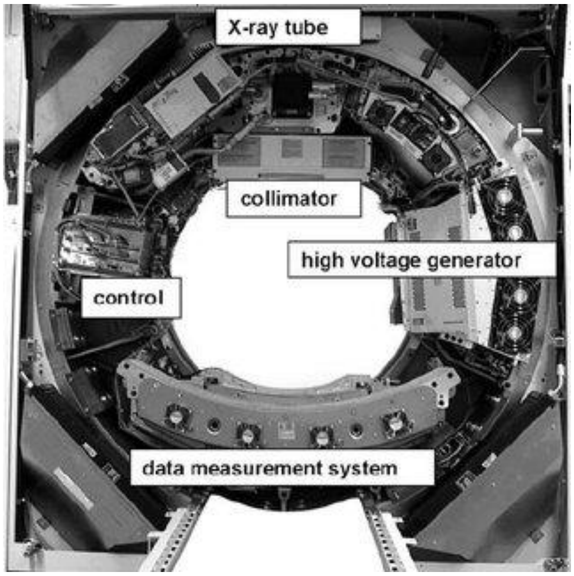


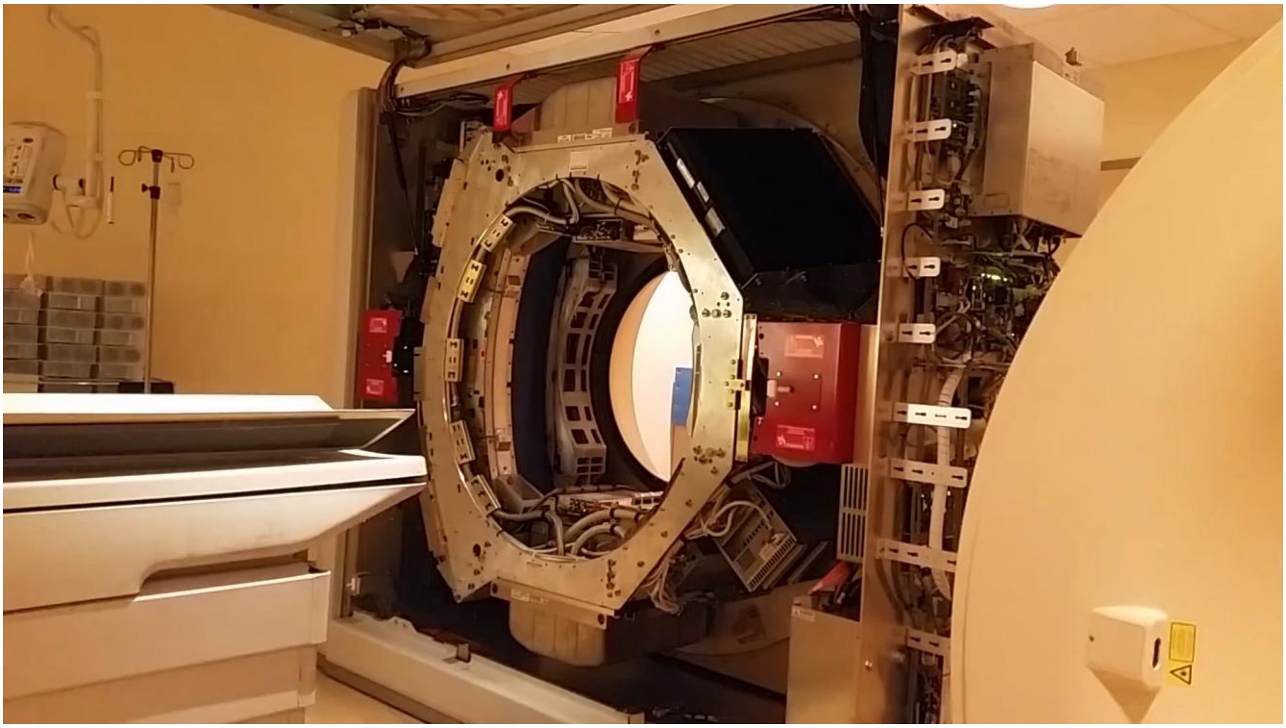
- All makes, and models of CT scanners are similar in that they consist of a scanning gantry, x-ray generator, computer system, operator's console, and physician's viewing console.
- CT can be broken down into three segments: data acquisition, image reconstruction, and image display.
- Data are acquired when x-rays pass through a patient to strike a detector and are recorded. The major components involved in this image creation phase are the gantry and the patient table.

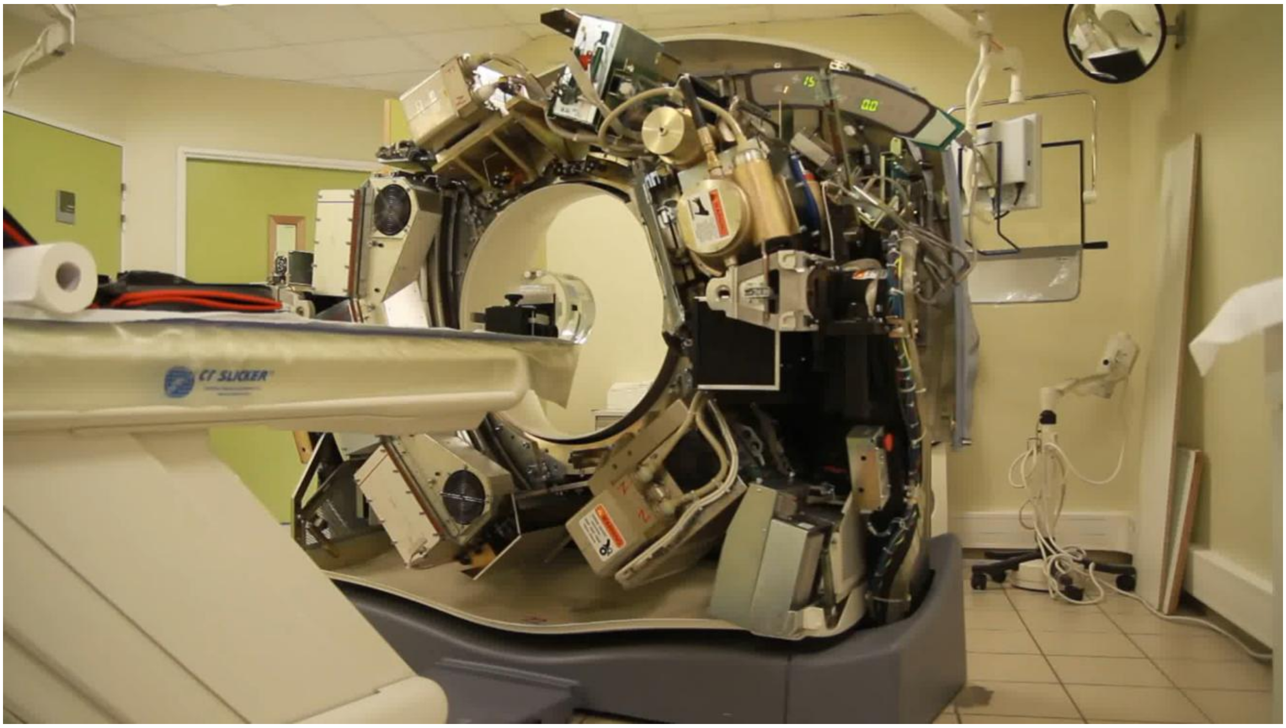
- The gantry is the ring-shaped part that houses many of the components necessary to produce and detect x-rays.
- The range of gantry aperture size is typically 70 to 90cm, and can be tilted either forward or backward (usually ±15° to ±30°) as needed to accommodate a variety of patients and examination protocols.
- The gantry also includes a laser light that is used to position the patient within the scanner.






X-ray tube
collimator
high voltage generator
control
data measurement system



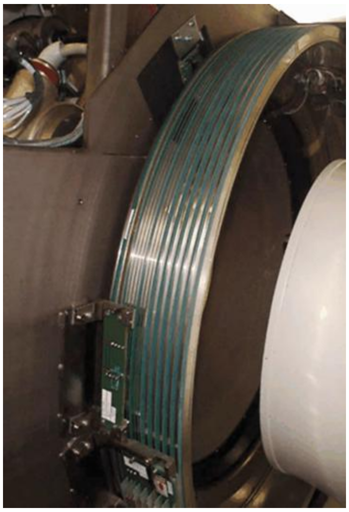


- Slip rings are electromechanical devices that use a brush-like apparatus to provide continuous electrical power and electronic communication across a rotating surface.
- They permit the gantry frame to rotate continuously, eliminating the need to straighten twisted system cables.
- Slip rings used to bring power to x-ray tube

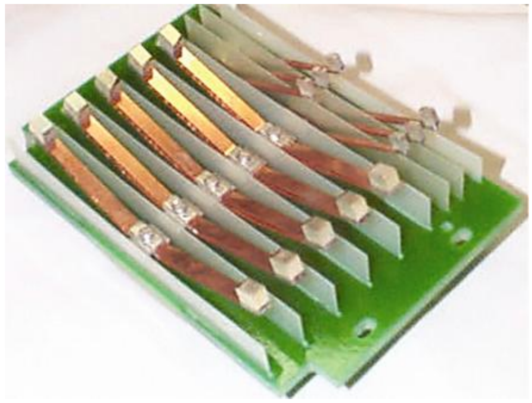
on rotating gantry of a helical CT machine

and, for some designs, to acquire

information from the detector array.



- The shiny metal strips carry electric signals that are swept off by special brushes. The brushes are not in the form of bristles but rather of metal blocks (in this case a silver alloy).
- The five pairs of larger brushes provide the voltage required by the x-ray tube, and the three pairs of smaller ones transfer signals from the gantry controller.



- ***High-frequency generators*** can be located within the gantry.
- Generators produce high voltage and transmit it to the x-ray tube. The power capacity of the generator is listed in kilowatts (kW), and determines the range of exposure techniques (i.e., kV and mA settings) available on a particular system.
- CT generators produce high kV (generally 120–140 kV) to increase the intensity of the beam, which will increase the penetrating ability of the x-ray beam and thereby reduce patient dose.
- In addition, a higher kV setting will help to reduce the heat load on the x-ray tube by allowing a lower mA setting. Reducing the heat load on the x-ray tube will extend the life of the tube

- The CT tube is designed to handle the stresses of the scanning protocols that often require multiple long exposures performed on numerous patients daily.
- The way a tube dissipates the heat that is created during x-ray production is critical.
- All manufacturers list generator and tube cooling capabilities in their product specifications.

➢ KW: the system generator’s maximum power in KW.

➢ MHU: the anode heat capacity in million heat units (MHU), e.g. 7 MHU tube will last more than 150 million mAs.

➢ KHU: the maximum anode heat dissipation rate in thousand heat units (KHU).

- These specifications can be helpful in comparing various CT systems. Also it is important to compare the length of protocols that the tube will allow and how quickly they can be repeated.

The Heat Units (HU) Formula for High-Frequency Generators

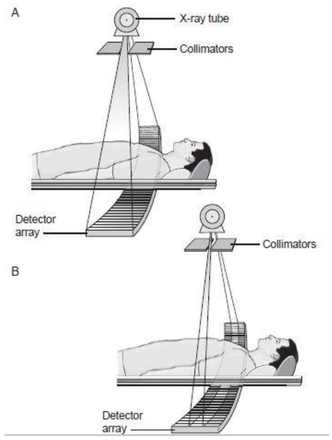
The Formula: $HU = 1.4 \times kVp \times mA \times s$. This formula calculates the amount of thermal energy (heat) deposited into the x-ray tube's anode target during a single exposure. Since x-ray production is highly inefficient (approximately 99% of the kinetic energy from the electron stream is converted to heat, and only 1% to x-rays), managing this thermal load is one of the biggest engineering challenges in CT design. The formula variables are:

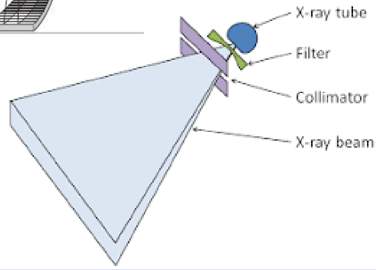
- kVp (Peak Kilovoltage): The electrical potential applied across the tube, determining the maximum energy of the electrons hitting the target.
- mA (Milliamperage): The tube current, representing the number of electrons flowing from the cathode to the anode.
- s (Time in seconds): The duration of the exposure.
- The 1.4 Constant (The Generator Factor): Modern CT scanners use high-frequency generators that convert the power to a nearly constant, flat-line voltage with less than 1% ripple. Since the voltage stays at its absolute peak (kVp) for the entire exposure, it deposits roughly 40% more heat into the anode. The factor of 1.4 is derived from the Root Mean Square of the waveform. 1 Heat Unit is roughly equivalent to 1 Joule of energy ($1\ HU \approx 1\ J$)

Example: Calculate the total Heat Units (HU) generated in a CT scanner with a high-frequency generator running a scanning protocol at 120 kVp, 300 mA, for an exposure time of 15 seconds. Calculate the total Heat Units (HU) generated.

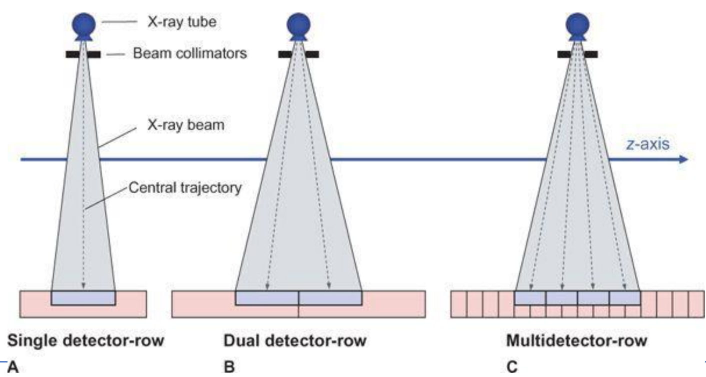
- Formula (High-Frequency Generator): $HU = 1.4 \times kVp \times mA \times time\ (s)$
- Calculation: $HU = 1.4 \times 120 \times 300 \times 15$ >>> $HU = 756{,}000$ Heat Units (or 0.756 MHU)
- **Engineering Context:** If the tube has a maximum capacity of 7.0 MHU, this single scan consumes about 10.8% of the tube's total thermal capacity. Proper heat dissipation (KHU) is critical to prevent damage before the next scan can begin.

- Collimators restrict the x-ray beam to a specific area, thereby reducing scatter radiation.
- Scatter radiation reduces image quality and increases the radiation dose to the patient.
- Reducing the scatter improves contrast resolution and decreases patient dose.
- Collimators control the slice thickness by narrowing or widening the x-ray beam. The source collimator is located near the x-ray source and limits the amount of x-ray emerging to thin ribbons.

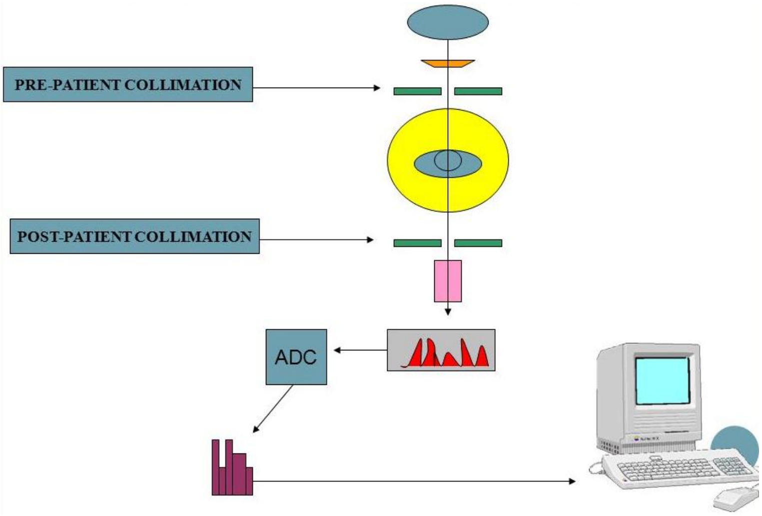




- The source collimator affects patient dose and determines how the dose is distributed across the slice thickness (i.e., dose profile).
- The source collimator resembles small shutters with an opening that adjusts, dependent on the operator's selection of slice thickness.
- In MDCT systems (Multidetector computed tomography), slice thickness (0.5 to 10 mm) is also influenced by the detector element configuration.



- Some CT systems also use pre-detector collimation located below the patient and above the detector array.
- This collimator shapes the beam after it has passed through the patient.
- The primary functions of pre-detector collimators are to ensure the beam is the proper width as it enters the detector and to prevent scatter radiation from reaching the detector.

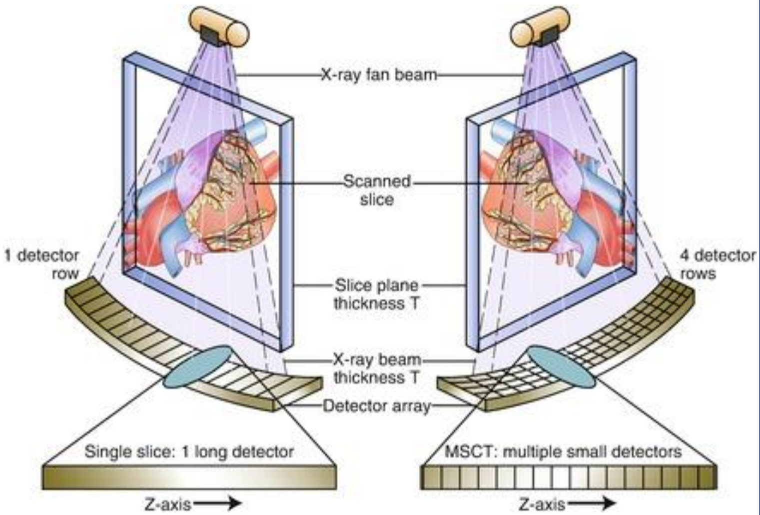


The Data Acquisition System (DAS) & ADC

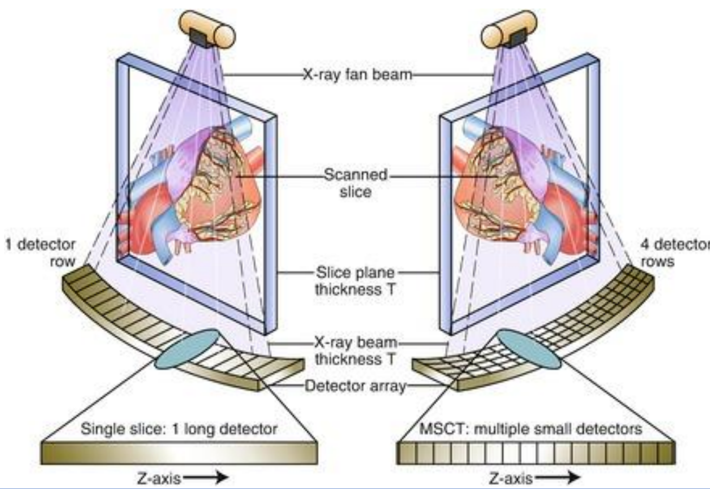
The DAS is the crucial bridge between the x-ray detectors and the computer system.

- **Step 1: Measurement.** Photons strike the solid-state detector, converting x-ray energy into light, which the photodiode converts into a tiny electrical current (Analog Signal).
- **Step 2: Amplification.** The DAS amplifies this extremely weak electrical current so it can be accurately measured.
- **Step 3: Logarithmic Conversion.** The system applies a logarithmic conversion corresponding to the beam's attenuation equation.
- **Step 4: Digitization (ADC).** The Analog-to-Digital Converter transforms the analog signal into digital binary code for the image reconstruction computer.

- The detectors collect information regarding the degree to which each anatomic structure attenuated the beam while passing through the patient.
- The detector array comprises detector elements situated in an arc or a ring, each of which measures the intensity of transmitted x-ray radiation along a beam projected from the x-ray source to that particular detector element.



- Also included in the array are elements referred to as reference detectors that help to calibrate data and reduce artifacts.
- The scan field of view determines the size of the fan beam, which, in turn, determines the number of detector elements that collect data.
- Detectors can be made from different substances, each with their own advantages and disadvantages



- *The optimal characteristics of a detector are as follows:*
- 1) high detector efficiency, defined as the ability of the detector to capture transmitted photons and change them to electronic signals;
- 2) low, or no, afterglow, defined as a brief, persistent flash of scintillation that must be taken into account and subtracted before image reconstruction;
- 3) high scatter suppression;
- 4) high stability, which allows a system to be used without the interruption of frequent calibration.

- *Overall detector efficiency is the product of a number of factors. These are:*
- 1) stopping power of the detector material;
- 2) scintillator efficiency (in solid-state types);
- 3) charge collection efficiency (in xenon types);
- 4) geometric efficiency, defined as the amount of space occupied by the detector collimator plates relative to the surface area of the detector;
- 5) scatter rejection.

- *Capture efficiency:* the ability with which the detector obtains photons that have passed through the patient.
- *Absorption efficiency:* the number of photons absorbed by the detector depends on the detector face's physical properties (e.g., thickness, material).
- *Response time:* is the time required for the signal from the detector to return to zero after stimulation of the detector by x-ray radiation so that it is ready to detect another x-ray event.
- The detector response is generally a function of the detector design.
- *Dynamic range:* is the ratio of the maximum signal measured to the minimum signal the detectors can measure.

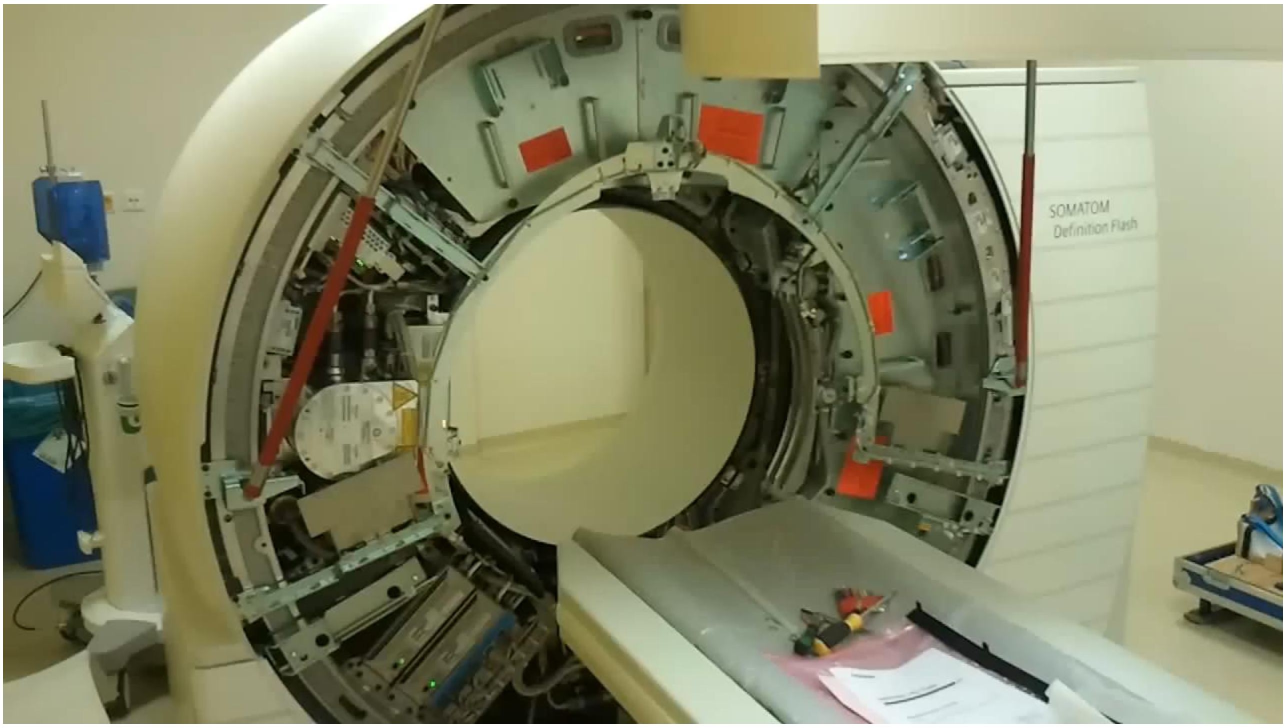
- All new scanners possess detectors of the solid-state crystal variety.
- Detectors made from xenon gas have been manufactured but have largely become obsolete as their design prevents them from use in MDCT systems.
- However, some xenon gas detector systems may still be in use on older models.

TABLE 2-1 Characteristics of Detectors

Solid-State Crystal	Pressurized Xenon Gas
High photon absorption	Moderate photon absorption
Sensitive to temperature, moisture	Highly stable
Solid material	Low-density material (gas)
Can exhibit afterglow	No afterglow
No front window loss	Losses attributable to front window and the spaces taken up by the plates

MDCT Array Types & Geometric Efficiency

- **Matrix (Symmetric) Arrays:** All detector rows are of equal thickness.
- **Adaptive (Asymmetric) Arrays:** Inner detector rows are thinner for high spatial resolution, while outer rows are thicker.
- **Geometric Efficiency:** The ratio of the active detector area to the total area, including the spacing bars (septa) separating the detectors.
- Example Calculation: Active Detector Width = 1.25 mm, Spacing Bar (Septa) Width = 0.25 mm
- Formula: Efficiency=(Active width/Total Width)×100 >>> Efficiency = (1.25/(1.25+0.25))=1.25/1.5=0.833
- Result: 83.3% Geometric Efficiency
- **Engineering Context:** ~16.7% of the X-rays hit the dead space (spacing bars). They contribute to the patient's radiation dose but are not recorded as useful image data.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