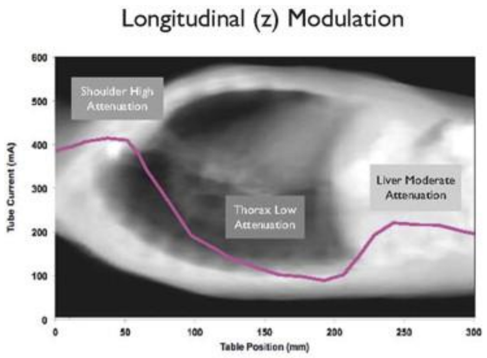
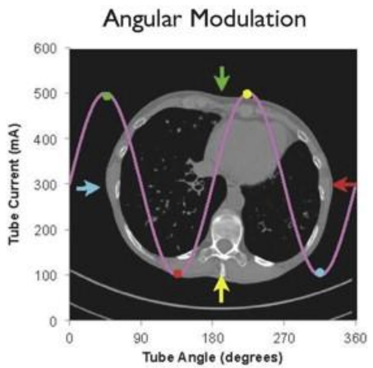
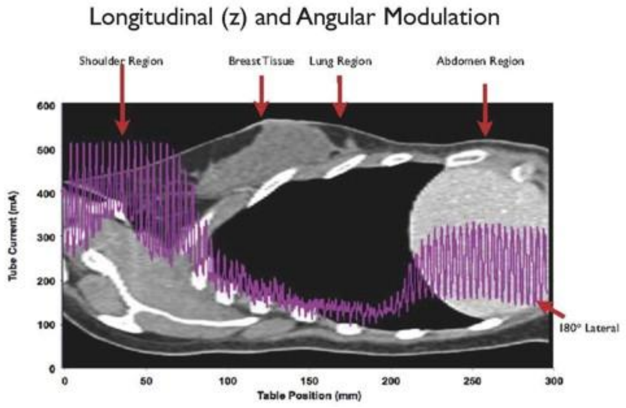
CT Radiation Exposure and Protection







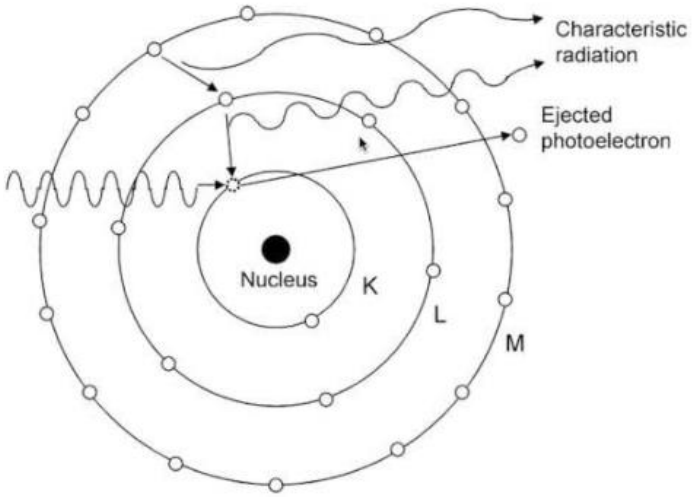
- **Exposure (R or C/kg):** Ionization produced in air; classic measurement with ion chamber. Defines how much radiation hits a point in air (patient surface, phantom).

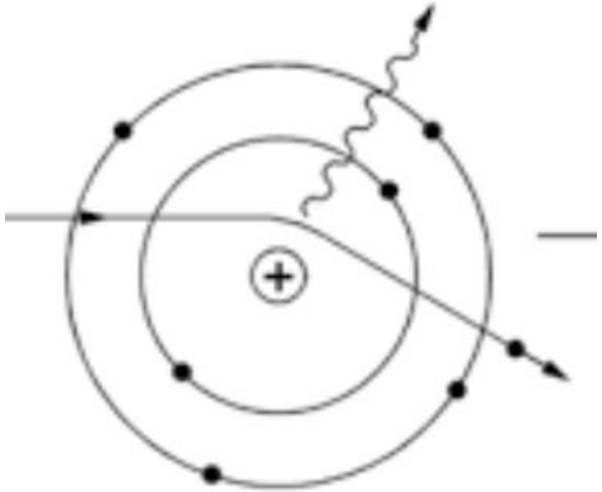
- **Absorbed dose (Gy):** Energy deposited per mass in tissue; primary quantity for biological effect and protection. 1 Gy = 100 rad.

- **Effective dose (Sv):** Risk-weighted dose accounting for radiation type and tissue weighting; compares partial-body to whole-body risk. 1 Sv = 100 rem

- **Radiation Effects: Stochastic vs. Deterministic**
- **Stochastic Effects (Probabilistic)**
- **Definition:** Random effects that occur by *chance* — the *probability* (not severity) increases with dose.
- **Key Feature: No threshold** — even the smallest dose can theoretically cause an effect.
- **Example:** Cancer, genetic mutations.
- **Model:** Based on the **Linear-No-Threshold (LNT)** model — assumes risk increases linearly with dose, with *no safe level*.
- **Importance:** Forms the **scientific foundation** of modern **radiation protection principles** (ALARA, dose limits).

- **Deterministic Effects (Tissue Reactions)**
- **Definition:** Biological effects that occur **after a threshold dose** is exceeded.
- **Key Feature: Severity increases with dose** — once threshold is passed, damage worsens predictably.
- **Examples:**
    - **Erythema (skin reddening)**
    - **Epilation (hair loss)**
    - **Cataracts**
    - **Skin burns, sterility** (at very high doses)
- **Relevance to CT:** Rare in diagnostic CT, but possible in **interventional or repeated high-dose exams**.
- **Protection Goal:** Keep exposures **below threshold levels** to prevent tissue injury.





- **X-Ray Production Refresher**

- **Bremsstrahlung Radiation (Braking Radiation)**
- Produced when high-speed electrons are **decelerated** or **deflected** by the positive electric field of tungsten nuclei in the anode.
- Generates a **continuous energy spectrum** — photon energies range from near zero up to the selected **kVp**.
- **Dominant type** of X-ray production in **CT and diagnostic imaging**

- **Characteristic Radiation**
- Occurs when an incoming electron **ejects an inner-shell electron** (usually K-shell) of the target atom.
- An **outer-shell electron drops** to fill the vacancy, emitting an X-ray photon of **discrete energy**.
- Contributes a **smaller fraction** of the total X-ray beam but adds distinct spectral lines

Practical Control Factors in CT

- **kVp** → controls **beam quality and** (penetration and energy spectrum).Quantity is also affected.
- **mAs** → controls **beam quantity** (photon flux and image noise).
- **Filtration** → removes **low-energy (non-diagnostic) photons**, improving beam quality and **reducing patient skin dose**.
    - *Added filtration = "harder" beam = better penetration with less unnecessary exposure.*

- **Typical kVp Settings in CT and CTA**
- **Standard CT Imaging:**
- **Adult body CT:** 120 kVp (most common)
- **Head CT:** 120–140 kVp (for dense skull penetration)
- **Pediatric CT:** 80–100 kVp (to lower dose)
- **Low-dose or screening CT (e.g., lung, colon):** 100–110 kVp

- Beam Frequency, Wavelength, and Intensity.

- **In CT, increasing kVp** raises the frequency and lowers the wavelength of x-ray photons, resulting in a beam with a higher mean energy. A higher-energy beam penetrates tissue more efficiently, resulting in less attenuation and better transmission through the patient. This enhances beam quality the penetrating power of the x-rays.

- **mAs** controls the quantity, or the total number of photons emitted. Filtration further improves quality by removing low-energy photons that would only increase skin dose without contributing to image formation. Other factors, such as **exposure time** and **pitch**, also affect the photon flux reaching each slice.

- The primary beam is the original X-ray beam emerging from the tube and striking the patient, while the remnant (or exit) beam is the portion that passes through the body and reaches the detectors to form the CT image. The balance between beam quality and quantity determines image contrast, noise, and patient dose.

Inverse Square Law and CT Staff Safety

- The **Inverse Square Law** states that **radiation intensity is inversely proportional to the square of the distance** from the source:
- **Dose ∝ 1 / distance^2**

- This means that **doubling your distance** from the radiation source reduces exposure by a factor of **four**, and **tripling it** reduces exposure by a factor of **nine**. In CT, the **patient becomes the main source of scatter radiation**, not the gantry itself, since most of the primary beam is confined within the scanner housing. Therefore, stepping even a few feet back during scanning **dramatically lowers occupational dose**.

- For **contrast injections**, technologists should practice **remote injection** whenever possible, either by **using power injectors with long tubing** or by initiating scans **from outside the room**. This approach minimizes unnecessary exposure while maintaining patient safety and workflow efficiency.

- **Key takeaway:** In CT, **distance is the most effective shield** — every extra step back matters.

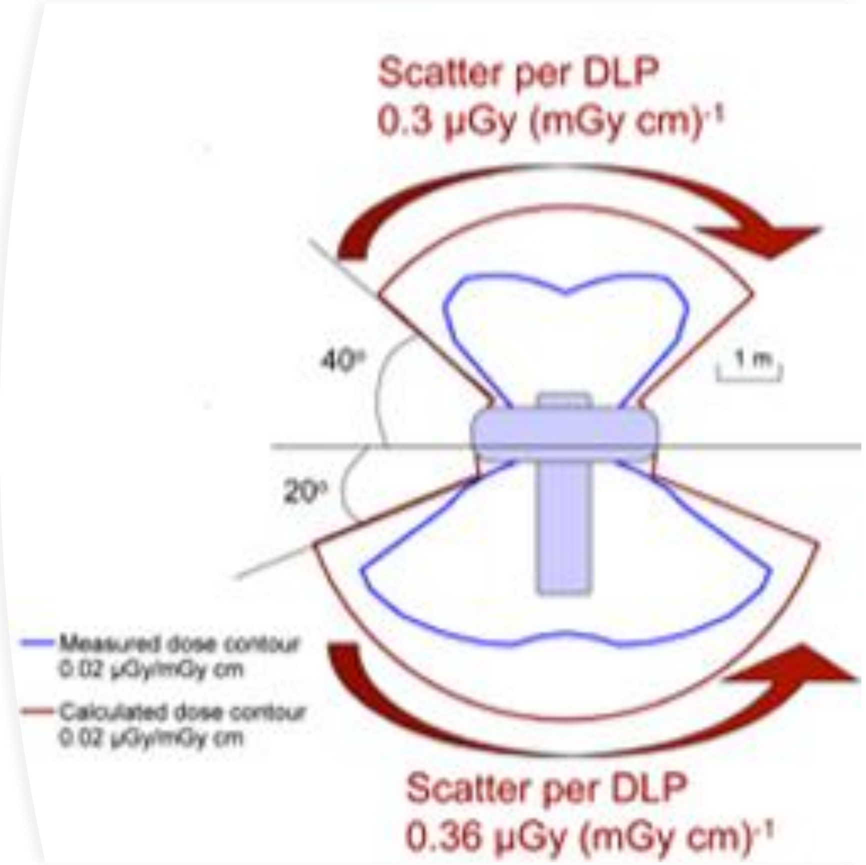
Photon Interactions with Matter

Photoelectric Effect

- Occurs when an x-ray photon is **completely absorbed** by an inner-shell electron.
- Dominates at **lower photon energies** and in **high atomic number (Z)** materials like **bone** or **iodinated contrast**.
- Creates **high image contrast** but also **increases patient dose**, since more photons are absorbed rather than transmitted.
- Clinically, this effect enhances the visibility of bone and contrast-filled vessels, especially at **lower kVp** settings.
-

Compton Scatter

- Occurs when a photon interacts with an **outer-shell electron**, losing part of its energy and changing direction.
- Dominates in **soft tissue** at the **medium-to-high photon energies** typical of CT.
- Reduces image contrast and contributes to **image noise and occupational exposure**.
- Scatter control (collimation, bowtie filters, shielding) helps reduce its impact on both image quality and staff dose.

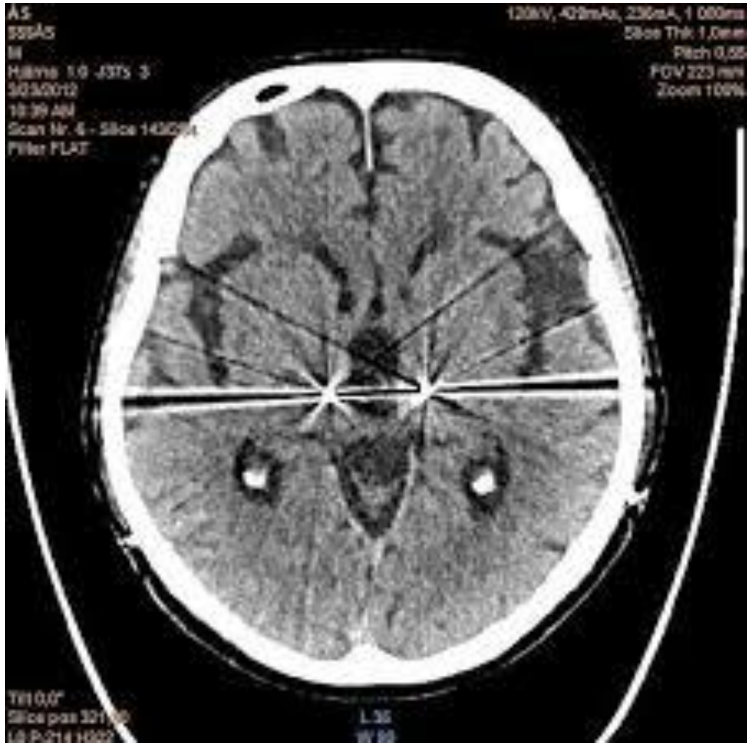


- **Clinical Takeaway:**

Selecting **kVp** is a balance:

- Lower kVp → ↑ photoelectric absorption → ↑ contrast (especially with IV contrast)
- Higher kVp → ↑ Compton scatter dominance → ↓ contrast, ↑ penetration

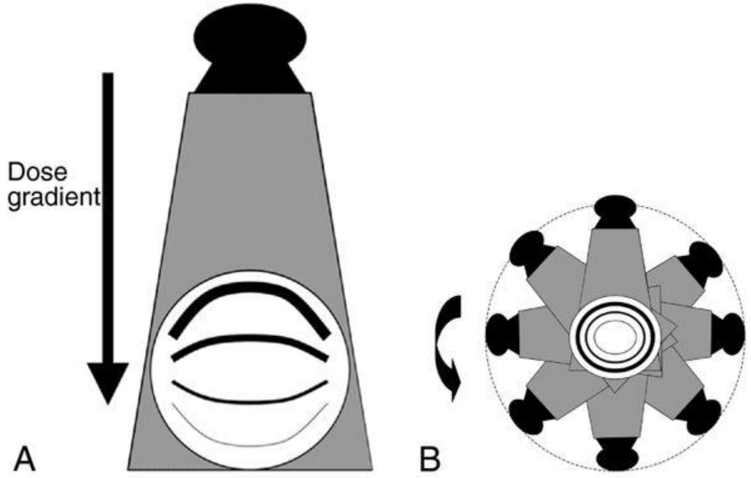
Optimal protocols tailor this balance to the patient's size and diagnostic task.



Attenuation Order (Lowest → Highest):

- **Air < Fat < Water/Soft Tissue < Muscle < Bone < Metal**

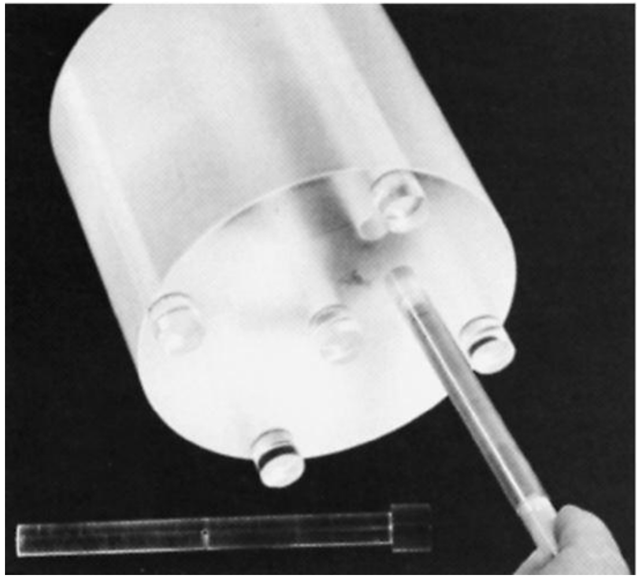
- **Air:** Minimal attenuation; appears **black** (≈ –1000 HU).
- **Fat:** Low attenuation; appears **dark gray** (≈ –100 to –50 HU).
- **Water/Soft Tissue:** Moderate attenuation; appears **gray** (≈ 0 HU).
- **Muscle:** Slightly higher density; **light gray** (≈ +40 HU).
- **Bone:** Very high attenuation; **white** (≈ +1000 HU).
- **Metal or contrast:** Extreme attenuation; may cause **streak artifacts** due to photon starvation or beam hardening.



CT Patient Exposure Geometry vs. Radiography

- **Radiography** uses a **single x-ray projection** through the patient onto a detector. The beam is emitted for a **fraction of a second**, striking the body from one direction only. Dose can be described simply as an **entrance skin exposure** — energy deposited at the skin surface from that one projection.
- In contrast, **Computed Tomography (CT)** exposes the patient **from multiple angles**. The x-ray tube **rotates 360°** around the patient while the **table moves continuously or step-wise** through the gantry. Each rotation delivers radiation through a thin **axial slice or helical volume**, meaning every tissue voxel is irradiated by photons from many directions.

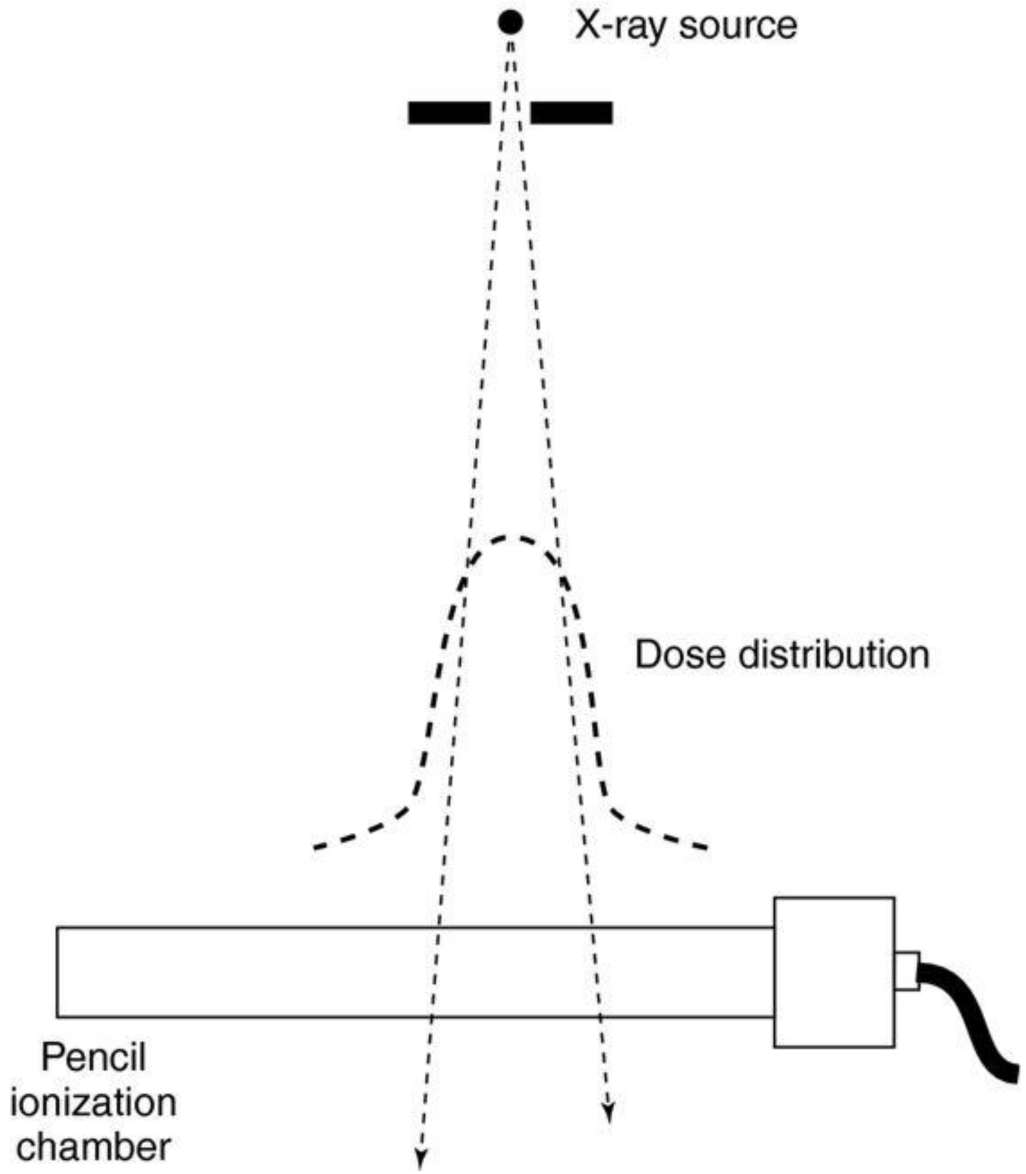
Dosimeters & Phantoms for CT



- **Historical Dosimetry Methods:**
- **Film badges:** Early qualitative method for detecting exposure; darkening indicated accumulated dose.
- **Thermoluminescent dosimeters (TLDs):** Provided quantitative dose readings by releasing light when heated used for both **patient dose studies** and **personnel monitoring**.
- **Ionization chambers:** Measure radiation exposure directly by collecting charge produced in air — the gold standard for **accurate dosimetry**.
-
- **CT-Specific Dosimetry: The Pencil Ionization Chamber**
- Modern CT dose measurements use a **100-mm pencil ionization chamber** (sometimes called a **pencil chamber**).
- It measures the **integrated dose** along the z-axis of the x-ray beam during one full rotation.
- This value forms the basis for **CTDI (Computed Tomography Dose Index)** calculations.

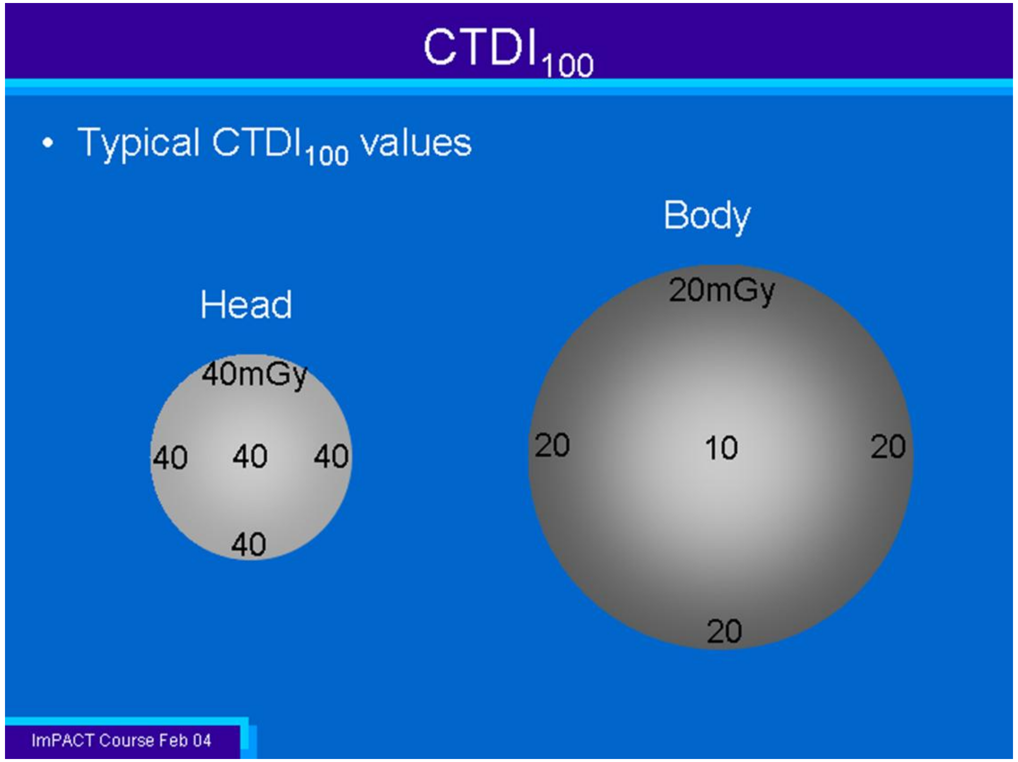
- Because of this complex **three-dimensional exposure geometry**, CT dose cannot be expressed as a single entrance skin dose. Instead, CT uses standardized **dose indices** that account for rotational geometry and table movement:
- **CTDIvol (Computed Tomography Dose Index, volume):** Measures the average dose within the scan field per rotation, standardized using phantoms.

- **DLP (Dose–Length Product):** CTDIvol × scan length; represents total energy delivered along the z-axis.
- These indices describe **dose distribution within a volume**, not just surface exposure. They enable consistent dose comparison across scanners and protocols, reflecting CT's **volumetric, rotational** nature rather than a single static projection as in radiography.

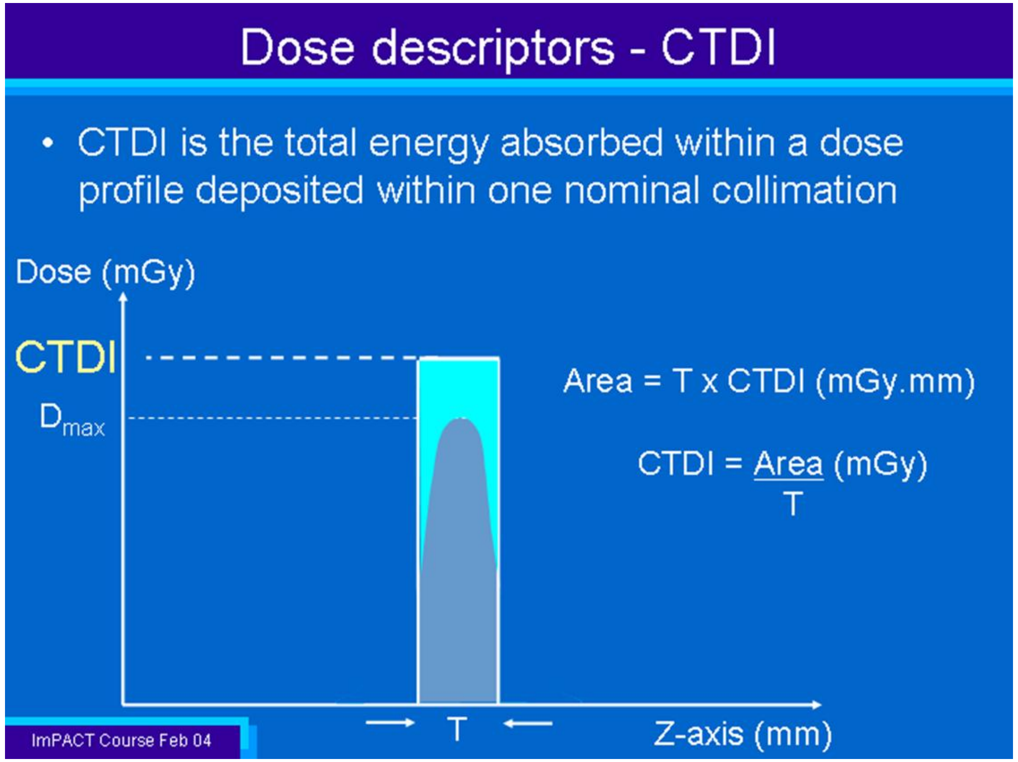

X-ray source
Dose distribution
Pencil
ionization
chamber

- **Phantoms for CTDI Measurement:**

- Made of **PMMA (polymethyl methacrylate)**, simulating x-ray attenuation of the human head and body.
- Two standard sizes are used:
    - **Head phantom:** 16 cm in diameter.
    - **Body phantom:** 32 cm in diameter.



- The CT dose index (CTDI) is an approximate measure of the dose received in a single CT section or slice.



CTDI100 is a fixed measurement taken with a 100-mm-long pencil ionization chamber and makes no reference to a specific number of slices.

Because of absorption, dose varies within the CT image across the acquired field of view. **CTDIw** is an internationally accepted, weighted dose index. It is calculated by summing two-thirds of the exposure recorded at the periphery of the field with one-third of the centrally recorded dose. This weighting yields a more accurate dose approximation.

CTDI$_{w}$ is calculated from measurements made with TLDs positioned at the center and periphery of the phantom to account for the variance in dose distribution.
The **CTDI**$_{w}$ is measured utilizing a conventional step-and-shoot mode of axial CT scanning and does not account for the affects of helical scanning on patient radiation dose.

$$\mathrm{CTDI}_{vol} = \frac{\mathrm{CTDI}_w}{\mathrm{Pitch}}$$

CTDI vol is used to approximate the radiation dose for each section obtained during a helical scan.

- **CTDIvol is the dose index reported on the console. It incorporates pitch (overlap or gap) but remains a standardized phantom-based metric, not an exact patient dose.**

- In the lower right dose panel, you see:

- **CTDIvol:** 41.91 mGy (for 32-cm body phantom)
- **DLP:** 41.91 mGy·cm
- **Scan length:** 10 mm (1 cm)
- **mAs:** 40

Current Image – Helical Abdominal Scan

•Here, **CARE Dose 4D** *is active*, clearly visible on the screen.

•ATCM adjusts **tube current (mA)** in real time based on patient attenuation:

- **Angular modulation:** Adjusts during rotation (lateral vs. AP thickness).
- **Longitudinal modulation:** Adjusts along z-axis (chest → abdomen → pelvis).

•Result: Dose is optimized for patient size and region — **lower in thin areas**, **higher in dense areas** — reducing total dose while maintaining uniform image noise.

In the second image, the **DLP of 188.47 mGy·cm** represents the **total radiation output for the entire scanned abdomen**, calculated by multiplying **CTDIvol (3.87 mGy)** by the **scan length (~48 cm)**—a standardized measure of the overall dose delivered along the z-axis.

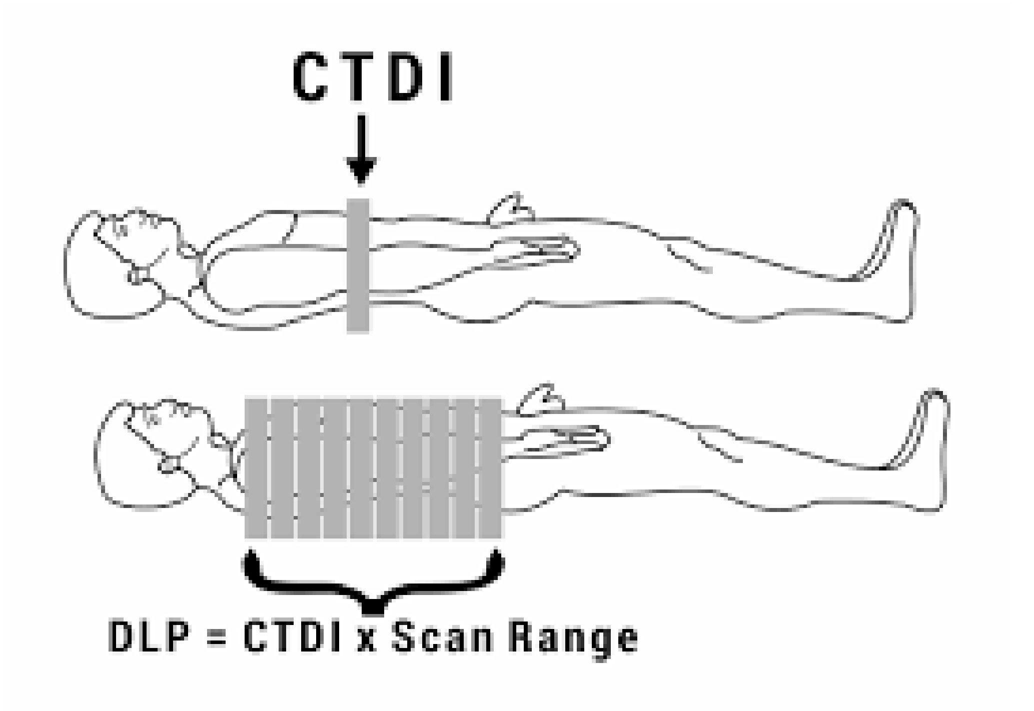
CTDI Accounting for Overlap and Gaps

Scanning Mode	Parameter Defining Overlap/Gaps	Does CTDI Account for It?	How It's Considered	Effect on Reported Dose
Axial (Conventional)	**Table Increment vs. Slice Thickness**	**Partially**	$CTDI_{100}$ measures one rotation; does **not automatically adjust** for overlapping or gapped slices. To represent total dose, $CTDI_{100}$ must be multiplied by the **scan-to-scan ratio** (e.g., overlap or gap correction).	Overlapping slices → ↑ total dose; gapped slices → ↓ total dose. CTDI itself remains constant per rotation.
Helical (Spiral)	**Pitch (Table travel / Beam width)**	**Yes (via CTDIvol)**	CTDIvol = CTDIw / Pitch. Lower pitch (<1) → overlap → ↑ CTDIvol; higher pitch (>1) → gaps → ↓ CTDIvol.	CTDIvol automatically adjusts for overlap or gap in continuous table motion.

DLP — Dose–Length Product

- **Definition:**

The **Dose–Length Product (DLP)** represents the **total radiation output** for an entire CT scan or series.
It combines the **average dose per slice** (**CTDIvol**) with the **scan length (cm)** to estimate the **overall energy delivered along the z-axis**.



Conceptual Meaning

•**CTDIvol** tells us how much dose is delivered *per slice* or *per unit length* of the patient.
•**DLP** multiplies that by how far along the patient we scan — so it reflects **both dose intensity and anatomical coverage**.
•Example:

- CTDIvol = 10 mGy
- Scan length = 40 cm
- → DLP = 10 × 40 = **400 mGy·cm**

What DLP Represents

It's a **measure of total radiation output**, not the actual patient's absorbed dose.
It correlates roughly with the **total energy imparted** to the scanned volume.
It allows **comparison between protocols**, **different scanners**, or **institutions**, even when slice thickness or pitch varies.

Typical Effective Dose Comparison: Radiography vs. CT

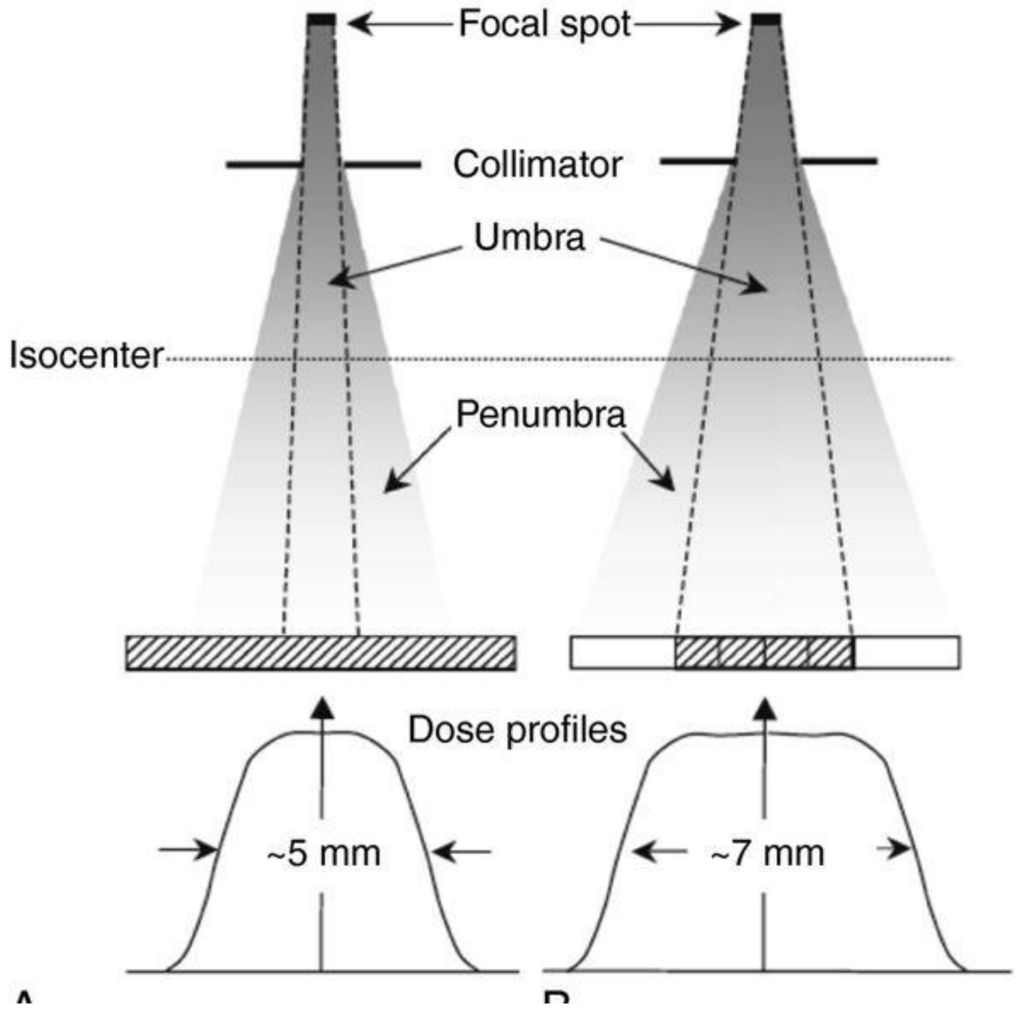
Exam Type / Body Region	Typical Radiography (mSv)	Typical CT (mSv)	Approx. Dose Ratio (CT ÷ X-ray)	Comments / Context
Chest	0.1	6–7	~60–70×	CT chest gives cross-sectional detail of lungs and mediastinum.
Abdomen	0.7	8–10	~12–15×	CT abdomen covers full 3D volume vs. single AP image.
Pelvis	0.6	6–8	~10–13×	Often combined with abdomen as one scan (CT A/P).
Head	0.1	2	~20×	CT head is volumetric, includes brain and bone detail.
Spine (Lumbar)	1.5	6	~4×	CT spine replaces multiple radiographs with 3D coverage.
Sinuses / Facial Bones	0.04	1–2	~25–50×	CT detects fine bone and sinus detail.
Chest–Abdomen–Pelvis (CT CAP)	—	12–18	—	Represents a full torso scan; among higher routine doses.
Low-Dose CT (Lung Screening)	—	1–2	—	Uses lower kVp/mA and iterative reconstruction.

The individual technical configuration of a CT system determines several characteristics that may affect patient dose:

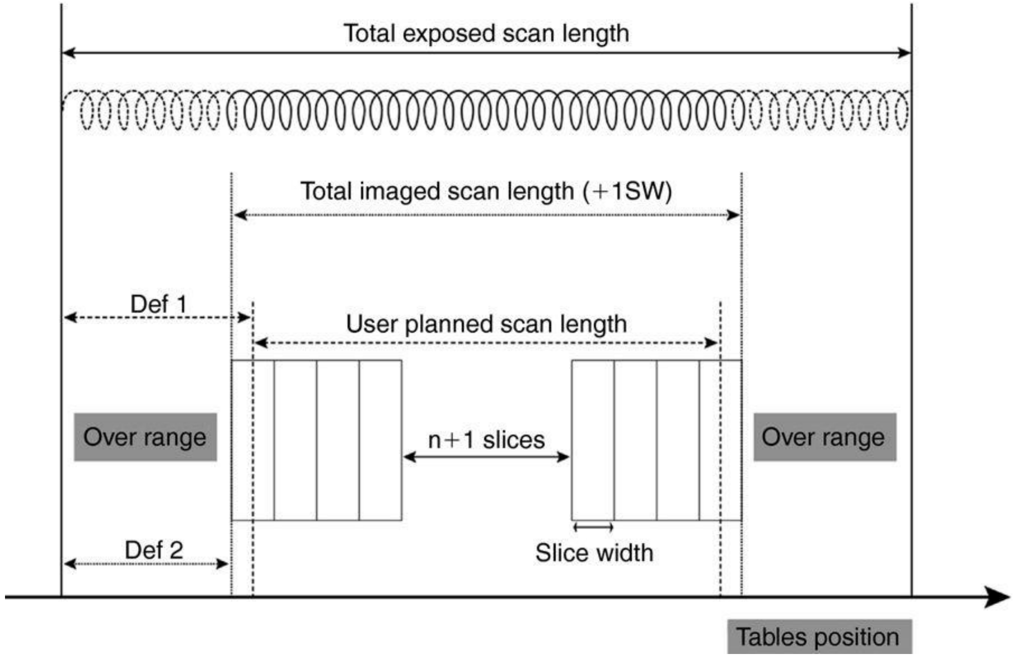
- Source–detector distance: As the distance from the x-ray tube to the detectors decreases, dose increases.
- Filtration
- Detector efficiency
- Beam shape
- Over-beaming
- Over-ranging
- Noise reduction algorithms
- Current modulation
- Patient centering within the gantry (isocenter)
- Physiologic gating
- ASIR

Filtration - additional filtration material along the periphery of the x-ray beam absorbs radiation where it is not necessary, thereby reducing the overall patient dose.

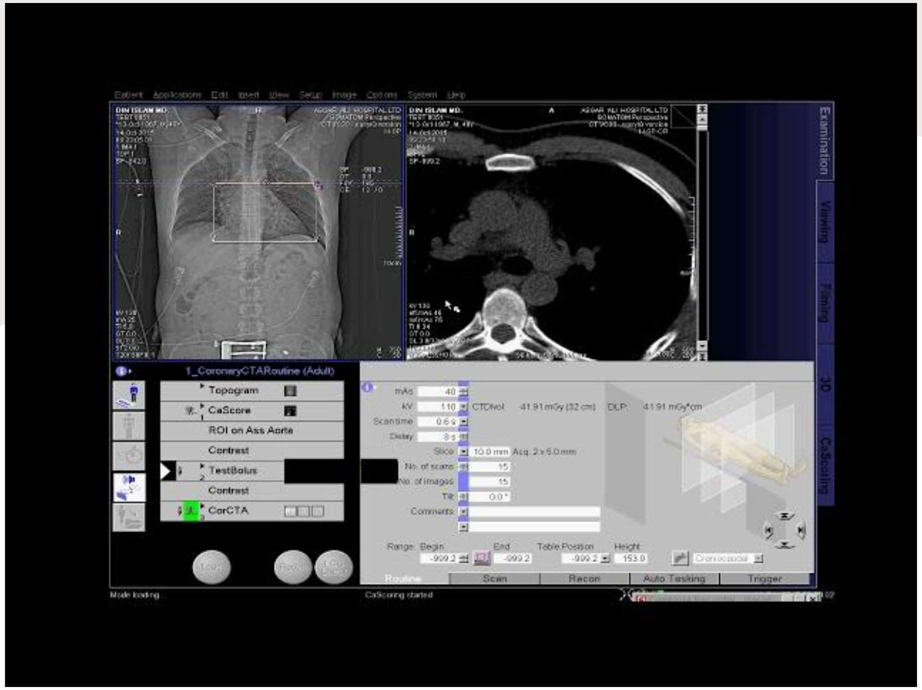
- In order to expose the widened detector array equally, MSCT utilizes a cone-shaped beam, instead of the fan-shaped beam traditionally used by single-slice CT (SSCT) systems. All detectors of the array must be exposed to x-radiation of equal intensity. The beam must be expanded even further to avoid exposing the detectors to undesirable "penumbra." This process is referred to as **overbeaming.**



- **Overranging** occurs when radiation dose is applied before and after the acquisition volume to ensure sufficient data collection for the interpolation algorithms inherent in spiral CT. Up to a half-rotation both before and after the spiral scan are common and add to the patient radiation dose.



- Noise reduction algorithms, or adaptive filters, are employed during the reconstruction process to reduce displayed noise within the CT image. Reduction of displayed noise allows for lower mAs (milliampere-seconds) settings during data acquisition.



- **The term constant mAs** refers to the selection of milliamperage and time (in seconds) separately or mAs on some scanners before the scan begins, keeping all other technical factors constant.

Effective Milliamperage-Seconds

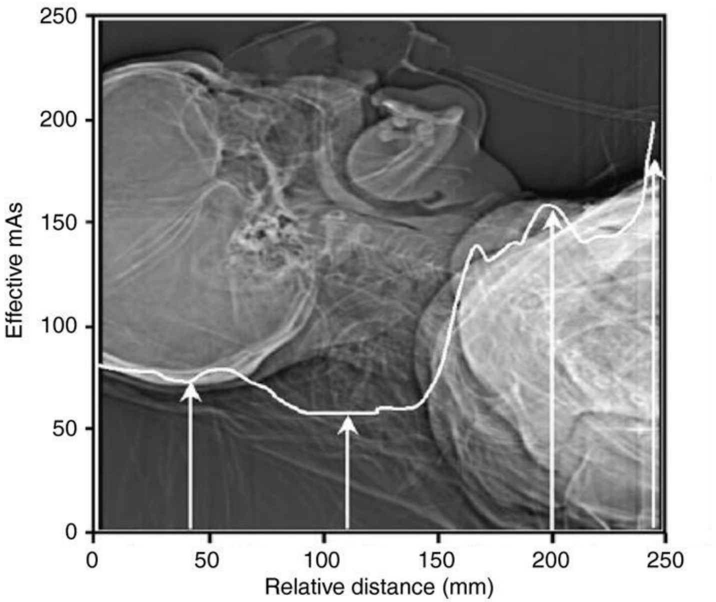
The effective mAs is a term used for multislice CT scanners that denotes the mAs per slice.

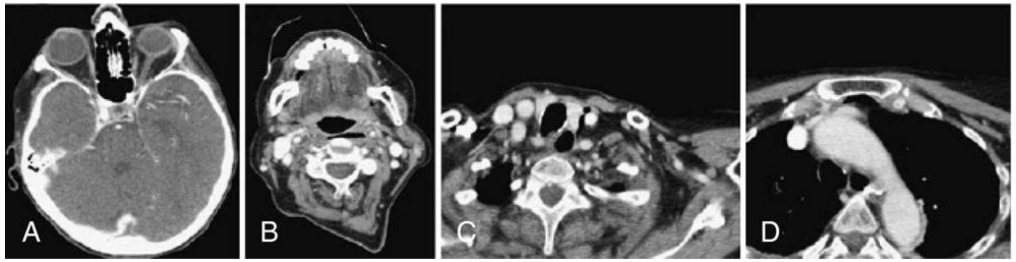
- In CT, ATCM refers to the automatic control of the mA in two directions of the patient (the x-y axis and the z-axis) during data acquisition (scanning process) by use of specific technical procedures that take into consideration not only the patient size but also the attenuation differences of the various tissues.

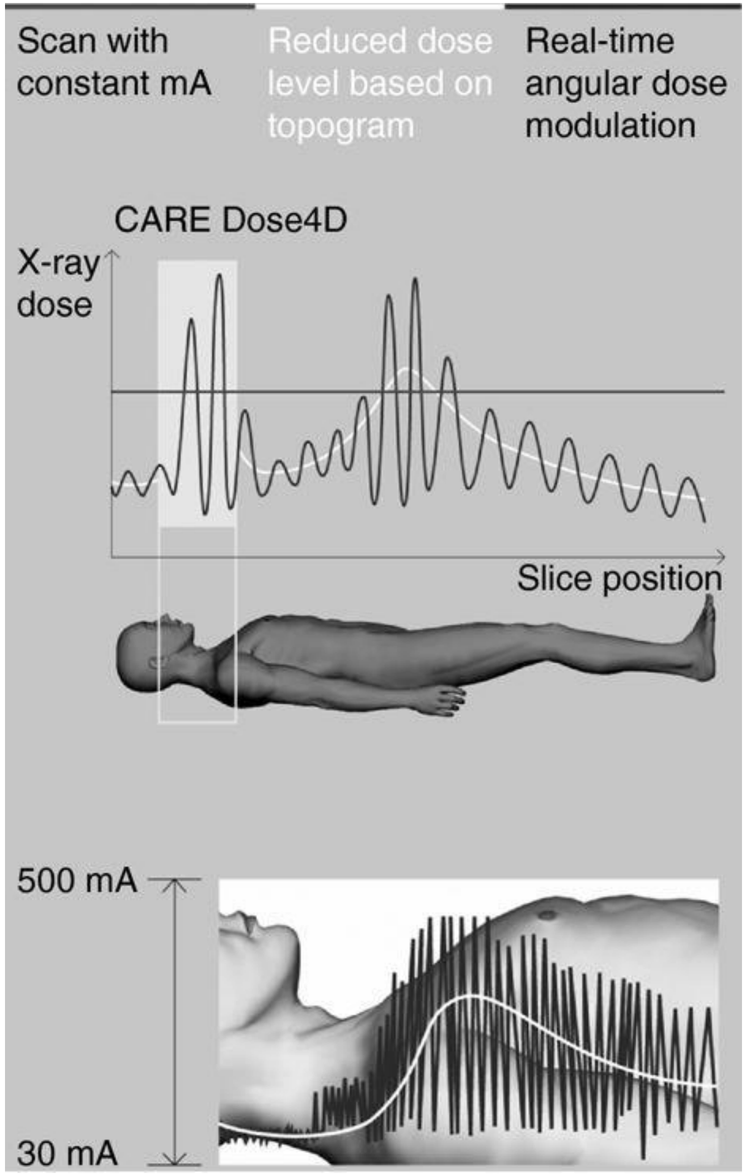
- The overall goal of ATCM is to provide consistent image quality despite the size of the patient and the tissue attenuation differences and to control the dose to the patient

- **Longitudinal (Z-Axis) Tube Current Modulation**

- Longitudinal (z-axis) tube current modulation (z-axis TCM) is based on differences in attenuation among body parts. For example, thicker body parts such as the abdomen and pelvis will attenuate the radiation more than thinner body parts, such as the head, neck, and chest regions. The technique of z-axis TCM is designed (using a specific computer algorithm) to change the mA automatically as the patient is scanned from, say, head to toe (along the z-axis) while maintaining a constant (uniform) noise level (image noise target values) for different thicknesses of body parts examined.





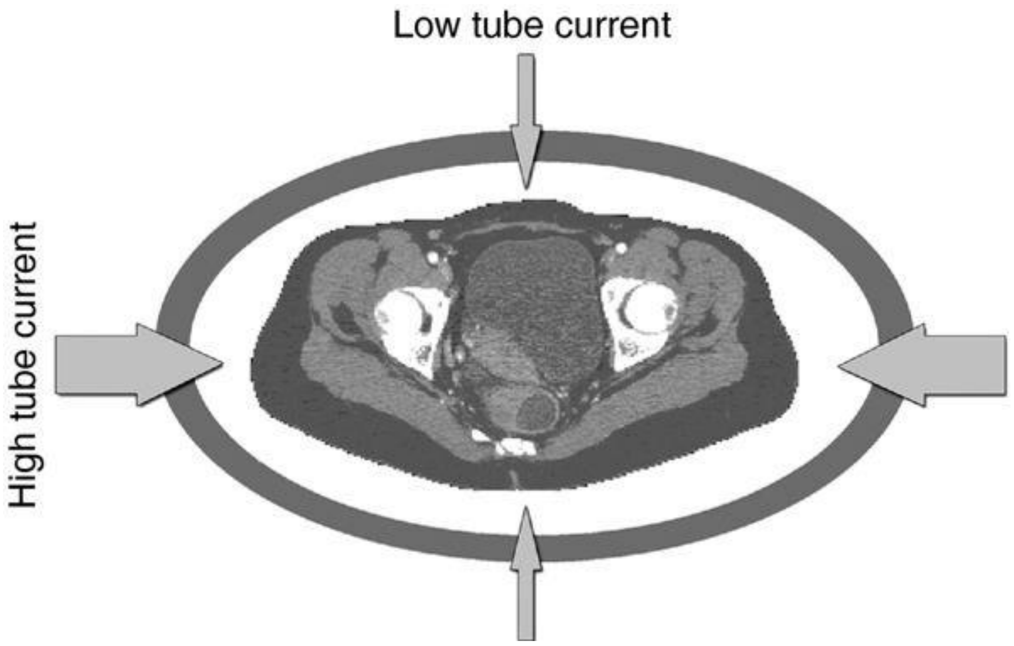


Scan with constant mA
Reduced dose level based on topogram
Real-time angular dose modulation
CARE Dose4D
X-ray dose
Slice position
500 mA
30 mA

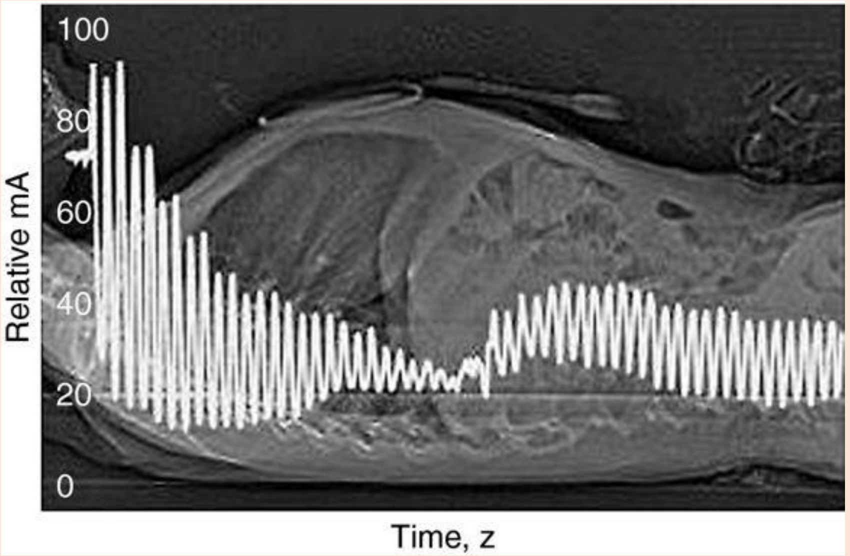
CT Prof. Stelmark

Angular (X-Y Axis) Tube Current Modulation

- Angular (x-y axis) tube current modulation (x-y axis TCM) is based on the fact that the radiation attenuation varies from the anteroposterior (AP) projection (low attenuation) to the lateral projection (high attenuation) as the tube rotates around (gantry rotation) the patient. Although the high attenuation projections will require higher mA values, the low attenuation projections will require lower mA values, as illustrated in Figure 10-15. The x-y axis TCM algorithm ensures that a constant (uniform) noise level is maintained during the scanning process.



Angular-Longitudinal Tube Current Modulation



Summary Table

Feature	Conventional (Axial)	Helical (Spiral)
Table motion	Step-and-shoot	Continuous
ATCM map source	Scout (topogram)	Scout (topogram)
mA variation	Changes **between slices**	Changes **continuousl**
Angular modulation	❌ No	✅ Yes
Effective mAs	Actual mAs per slice (varies slice-to-slice)	Average mAs per slice

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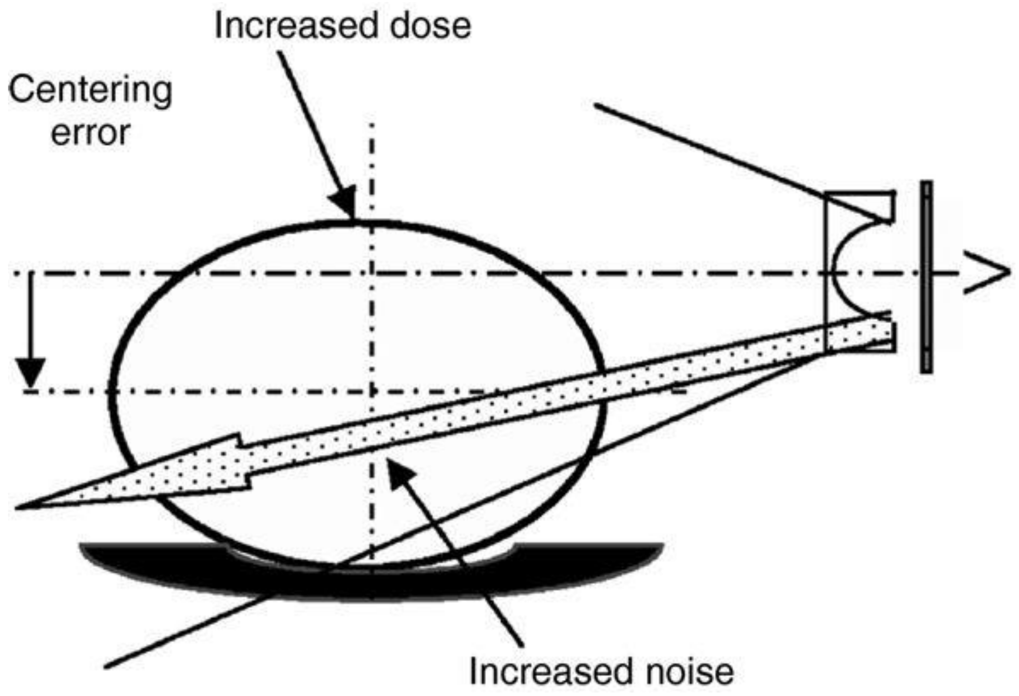
- A patient who is miscentered in the scan field of view can be expected to have degraded bowtie filter performance with an undesired increase in both dose and noise



During MSCT cardiac studies, prospective gating can be used to reduce the patient radiation dose. ECG-triggered tube current modulation allows for pulses of x-ray energy rather than continuous exposure to be used.

Radiation Protection Actions

Radiation protection actions include the use of time, shielding, and distance, which are intended to protect both patients and personnel in radiology. For example, because dose is proportional to the **time** of exposure, to protect personnel in CT, it is essential to minimize the time spent in the CT scan room during the exposure.

Distance, on the other hand, is a major dose reduction action because the dose is inversely proportional to the square of the distance. This means that the further one is away from the radiation source, the less the dose received. In CT, because the patient is the main source of scatter, technologists should stand back as far away from the patient as possible if there is a need to be present in the scan room during scanning. This implies that the use of a power injector that can becontrolled from outside the scan room is recommended. If a hand injector is used, then a long tubing should be used.

- Shielding is intended to protect not only patients (gonadal, breast, eyes, and thyroid shielding) but also personnel and members of the public. Patients are often concerned about the exposure of their gonads during a CT examination. Because most of the gonadal exposure will come from internal scatter and not from the primary beam (unless the gonadal region was being examined), there is no need for this concern.

- Technologists, however, could place gonadal shielding on the patient because it may alleviate any fears about the risks of being exposed to radiation. Shields can also be used to protect the eyes, breast, and thyroid of patients undergoing CT examinations .

- Although some of the shields are made of flexible, **nonlead (bismuth)** composition, lead may also be used to provide significant dose reduction to these critical organs.

- Bismuth shields (for breast, thyroid, or eye protection) are designed to reduce radiation to superficial organs during the actual CT scan. However, they should never be placed on the patient before the scout (topogram) is acquired. Why Not During the Scout?

- Automatic Tube Current Modulation (ATCM) uses the scout image to analyze patient size, shape, and density. If a bismuth shield is present during the scout, the system "sees" an area of artificially high attenuation (appears very dense). The scanner will then increase tube current (mA) in that region to compensate, thinking the area is thicker tissue. This results in a higher overall dose, completely defeating the purpose of the shield and potentially increasing exposure to adjacent tissues.

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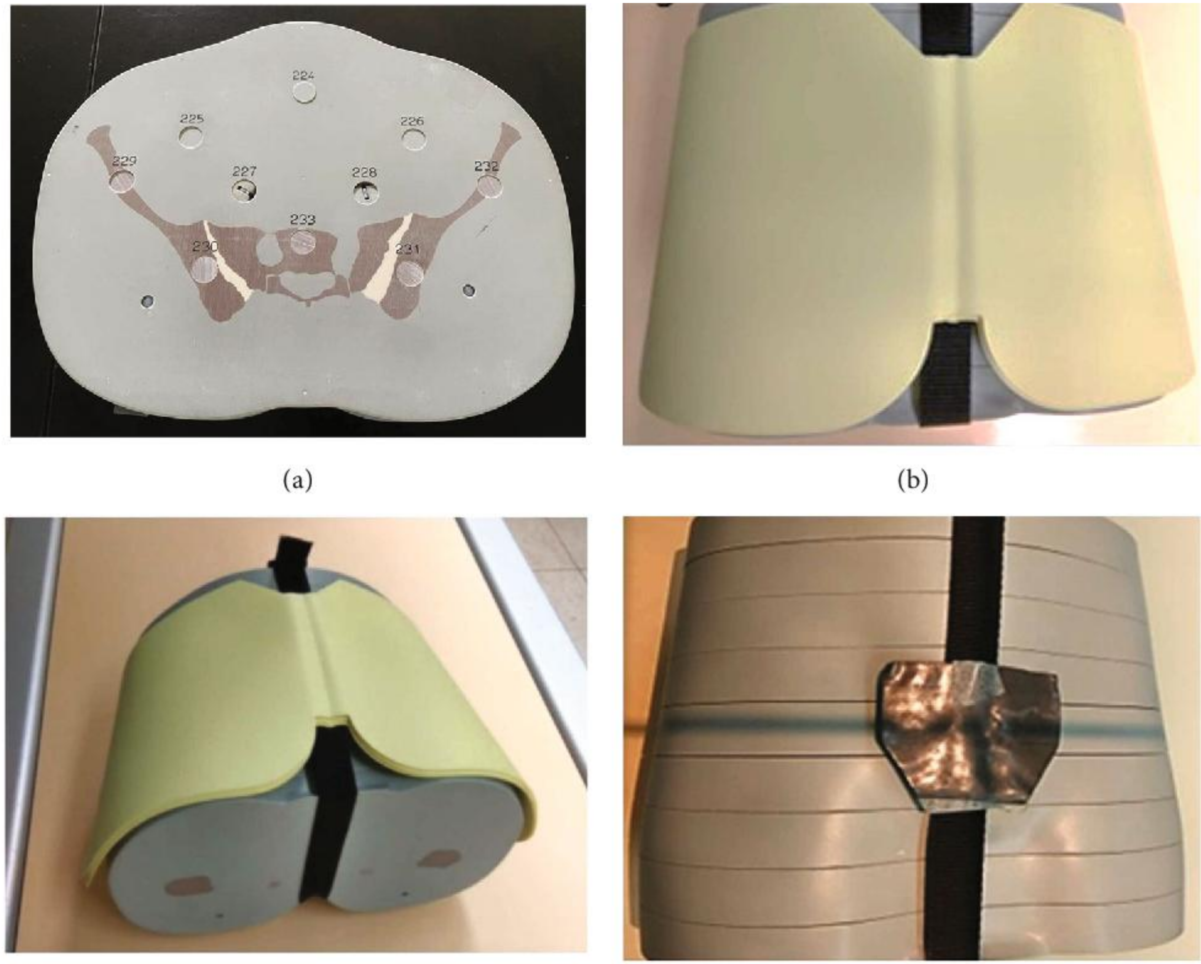

(a) (b)

(c) (d)

Role of Iterative Reconstruction (IR) and AI in Low-kVp Imaging (CTA 80 kVp)

1. **The Challenge Without IR/AI**

•When we lower kVp to 80–100, photon energy drops → **less penetration** and **more noise**.

•To fix that, the scanner automatically **raises mA**, sometimes increasing dose.

•Before IR existed, lowering kVp was risky — you often lost image quality or negated dose savings

Iterative Reconstruction (IR): The Game Changer

IR algorithms (ASiR, iDose, SAFIRE, ADMIRE, etc.) repeatedly **model and refine the image**, removing random noise while preserving real edges.

This allows scanners to **maintain diagnostic quality** at **much lower kVp and mAs**.
In practice: you can drop kVp to 80 or 100 and still get clean, high-contrast angiograms — **dose ↓ 30–60%** compared to filtered back projection (FBP)

AI-Based Reconstruction (Deep Learning IR)

The newest systems use deep convolutional neural networks trained on millions of image pairs (low-dose vs. full-dose).They recognize and suppress noise patterns far better than model-based IR.Effect:Sharper details, natural texture (less “plastic” look)Even greater dose savings — up to 70–80% in CTA and low-dose protocolsEnables 80 kVp or even 70 kVp imaging for slim patients with excellent quality.

Parameter	FBP (Old)	Iterative (IR)	AI / Deep Learning IR
Noise control	Poor	Good	Excellent
Dose reduction	Limited	30–60%	60–80%
Image texture	Grainy	Smooth / plastic	Natural
Feasible low kVp	≥ 100 kVp	80–100 kVp	70–100 kVp