BIOMEDICAL INSTRUMENTATION

(24BM502)

UNIT IV

MEDICAL IMAGING AND DIAGNOSTIC TECHNOLOGIES

Topics Covered

1. Principles of Medical Imaging
2. X-Ray Imaging Systems
3. Computed Tomography (CT)

1. Principles of Medical Imaging

Medical imaging is the set of techniques used to create visual representations of the interior of the human body for clinical analysis and medical intervention. Every imaging modality works on the same basic idea: some form of energy is sent into or through the body, the tissues interact with that energy in a way that depends on their physical or chemical properties, and the resulting signal is captured and converted into an image that a clinician can interpret.

1.1 The Electromagnetic Spectrum in Imaging

Most medical imaging modalities use electromagnetic (EM) radiation of some form. The EM spectrum spans a huge range of wavelengths and frequencies, and only certain bands are useful for imaging the body.

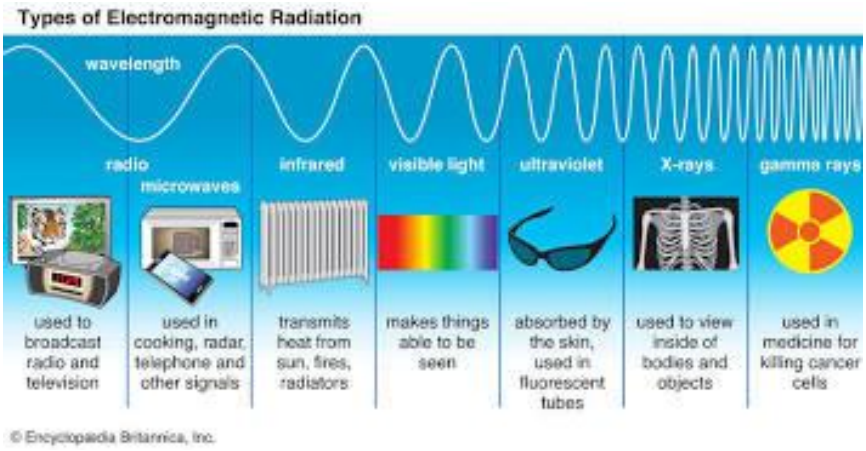


Fig. 1.1 The electromagnetic spectrum, from radio waves to gamma rays

- **Radio waves:** longest wavelength, lowest frequency; used in broadcasting and, notably, in MRI signal detection
- **Microwaves:** shorter than radio waves; used in radar and satellite communication
- **Infrared (IR):** felt as heat; basis of thermal/infrared imaging
- **Visible light:** the narrow band the human eye can detect; used in endoscopy and optical imaging
- **Ultraviolet (UV):** higher energy than visible light; limited direct diagnostic imaging use
- **X-rays:** high-energy radiation capable of penetrating soft tissue; the basis of radiography and CT
- **Gamma rays:** shortest wavelength, highest energy; emitted by radioactive decay and used in nuclear medicine (PET, SPECT)

1.2 How a Diagnostic Image Is Formed

Although the energy source differs across modalities, the creation of a spatial image depends on differential interaction of that energy with different tissue types. In projection X-ray imaging, this process can be broken into three stages:

- **Density variation:** internal structures such as bone, soft tissue, tumours, or air pockets differ in physical density

- **Differential absorption:** as radiation passes through the body, denser structures absorb (attenuate) more of it, while less dense structures allow more to pass through
- **Emerging pattern:** the radiation leaving the body carries a spatial pattern of intensity variations — effectively an "invisible shadow image" of the internal anatomy — which is then captured by a detector or film

This same logic of differential interaction, followed by detection and image reconstruction, underlies every modality studied in this unit — only the type of energy and the physical interaction mechanism change.

1.3 Overview of Imaging Modalities

Modality	Energy Used	What It Best Reveals
X-Ray	Ionising EM radiation (X-rays)	Bone detail, quick 2-D projection images
CT	Ionising EM radiation (X-rays), computer-reconstructed	3-D cross-sectional anatomy, excellent bone detail
MRI	Radiofrequency waves + strong magnetic field (non-ionising)	Excellent soft-tissue contrast
Ultrasound	Mechanical sound waves (non-ionising)	Real-time soft-tissue and flow imaging
PET	Gamma rays from an injected radiotracer	Functional / metabolic activity

1.4 Ionising vs. Non-Ionising Radiation

The energy used for imaging is classified by whether it carries enough energy to remove electrons from atoms (ionisation). This distinction is the basis of radiation-safety practice in every imaging department.

- **Ionising radiation:** X-rays and gamma rays carry enough energy to ionise atoms and can damage cell DNA; the dose delivered to the patient must therefore be justified and kept as low as reasonably achievable (the ALARA principle).
- **Non-ionising energy:** radio waves (MRI) and ultrasound (mechanical sound waves) do not ionise tissue and have no known cumulative radiation hazard at diagnostic levels, which makes them preferred for repeated studies and for pregnancy and paediatric imaging.

1.5 Image Quality Parameters

Whatever the modality, the quality of a diagnostic image is judged by the same few parameters. Improving one usually costs something in another, which is why imaging protocols are always a compromise.

Parameter	Meaning	Influenced by
Spatial resolution	Ability to separate two small, closely spaced structures	Focal spot / detector size, pixel size, probe frequency
Contrast resolution	Ability to distinguish tissues with very similar properties	Beam energy, scatter, tissue properties, contrast agents

Parameter	Meaning	Influenced by
Noise	Random variation in pixel values that grainy-ises the image	Dose (photon count), detector sensitivity, reconstruction filter
Signal-to-noise ratio (SNR)	Strength of useful signal relative to noise	Increases with dose, scan time, field strength
Artifacts	Features in the image that do not represent real anatomy	Patient motion, metal, hardware faults, physics limits
Temporal resolution	How quickly images can be acquired	Scan speed, frame rate, gantry rotation time

Key Takeaways

- Medical imaging works by sending energy into the body and capturing how tissues modify that energy.
- The electromagnetic spectrum provides the energy for X-ray, CT, and PET imaging; MRI uses RF waves with a magnetic field; ultrasound uses mechanical sound, not EM radiation.
- Differential absorption of radiation by tissues of different density is the core principle behind X-ray and CT image formation.
- Each modality trades off between radiation exposure, soft-tissue contrast, spatial resolution, cost, and speed — the reason several modalities coexist in clinical practice.

2. X-Ray Imaging Systems

An X-ray imaging system generates a beam of X-rays, shapes and limits that beam, passes it through the patient, and records the emerging pattern on a detector or film. This section covers the major components of a conventional X-ray system: the X-ray tube, the collimator, the anti-scatter grid, and the X-ray film.

2.1 The X-Ray Tube

The X-ray tube is the source of the X-ray beam. Electrons are generated at a heated cathode filament and accelerated across a vacuum to strike a metal target (the anode), where their sudden deceleration produces X-rays. Two anode designs are used, depending on the power requirement of the machine.

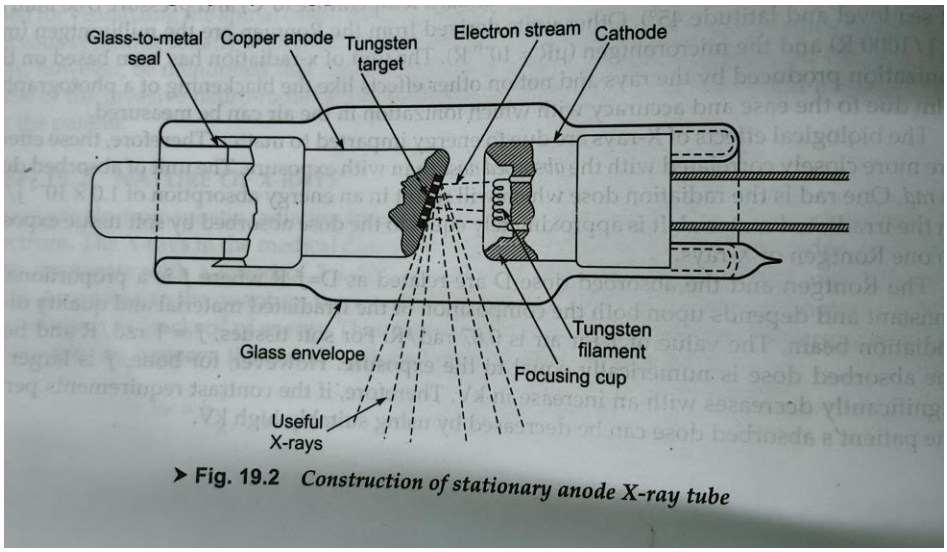


Fig. 2.1 Construction of a stationary-anode X-ray tube

Stationary Anode

- Consists of a small tungsten target (about 15 mm × 20 mm × 3 mm) soldered into a copper block.
- Tungsten is used for its high atomic number (74), which improves X-ray production efficiency, and its very high melting point (3400°C).
- Copper is an excellent thermal conductor and rapidly carries heat away from the target to be cooled by surrounding oil.
- Used primarily in low-capacity, low-power machines.

Rotating Anode

- Designed for high-power systems, where heat dissipation is the major limiting factor.
- A rotating disk of tungsten or a tungsten–rhenium alloy (which resists thermal stress) spins at 3,000–3,600 rpm, or up to 9,000–10,000 rpm in high-speed designs.
- Rotation continuously brings a fresh, cool region of the target into the electron beam path, instead of concentrating heat at one spot.
- A bevelled edge (typically 5°–20°) exploits the line focus principle — the effective focal spot stays small (for sharp images) while the heat is spread over a larger physical track on the disk.

2.2 The Collimator

The collimator is a beam-limiting device placed between the X-ray tube and the patient. It performs two vital functions in diagnostic radiology: reducing patient radiation dose and increasing image contrast, by restricting the X-ray beam to only the anatomical region of clinical interest.

- **Lead shutters:** adjustable lead strips (long shutters and cross shutters) that can be moved relative to one another to shape the beam into any required rectangular field size
- **Entrance barriers:** a conical entrance barrier plus a fixed barrier near the tube housing port, which block off-focus radiation (radiation arising from parts of the anode other than the true focal spot) before it can escape the housing
- **Optical positioning simulator:** a light source inside the collimator that projects a visible light field exactly matching the invisible X-ray field, letting the radiographer align the beam accurately before exposure

2.3 Anti-Scatter Grids

As the X-ray beam passes through the patient, some photons are scattered in random directions. Scattered radiation that reaches the film reduces image contrast. A grid — a thin structure of alternating radiopaque (lead) strips and radiolucent interspace material, placed between the patient and the film — absorbs much of this scattered radiation while allowing the primary (useful) beam through.

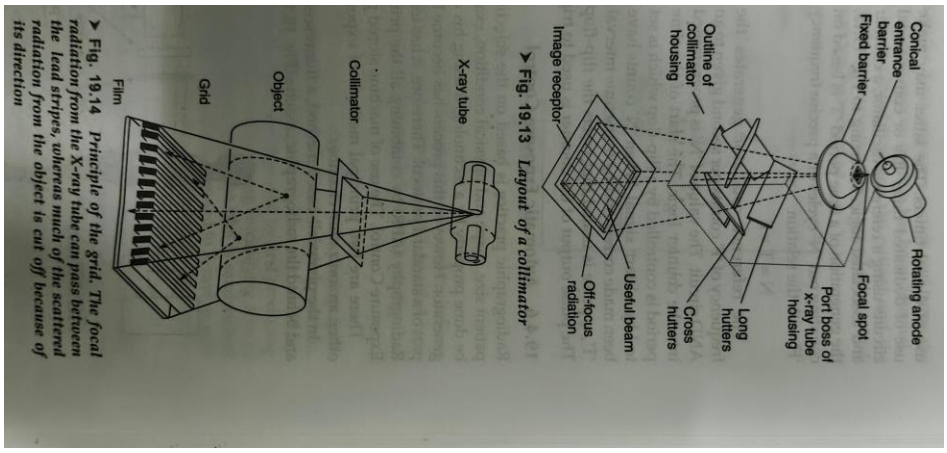


Fig. 2.2 Layout of a collimator (top) and principle of the anti-scatter grid (bottom)

Types of Grids

- **Stationary grid:** fixed in position during exposure; simple, but the lead strips can produce faint grid lines on the image
- **Movable grid (Potter-Bucky diaphragm):** moves during the exposure so the grid lines are blurred out and become invisible on the final image

Grids are further classified by the arrangement of their lead strips:

- Linear / parallel grid — strips run parallel to one another.
- Focused grid — strips are angled so their projections converge at a specific focal distance, matching the diverging X-ray beam.
- Crossed grid — two linear grids stacked at 90° to each other, giving very effective scatter clean-up but requiring precise beam alignment.

2.4 Ultrastructure of X-Ray Film

Conventional X-ray film is a layered structure, built up on both sides of a supporting base so that the emulsion can be exposed from either side.

Layer	Also Known As	Role
Base	—	Supporting layer; strong, transparent, flexible; resistant to moisture and chemicals; usually polyester (polyethylene terephthalate)
Subbing layer	Adhesive layer	A mixture of gelatin and a solvent of the base; bonds the base to the emulsion layer; dyes are sometimes added to prevent the crossover effect
Emulsion layer	—	Where the image is produced and stored; tiny silver halide grains dispersed in gelatin — larger grains give greater sensitivity
Supercoat	Anti-abrasive layer	A thin layer of clear gelatin; protects the emulsion from damage and gives the film its glossy surface finish

2.5 Production of X-Rays and Exposure Controls

When the fast electrons strike the anode, less than 1% of their energy becomes X-rays; more than 99% is converted to heat, which is why anode cooling is such a central design problem. Two processes produce the X-ray photons:

- **Bremsstrahlung (braking) radiation:** an electron is slowed and deflected by the nucleus of a tungsten atom and loses energy as an X-ray photon; this produces a continuous spectrum of photon energies and forms most of the beam.
- **Characteristic radiation:** an incoming electron knocks out an inner-shell electron of a tungsten atom; an outer-shell electron drops into the vacancy and releases a photon of a specific, fixed energy, giving sharp peaks on the spectrum.

Controlling the Exposure

- **kVp (kilovoltage peak):** the tube voltage; it sets the maximum photon energy and so the penetrating power of the beam. Higher kVp penetrates more but reduces subject contrast.
- **mAs (milliampere-seconds):** tube current multiplied by exposure time; it sets the quantity of X-rays produced and therefore image blackening, noise, and patient dose.

2.6 Intensifying Screens and Digital Radiography

Film alone absorbs only a small fraction of the X-ray beam, so film is normally sandwiched between intensifying screens. Modern departments have largely moved to digital detectors that remove chemical processing altogether.

- **Intensifying screens:** phosphor layers (for example rare-earth compounds) convert each absorbed X-ray photon into many visible-light photons that expose the film, allowing a much lower patient dose.

- **Computed radiography (CR):** a reusable photostimulable phosphor plate stores the latent image; a laser scanner later releases it as light, which is digitised.
- **Digital radiography (DR):** a flat-panel detector converts X-rays directly to an electronic signal. Indirect detectors use a scintillator (such as caesium iodide) plus a photodiode array; direct detectors use amorphous selenium to convert X-rays straight into charge.

System	How the image is recorded	Key advantage / drawback
Film–screen	Screen light exposes silver halide film; chemical development	Mature and cheap; fixed contrast, no post-processing, chemicals needed
CR	Phosphor plate read out by laser in a separate reader	Works with existing machines; slower workflow than DR
DR	Flat-panel detector gives the image within seconds	Fast, wide dynamic range, lower dose; high equipment cost

2.7 Clinical Uses and Radiation Protection

- **Common uses:** chest radiography (lungs, heart size), bone and joint fractures, dental and skull films, abdominal studies, mammography, and fluoroscopy for real-time imaging during contrast studies and interventional procedures.
- **Protection:** the three basic protection principles are minimum time, maximum distance from the source, and adequate shielding (lead aprons, thyroid collars, lead-glass screens); collimation, proper kVp/mAs selection, and checking for pregnancy before exposure further reduce dose.

Key Takeaways

- The X-ray tube produces X-rays by decelerating electrons in a tungsten target; over 99% of the energy becomes heat, hence the rotating anode.
- The collimator limits the field, the grid removes scatter, and the film or digital detector records the emerging intensity pattern.
- kVp controls beam penetration and contrast, while mAs controls the quantity of radiation and noise.
- Digital radiography (CR and DR) has largely replaced film–screen systems because of faster workflow and wider dynamic range.

3. Computed Tomography (CT)

3.1 Basic Principle

Computed Tomography (CT) is a non-invasive imaging technique that combines X-ray measurements with computer algorithms to generate cross-sectional images of the body. The name reflects the process: "Computed" because data from many X-ray measurements are processed mathematically, and "Tomography" because the body is imaged by sections, or slices. Unlike conventional X-rays, which produce a single flattened 2-D projection, CT reconstructs full 3-D volumetric data, displayed as a stack of 2-D cross-sectional slices, by acquiring projections from many angles around the patient.

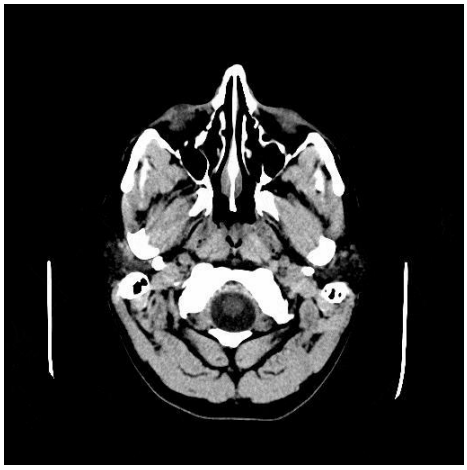


Fig. 3.1 Axial CT slice through the head — one image in a stacked 3-D dataset

3.2 Why CT? Limitations of Conventional Radiography

- **Inefficient X-ray absorption:** older film/screen cassettes absorbed only about 25% of the X-ray beam, wasting most of the dose and information
- **High scatter-to-primary ratio:** scattered photons often made up more than 50% of detected X-rays, reducing subject contrast
- **Superimposition and conspicuity:** projecting a 3-D volume onto a 2-D image causes overlying structures to obscure one another
- **Contrast vs. latitude trade-off:** film needs a wide exposure latitude, which necessarily limits how much contrast it can display

3.3 Attenuation, Hounsfield Units, and Windowing

As X-rays pass through the body, each tissue absorbs (attenuates) the beam by a different amount depending on its density: high-density tissue such as bone absorbs strongly, so a reduced signal reaches the detector, while low-density tissue such as lung or air absorbs weakly, giving a greater detected signal. Each tissue's attenuation is expressed as a CT number, measured in Hounsfield Units (HU) — named after Sir Godfrey Hounsfield, inventor of the first clinical CT scanner (1971).

- **Reference points:** water = 0 HU, air ≈ −1000 HU, dense bone ≈ +1000 HU

The HU scale is calculated as $HU = ((\mu(material) - \mu(water)) / \mu(water)) \times K$, where μ is the attenuation coefficient of the voxel (or of water), and K is a standardised scaling constant (1,000).

A typical display can show far fewer grey shades (roughly 100–250) than the full HU range (−1000 to +1000), so many tissues would otherwise look identical. Windowing lets the viewer select a window width (the range of HU displayed) and a window level (the centre HU value), letting soft tissue, lung, or bone detail be brought out interactively for the clinical question at hand.

3.4 Scanning Geometry and Image Reconstruction

- **Scanning geometry:** in early designs, a narrow X-ray beam and a linked detector scan across the patient in a linear translation, sweeping through one axial slice; the X-ray path for one measurement is a ray, and the full set of rays from one translation is a view; after each translation the tube–detector assembly rotates by a small angle and repeats — the translate–rotate motion — with views collected over 180–360° providing sufficient data for reconstruction
- **Voxels and attenuation sums:** each slice is treated as a matrix of small volume elements (voxels), each with its own attenuation coefficient μ; for a ray through a row of voxels, the measured transmission relates to the sum of the attenuation values along its path ($X_i = \mu_1 + \mu_2 + \ldots + \mu_n$, where $X_i = -\ln(N_i/N_0)$); enough such equations, from many rays at many angles, allow every voxel's attenuation to be solved for
- **Algebraic Reconstruction (ART):** an iterative approach — start from an estimate, compare it against each new view's actual measurements, and distribute any mismatch as a correction along the ray, refining the estimate view by view until predicted and actual measurements agree within tolerance
- **Backprojection and Filtered Backprojection (FBP):** simple backprojection divides each view's measurements evenly back along the ray path and sums across all angles — fast, but blurry; filtering (convolution) each view before backprojection sharpens the smeared edges; this combination, FBP, remains the classic reconstruction algorithm for slice-by-slice CT

3.5 Four Generations of CT Scanner Design

Generation	Geometry / Motion	Key Feature
First (1971)	Single pencil beam + single detector, translate–rotate	Scan time 5–6 min/slice; 160 rays × 180 views per slice
Second (1974)	Multiple narrow beams/detectors, translate–rotate	Scan time cut to ~20 s; enabled breath-hold body scanning
Third (1975)	Wide fanbeam, tube + curved detector rotate together	Scan time under 5 s; prone to ring artifacts from detector drift
Fourth (1976)	Stationary detector ring, only tube rotates	Avoids ring artifacts; large ring, high detector count, costly
Helical / Spiral	Slip-ring gantry + continuous table motion	Continuous helical data acquisition; interpolation reconstructs in-plane data

Slip-ring technology transfers electrical power to the rotating gantry without fixed cables, enabling continuous, non-stop rotation. Combined with continuous table movement, this traces a helical (spiral) path through the patient. The helical pitch (table movement per rotation ÷ slice thickness) governs the trade-off between scan coverage and accuracy.

3.6 Clinical Applications

- **Contrast imaging:** IV contrast highlights the cardiovascular system (aneurysms, dissections, atherosclerotic disease) and helps characterise tumours; CT urography evaluates the urinary tract; oral contrast evaluates the digestive tract
- **Intracranial haemorrhage:** CT is the primary investigation for suspected brain bleeds because of its speed and sensitivity to acute blood — extradural (lentiform, arterial, trauma-related), subdural (crescent-shaped, torn bridging veins), subarachnoid (ruptured aneurysm), and intracerebral (localised, from hypertension or trauma) bleeds each have a characteristic appearance

3.7 CT Compared with Other Modalities

Factor	CT (abdomen)	MRI	X-ray (chest)	Ultrasound
Duration	3–7 min	30–45 min	2–3 min	5–10 min
Cost	Cheaper	Expensive	Cheap	Cheap
Dimensions	3-D	3-D	2-D	2-D
Soft-tissue detail	Poor–moderate	Excellent	Poor	Poor
Bone detail	Excellent	Poor	Excellent	Poor
Radiation dose	~10 mSv	None	~0.15 mSv	None

3.8 Common CT Image Artifacts

Because CT images are reconstructed from measurements, errors in the data appear as characteristic artifacts. Recognising them avoids misdiagnosis.

Artifact	Appearance	Typical cause
Beam hardening	Dark streaks and "cupping" (centre darker than edges)	Low-energy photons absorbed first, so the beam becomes harder as it crosses dense bone
Motion	Blurring or double edges	Patient movement, breathing, heartbeat
Metal	Bright and dark star-shaped streaks	Dense implants, dental fillings, surgical clips
Ring	Concentric rings in the slice	A faulty or mis-calibrated detector in a third-generation scanner
Partial volume	Blurred edges, mixed density values	More than one tissue type inside a single voxel

3.9 Radiation Dose and Safety in CT

- **Dose:** a CT examination delivers a considerably higher dose than a plain radiograph (see the comparison table above), so every scan must be clinically justified.

- **Dose reduction:** adjusting tube current and voltage to patient size, limiting the scanned length, using iterative reconstruction, and avoiding repeat scans.
- **Contrast safety:** iodinated contrast is checked against known allergy and kidney function before injection, and patients are observed afterwards for reactions.

Key Takeaways

- CT reconstructs cross-sectional slices from X-ray projections taken at many angles; image values are expressed in Hounsfield Units and viewed through adjustable windows.
- Scanner design evolved from slow translate–rotate systems to fast helical scanning with slip rings and continuous table motion.
- CT gives excellent bone and acute-bleed detail in minutes, at the cost of higher radiation dose than plain X-ray and weaker soft-tissue contrast than MRI.