Chapter 3: X-ray Radiography and Computer Tomography

X-ray Physics

Contents

A Brief history

X-ray terminology

X-ray generation

- Bremsstrahlung and characteristic X-rays
- Basic structure of X-ray tube
- Practical considerations

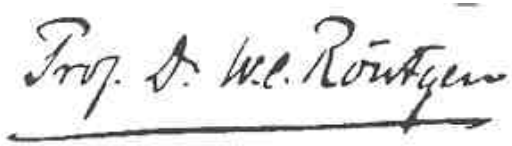
Attenuation of X-rays photons in media

- Interaction of X-rays
- Measures of X-ray attenuation

X-ray Terminology

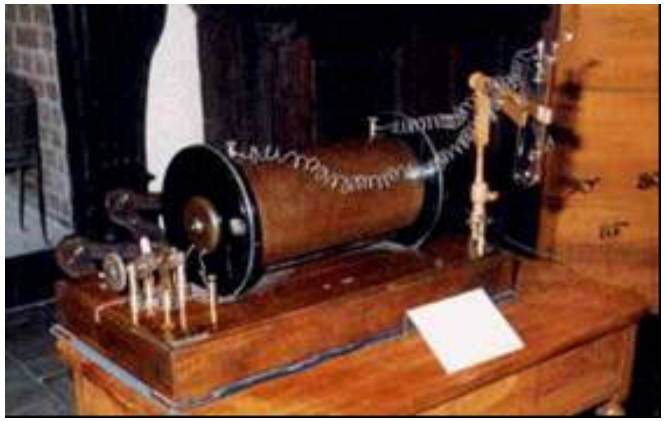
- Film Radiography
- Fluoroscopy
- Digital Radiography
- Angiography
- Computed Tomography

Wilhelm Röntgen, Wurtzburg



Nov. 1895	– Announces X-ray discovery
Jan. 13, 1896	– Images needle in patient's hand
	– X-ray used presurgically
1901	– Receives first Nobel Prize in Physics
	– Given for discovery and use of X-rays.

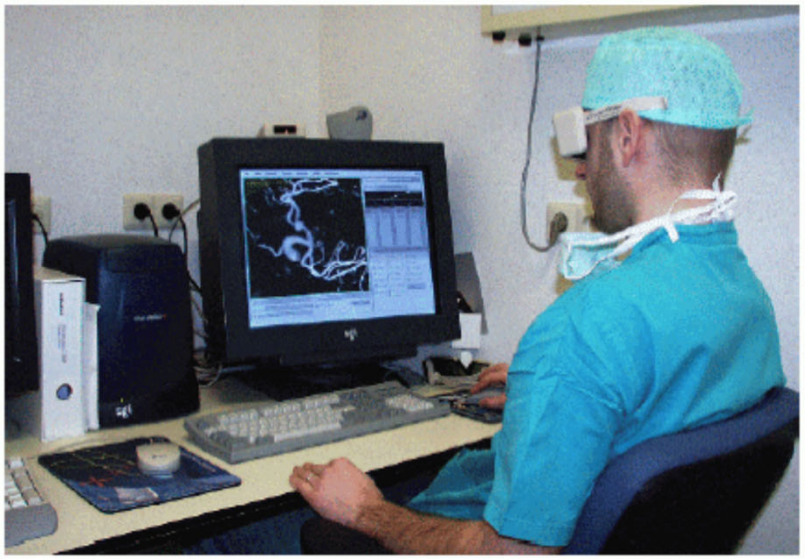
Roentgen's X-ray Setup (1895)

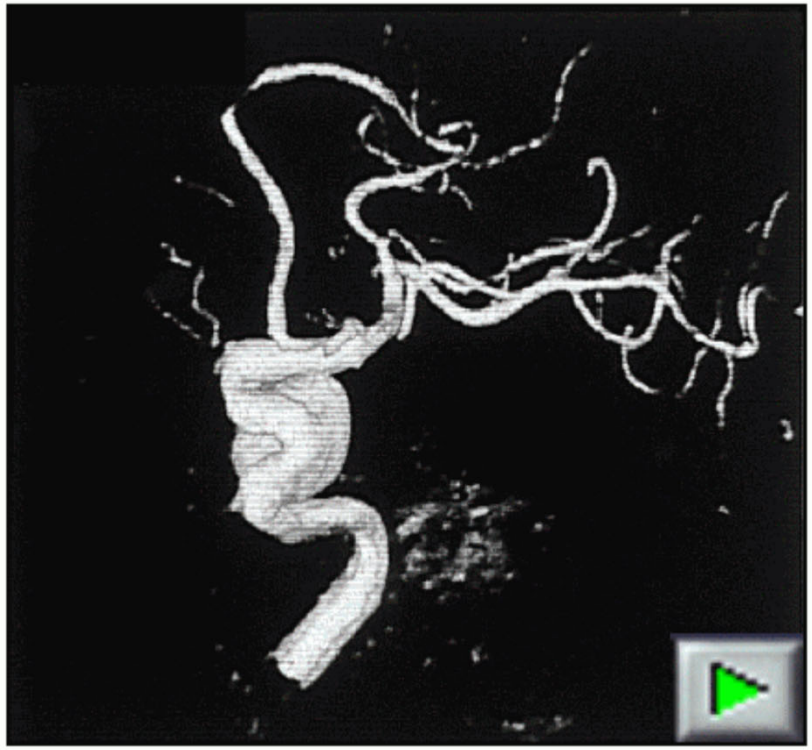


Roentgen's experimental apparatus (Crookes tube) that led to the discovery of the new radiation on 8 Nov. 1895 – he demonstrated that the radiation was not due to charged particles, but due to an as yet unknown source, hence "x" radiation or "x-rays"

Known as "the radiograph of Bera Roentgen's hand" taken 22 Dec. 1895

X-ray Imaging Examples

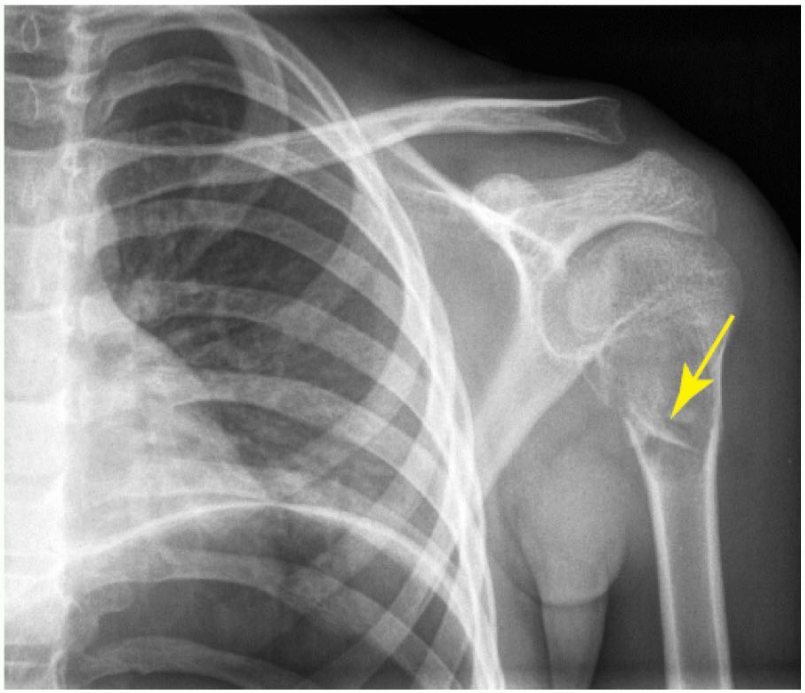




3D image of the blood vessels viewed by means of stereoscopic glasses.

X-ray Imaging Examples

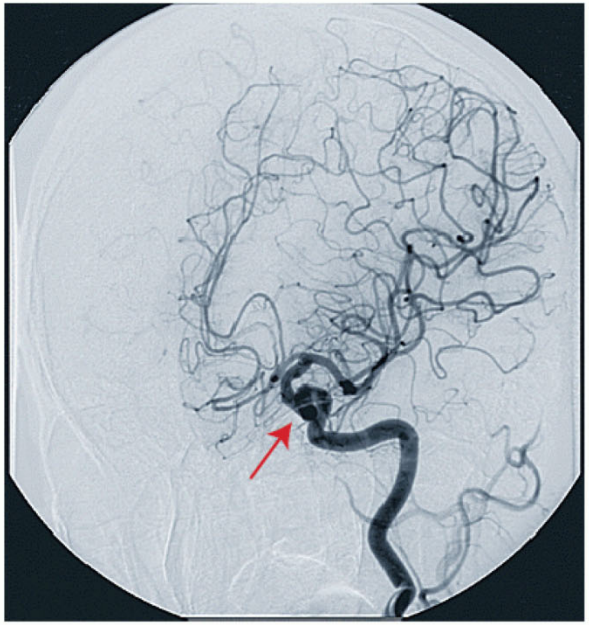
Double mandibular fracture with strong displacement to the left.



Solitary humeral bone cyst known as ”fallen leaf sign”

X-ray Imaging Examples

Cerebral angiogram obtained by injecting a iodine containing fluid into the arteries. The contrast dye subsequently fills the cerebral arteries, capillaries and veins.



Cerebral angiogram showing an aneurysm or saccular dilation of a cerebral artery.

K-edge Subtraction X-ray Imaging

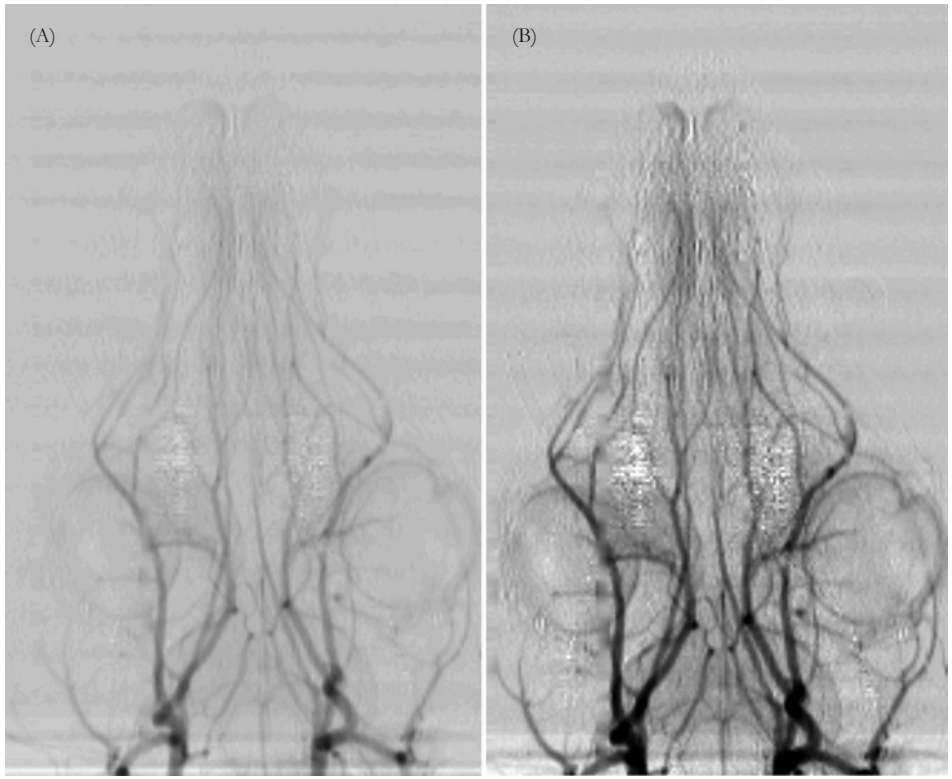


Figure 3. *Image of intracerebral arteries acquired in radiography mode with the germanium detector in antero-posterior projection, using synchrotron K-edge digital subtraction angiography: (A) early filling phase, (B) late filling phase.*

Synchrotron-based intravenous cerebral angiography in a small animal model, PMB, 2004

K-edge Subtraction X-ray Imaging

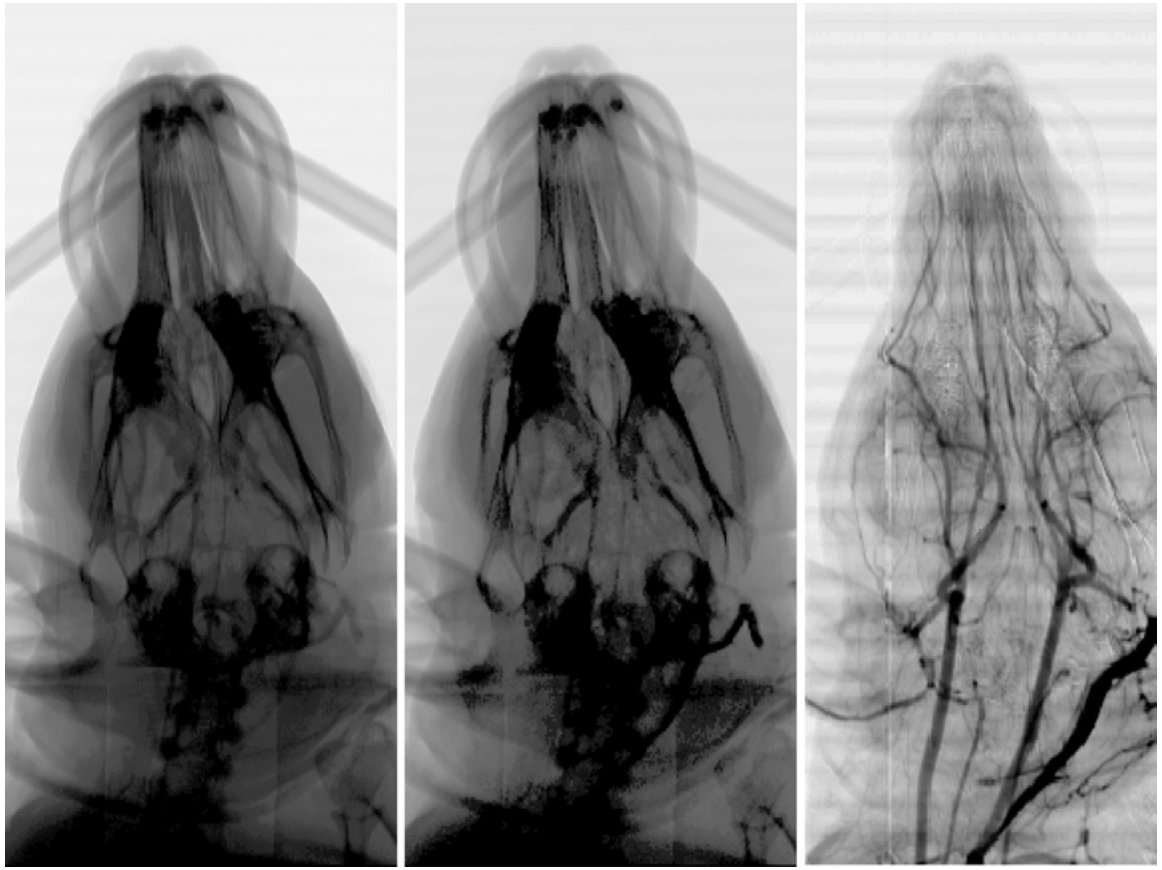


Figure 4. Images acquired below (A) and above (B) the K-edge of iodine, both without easily discernible contrast in the cerebral arteries. The subtracted image (C), however, shows very good contrast in the cerebral arteries.

Synchrotron-based intravenous cerebral angiography in a small animal model, PMB, 2004

Questions

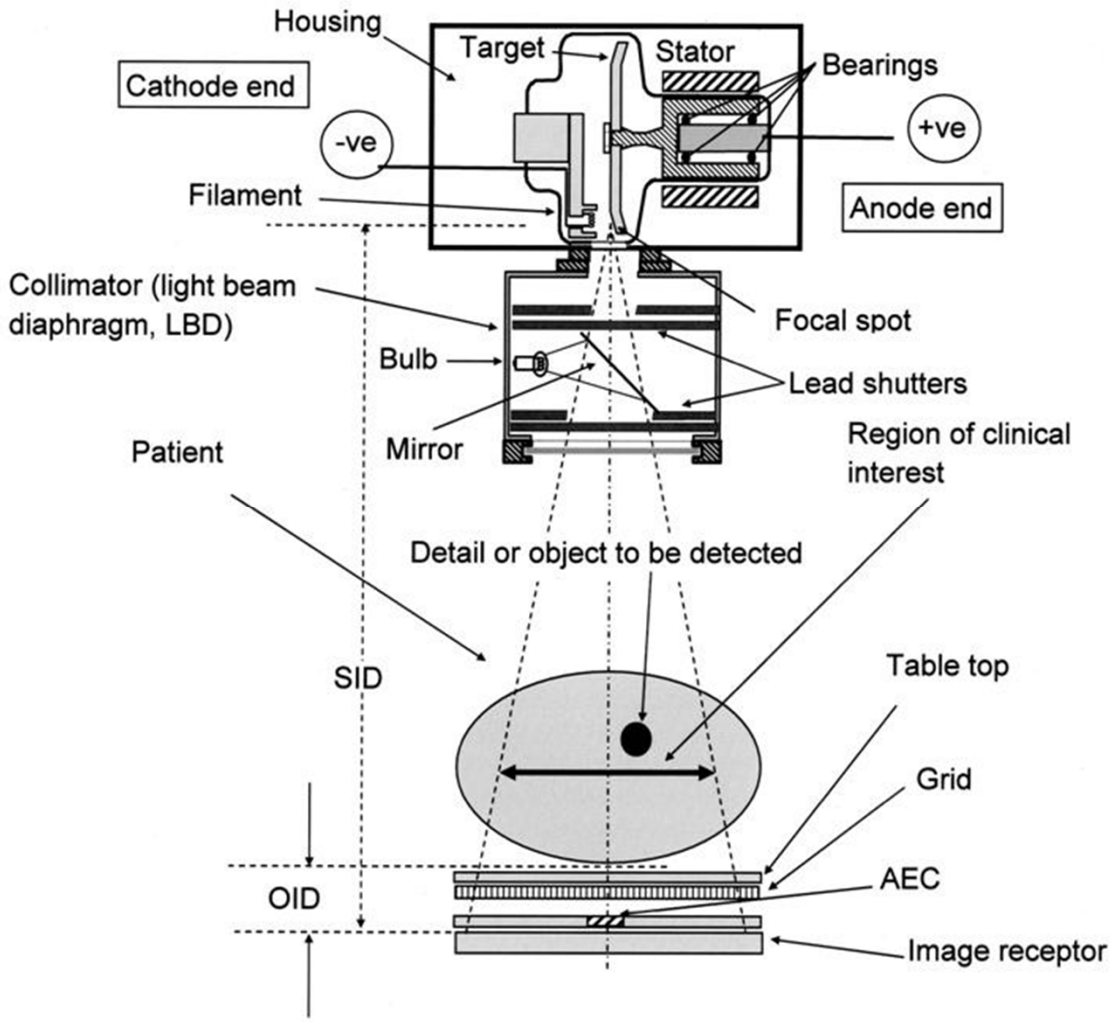
Questions concerned in this lecture:

- Why X-ray is the most used ionizing radiation for diagnostic imaging?
- What are we measuring with X-ray radiography?
- What are the physical factors that contribute to the contrast in an X-ray image?
- What is the fundamental limitation of X-ray for diagnostic purpose?

To answer these questions, we need to know the basics of X-ray physics.

- How X-rays are generated?
- What are the characteristics of these particles?
- How are X-rays attenuated in objects and in X-ray detector?

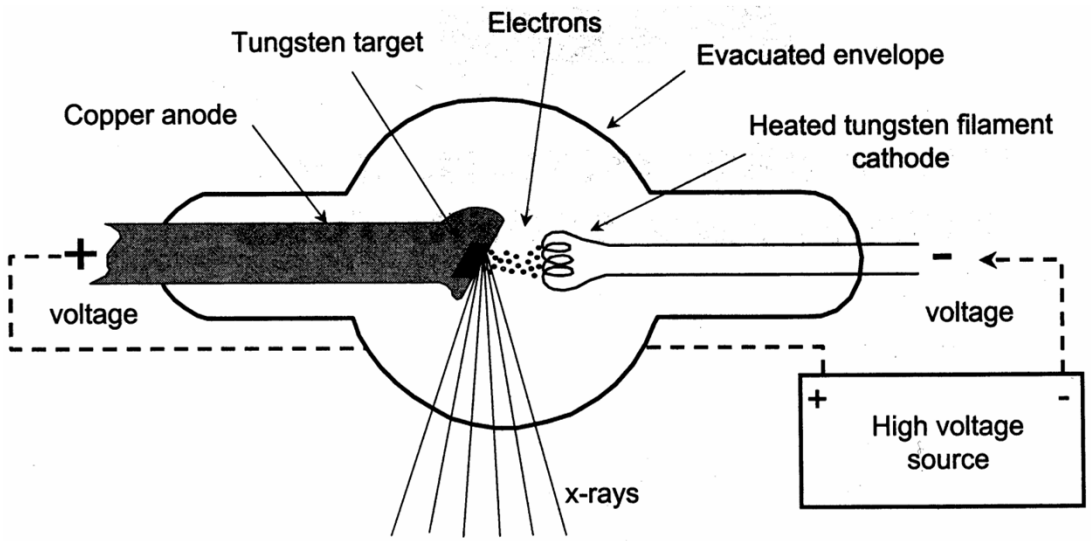
A Typical X-ray Radiography System



https://humanhealth.iaea.org/HHW/MedicalPhysics/TheMedicalPhysicist/Studentscorner/HandbookforTeachersandStudents/Chapter_06.pdf

X-ray Sources

X-ray Generation – Bremmstrahlung



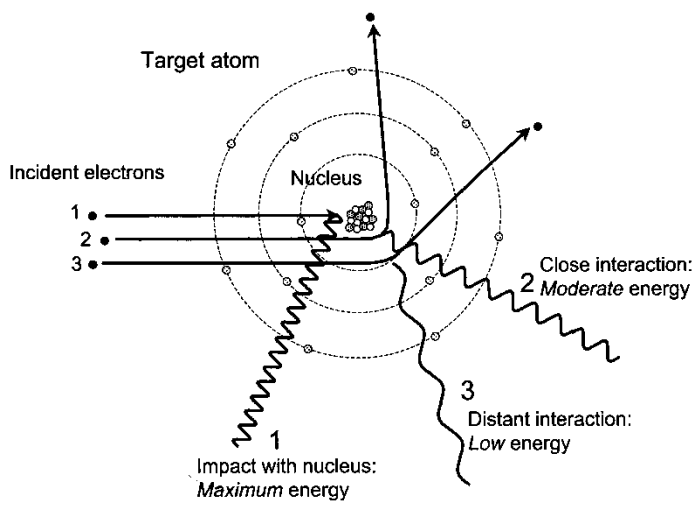
- e^- released from the cathode are accelerated towards the anode with a gain in KE as the e^- drops through the applied potential difference (kilovoltage potential - kVp)

Unfortunately,

- About 99% of the KE converted to heat via collision-like interactions
- About 1% of the KE converted into x-rays via strong Coulomb (electrostatic) interactions → Bremsstrahlung

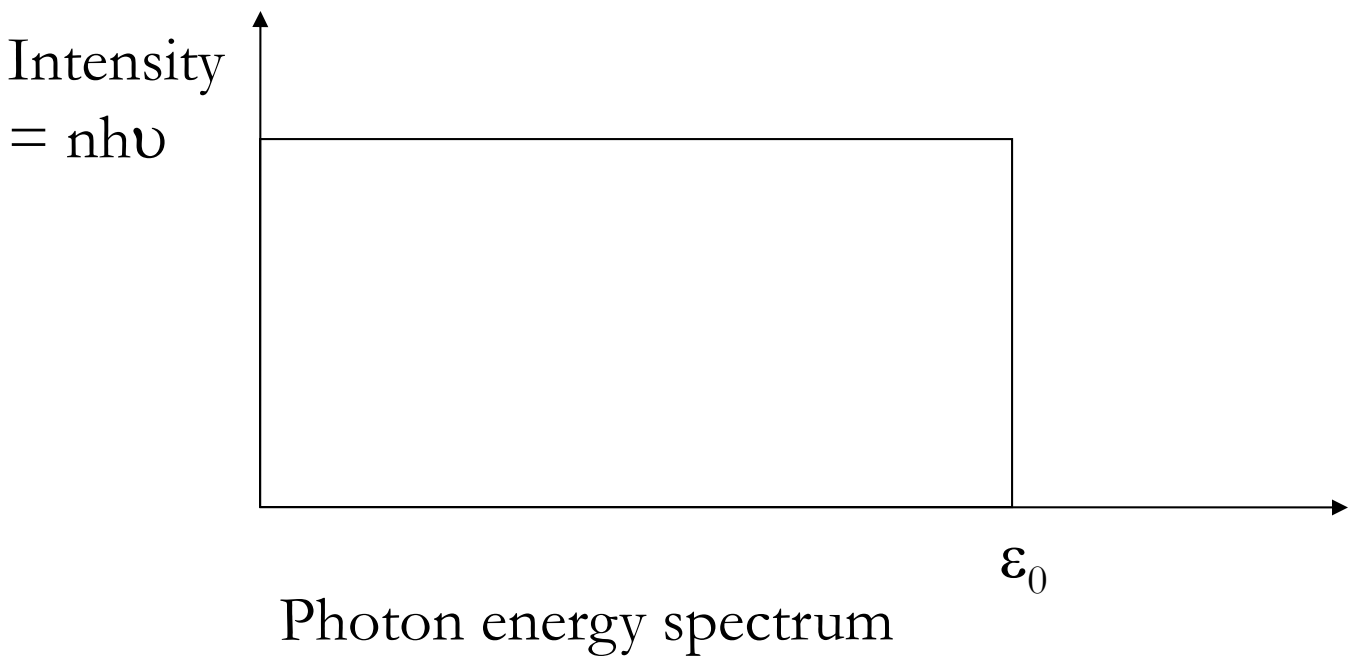
X-ray Generation – Bremmstrahlung

- Target nucleus positive charge ($Z \cdot p^+$) attracts incident e^-
- Deceleration of an incident e^- occurs in the proximity of the target atom nucleus
- E lost by e^- is gained by the EM photon (x-ray) generated
  - The impact parameter distance, the closest approach to the nucleus by the e^- determines the amount of E loss
  - The Coulomb force of attraction varies strongly with distance ($\propto 1/r^2$); ↓ distance → ↑ deceleration and E loss → ↑ photon E
  - Direct impact on the nucleus determines the maximum x-ray E (E_{max})



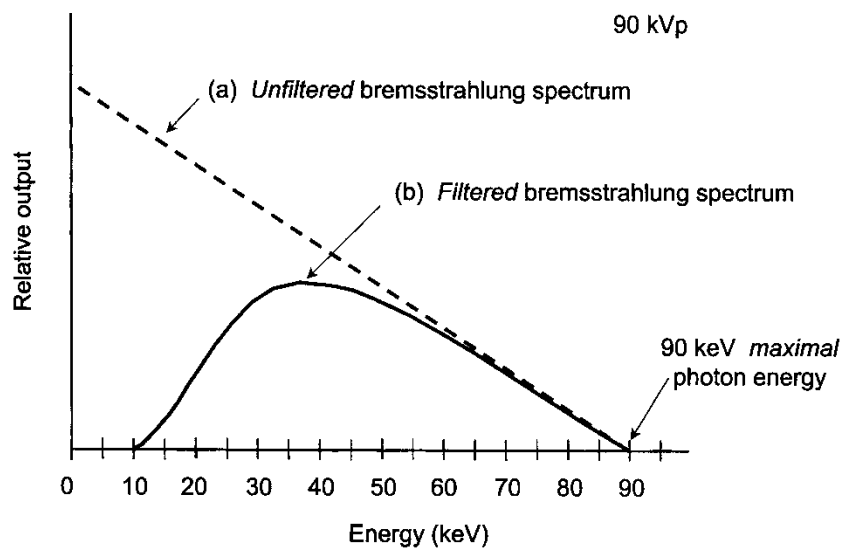
X-ray Generation – Bremmstrahlung

Interestingly, this process creates a relatively uniform spectrum.



X-ray Generation – Bremmstrahlung

- The unfiltered Bremsstrahlung spectrum (intensity) contains a large number of very low E photons and ↓ approx. linearly as photon E ↑.
- The peak voltage (kVp) applied across the electrodes of the x-ray tube determines the highest x-ray E (E_{max})
- The lowest E of the unfiltered x-ray spectrum is not easily determined, due to severe attenuation of these photons by the material and thickness of the x-ray tube envelope
- X-ray production efficiency is influenced by the target Z and acceleration potential (kVp)

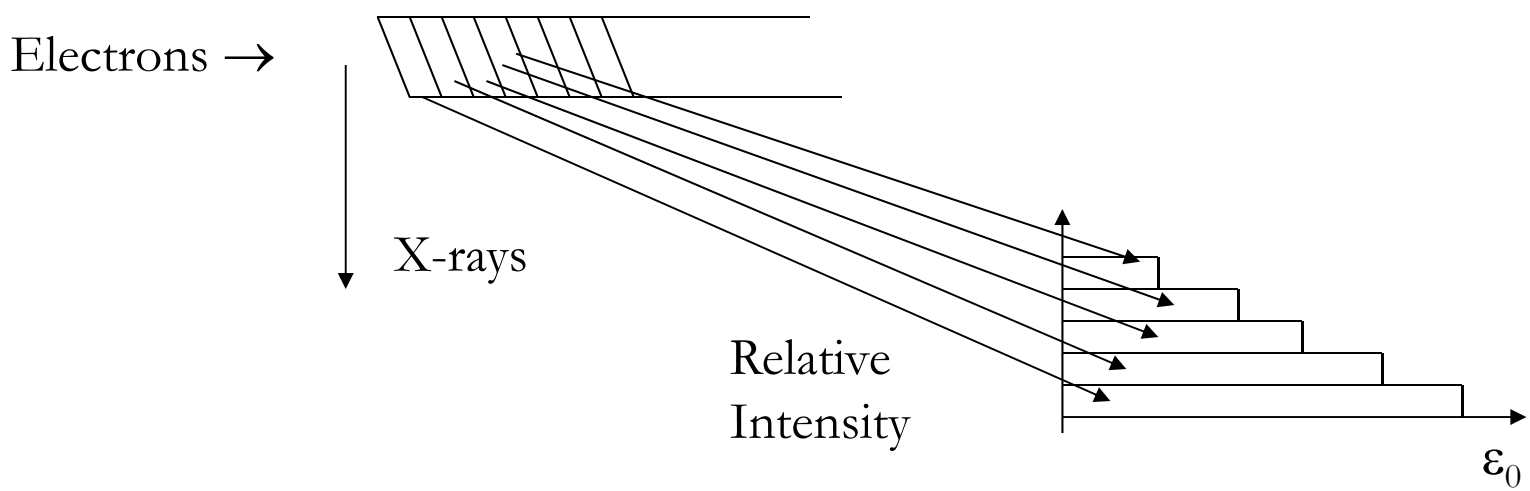


$$-\left(\frac{dE}{dx}\right)_r = \frac{NEZ(Z+1)e^4}{137 m_e^2 v^4}\left(4\ln\frac{2E}{m_e c^2} - \frac{4}{3}\right)$$

Specific energy loss by Bremmstrahlung

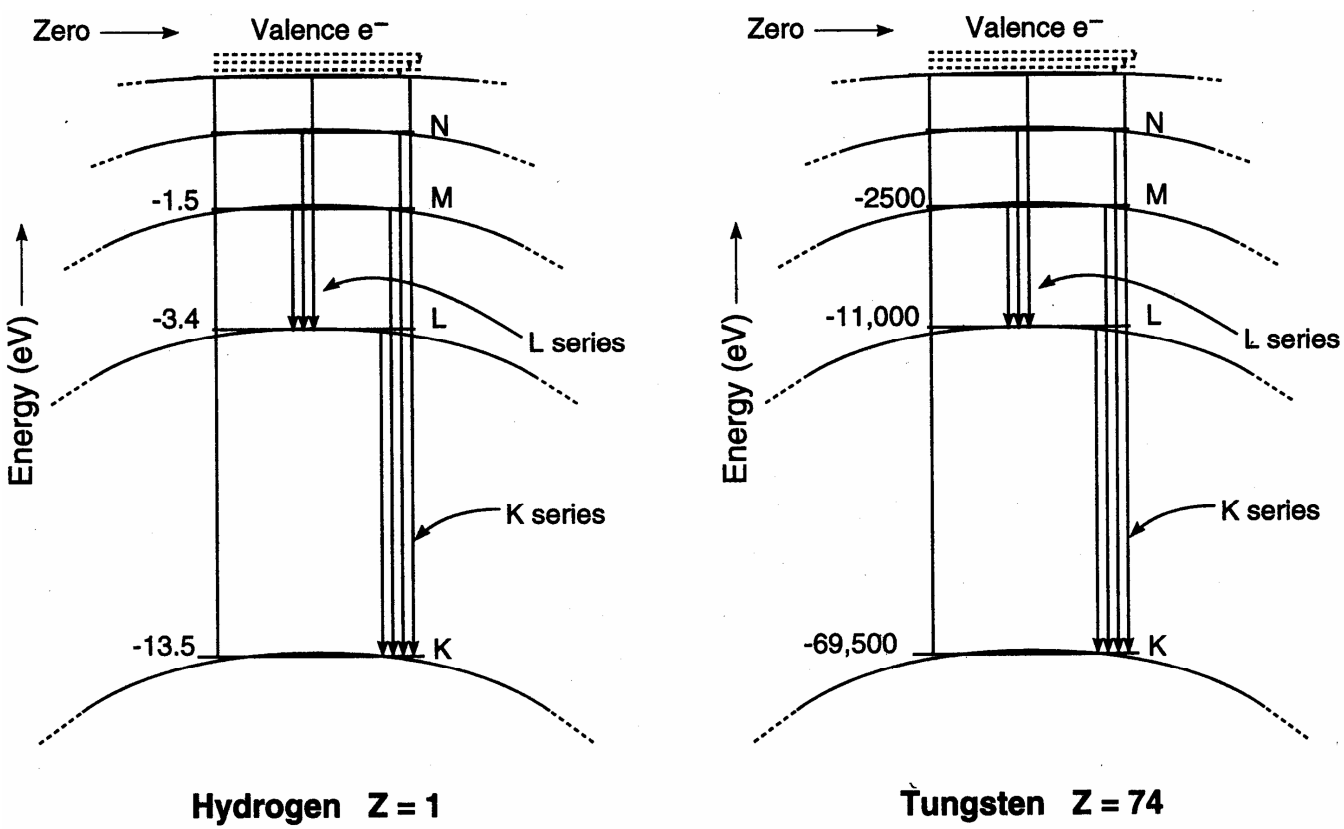
The Unfiltered Bremsstrahlung Spectrum
Thick Target X-ray Formation

We can model target as a series of thin targets. Electrons successively loses energy as they moves deeper into the target.



Each layer produces a flat energy spectrum with decreasing peak energy level.

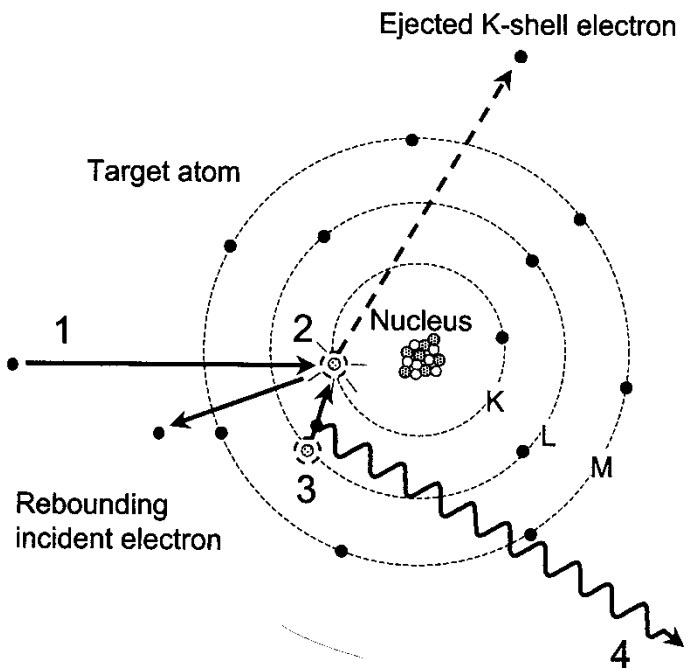
X-ray Generation – Characteristic X-rays



Electron binding energy

X-ray Generation – Characteristic X-rays

- e^- of the target atom have a binding energy (BE) that depends on atomic Z (rem: $BE_K \propto Z^2$) and the shell ($BE_K > BE_L > BE_M > \ldots$)
- When e^-(KE) incident on the target exceeds the target atom e^-(BE), it's energetically possible for a collisional interaction to eject the bound electron and ionize the atom.
- What would happen then?



X-ray Generation – Characteristic X-rays

TABLE 5-1. ELECTRON BINDING ENERGIES (keV) OF COMMON X-RAY TUBE TARGET MATERIALS

Electron Shell	Tungsten	Molybdenum	Rhodium
K	69.5	20.0	23.2
L	12.1/11.5/10.2	2.8/2.6/2.5	3.4/3.1/3.0
M	2.8–1.9	0.5–0.4	0.6–0.2

- The target materials used in x-ray tubes for diagnostic medical imaging include: W (Z=74), Mo (Z=42) and Rh (Z=45): BE $\propto Z^2$.
- As the E of the incident e^- increases above the threshold E for characteristic x-ray production, the % of char. x-rays ↑ (5% @ 80 kVp vs. 10% @ 100 kVp for W).
- A variety of E transitions occur from adjacent (α) and non-adjacent (β) e^- shells → discrete E lines superimposed on the continuous bremsstrahlung spectrum.

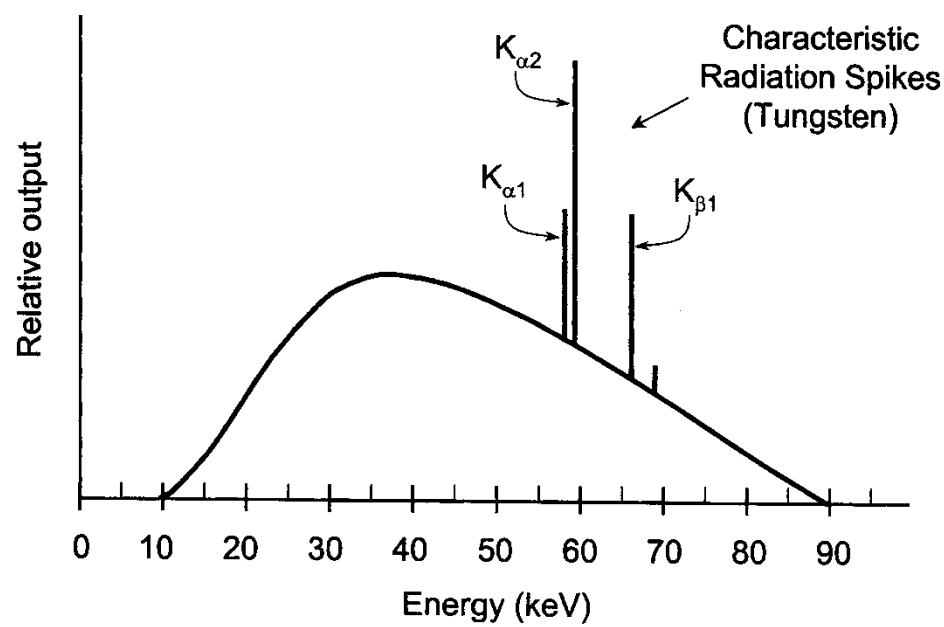
X-ray Generation – Characteristic X-rays

- Within each shell (other than K) there are discrete E orbitals (ℓ = 0, 1, ... , n-1) → characteristic x-ray fine E splitting
- Characteristic x-rays other than those generated through K-shell transitions are unimportant

TABLE 5-2. K-SHELL CHARACTERISTIC X-RAY ENERGIES (keV) OF COMMON X-RAY TUBE TARGET MATERIALS[a]

Shell Transition	Tungsten	Molybdenum	Rhodium
$K_{\alpha 1}$	59.32	17.48	20.22
$K_{\alpha 2}$	57.98	17.37	20.07
$K_{\beta 1}$	67.24	19.61	22.72

[a]*Note:* Only prominent transitions are listed.



X-ray Generation – Characteristic X-rays

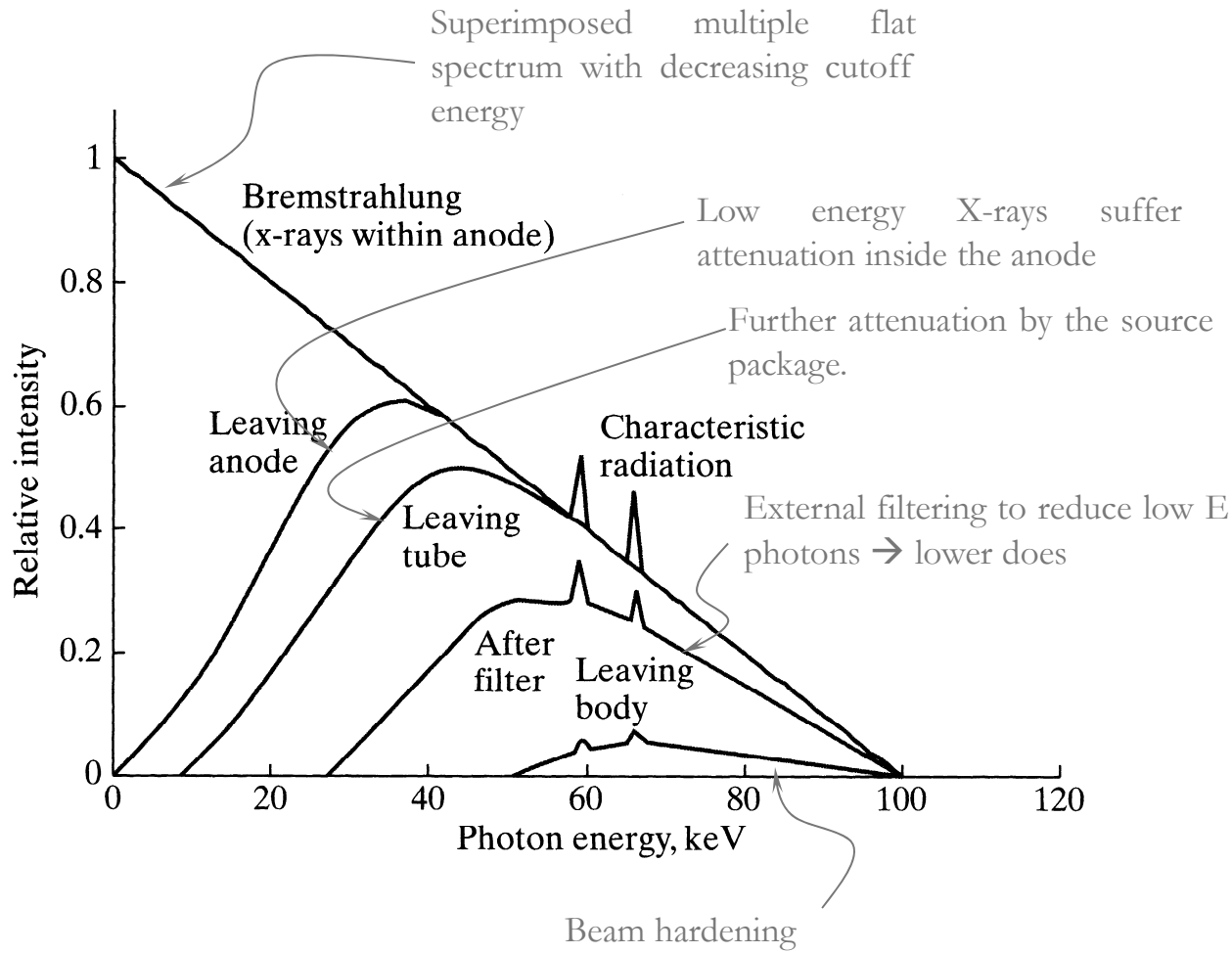


Figure 5.5
Relative intensity of x-ray photons. (Adapted from Webster, 1998. This material is used by permission of John Wiley & Sons, Inc.)

Practical Considerations on X-ray Source

X-ray Generation – X-ray Tube

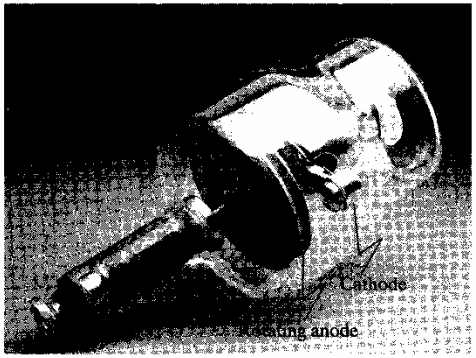


Figure 5.3
An x-ray tube.

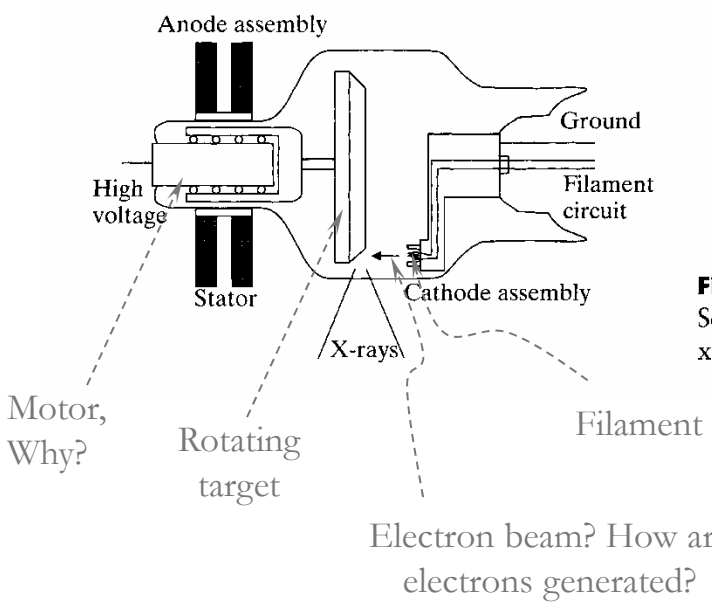
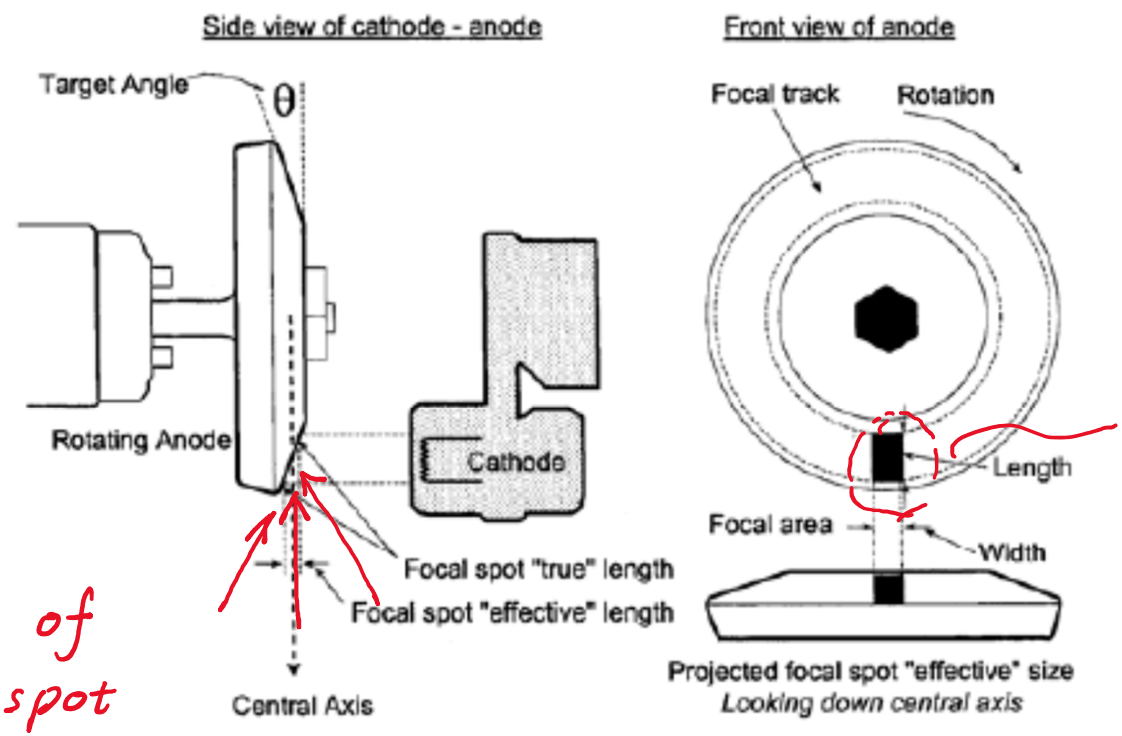


Figure 5.4
Schematic diagram of an x-ray tube.

X-ray Generation – Anode Design



not circular

The size of the focus spot varies with angle.

FIGURE 5-13. The anode (target) angle, θ, is defined as the angle of the target surface in relation to the central ray. The focal spot length, as projected down the central axis, is foreshortened, according to the line focus principle (lower right).

References: Bushberg text
The Physics of Medical Imaging, Webb, IOP Publ.

X-ray Generation – Filament

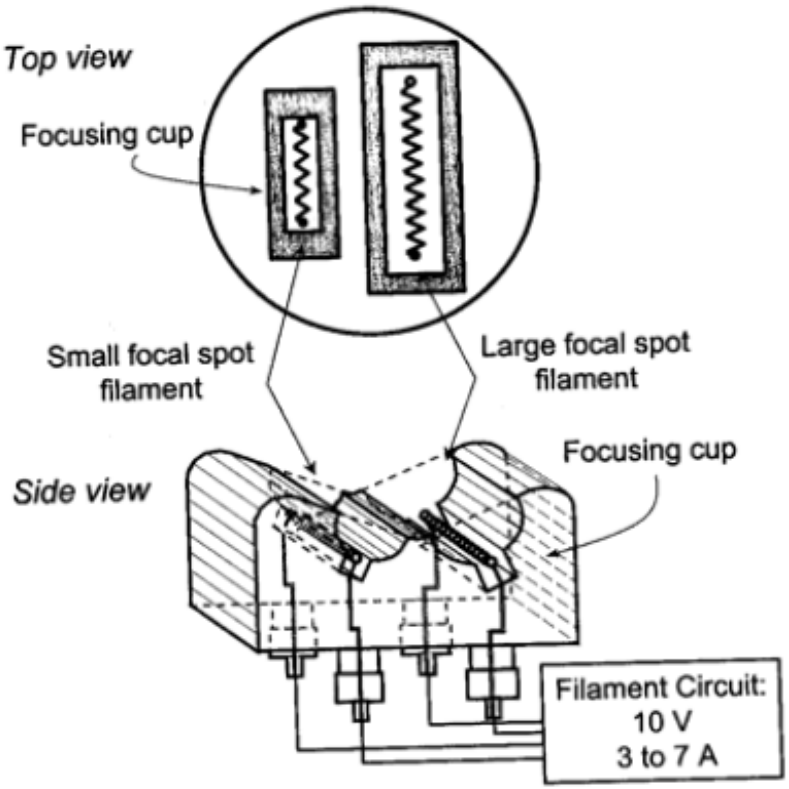


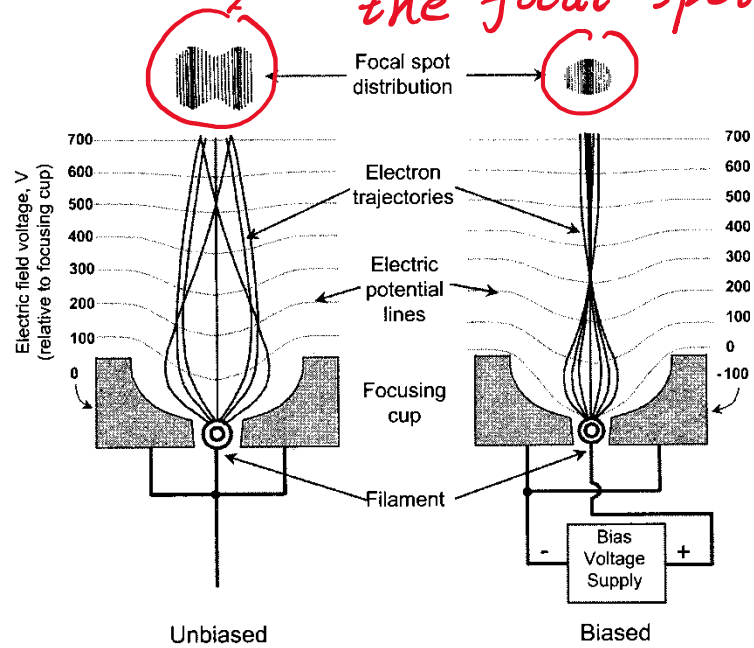
FIGURE 5-7. The x-ray tube cathode structure consists of the filament and the focusing (or cathode) cup. Current from the filament circuit heats the filament, which releases electrons by thermionic emission.

References: Bushberg text
The Physics of Medical Imaging, Webb, IOP Publ.

X-ray Generation – Cathode

- Shapes e^- distribution when at same V as filament (unbiased)
- Isolation from filament and application of a negative bias V constrains e- distribution further (biased)
- Focusing cup slot width determines the focal spot width
- Filament length determines focal spot length
- Small and large focal spot filaments (usu. 0.6 and 1.2 mm)

flux coming out of the focal spot is NOT uniform.

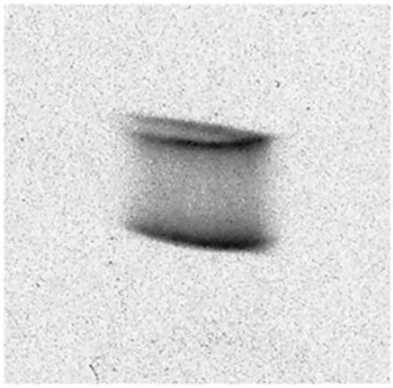


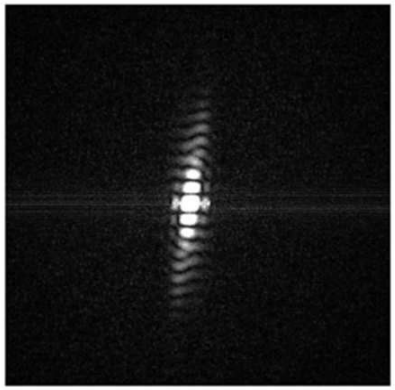
What limits the focal spot size? The power output of the tube!

X-ray Generation – Cathode

Geometrical Unsharpness

The **Focal Spot MTF** may be measured using a pinhole to determine the PSF or a slit to determine the LSF, and calculating the normalised modulus of the Fourier Transform of the spread function





(Left) A pinhole image a 2.0 mm focal spot showing a typical distribution of the X ray intensity

(Right) the corresponding 2D MTF

Power Rating Versus Focal Spot Size

TABLE 5-6. X-RAY TUBE FOCAL SPOT SIZE AND TYPICAL POWER RATING

Nominal X-ray Tube Focal Spot Size (mm)	Typical Power Rating (kW)
1.2–1.5	80–125
0.8–1.0	50–80
0.5–0.8	40–60
0.3–0.5	10–30
0.1–0.3	1–10
<0.1 (micro-focus tube)	<1

clinical sources

References: Bushberg text
The Physics of Medical Imaging, Webb, IOP Publ.

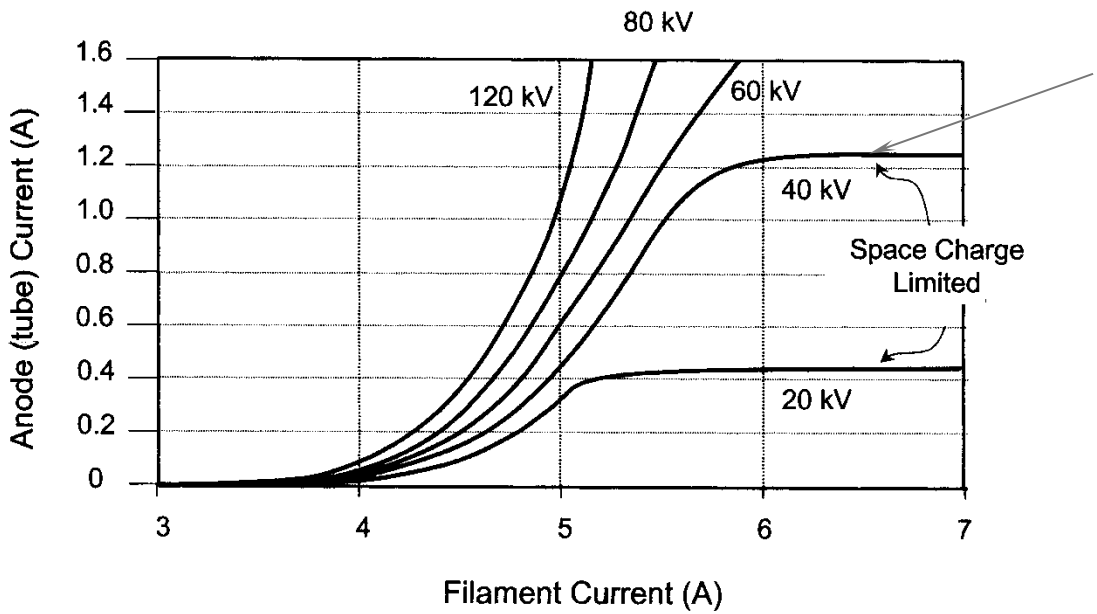
Factors Affecting X-ray Generation

1) Target material - affects quality and quantity of radiation emitted
2) Tube voltage – kVp - determines maximum energy of photons emitted in the bremsstrahlung spectrum, thus affecting quality of the beam and the overall exposure.

Rule of thumb 1: Exposure α $(kVp)^5$
Rule of thumb 2: Exposure α mAs

3) Tube current (mA)
4) Exposure time (s)– duration of x-ray production. Often the current and exposure time are expressed together as a product, in mAs.
5) Beam filtration – modifies the x-ray energy spectrum and the overall number of photons in the beam.
6) Generator waveform – affects the spectrum and quantity of photons emitted. Single phase system provides lower average energy and number of photons than does a three phase system.
7) Focal spot size – affects the number of photons being produced.

X-ray Generation – Tube Output

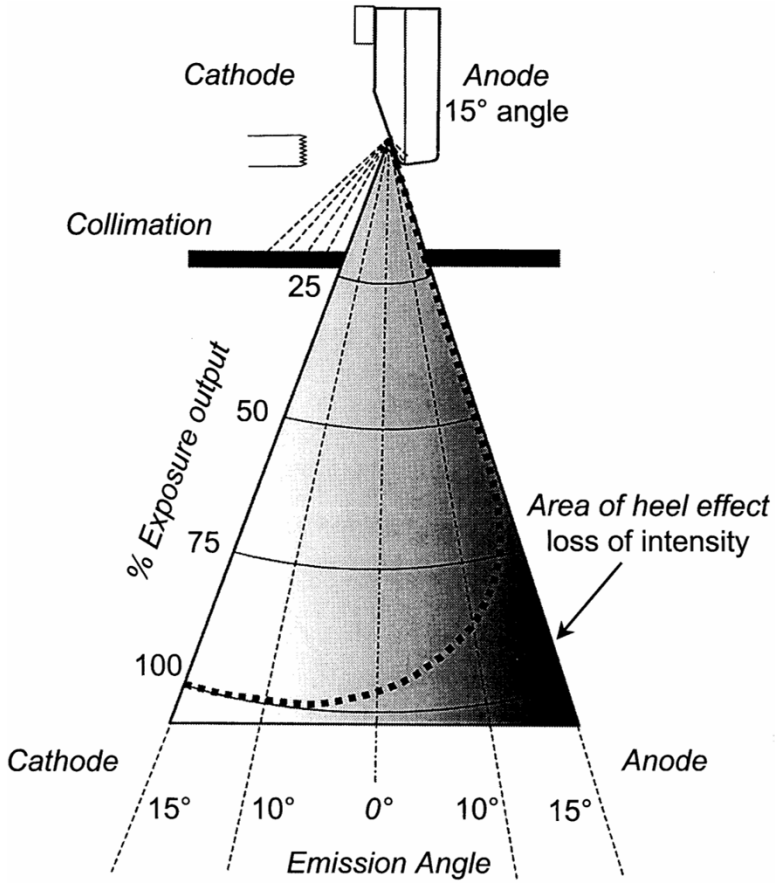


Increasing filament current does not increase tube current.

- Filament current (A) → filament temperature (T) → thermionic emission rate
- When kVp is low, an e^- cloud (space charge cloud) forms around filament
- Space charge cloud shields the electric field for tube voltages of ≤ 40 kVp → only some e^- are accelerated towards the anode: space charge limited
- ≥ 40 kVp the space charge cloud effect overcome by kVp applied and tube current (mA) limited only by the emission of e- from the filament: emission-limited operation
- Tube current about 5-10 times less than the filament current in the emission-limited range

X-ray Generation – Heel Effect

- Reduction of x-ray beam intensity towards the anode side of the x-ray field although x-rays generated isotropically.
- Self-filtration by the anode and the anode bevel causes greater intensity on the cathode side of the x-ray field
- Can use to advantage, e.g., chest exposure.
- Orient chest to anode side and abdomen to cathode side
- Less pronounced as SID ↑



X-ray Generation – Filtration

Filtration for geometrical compensation

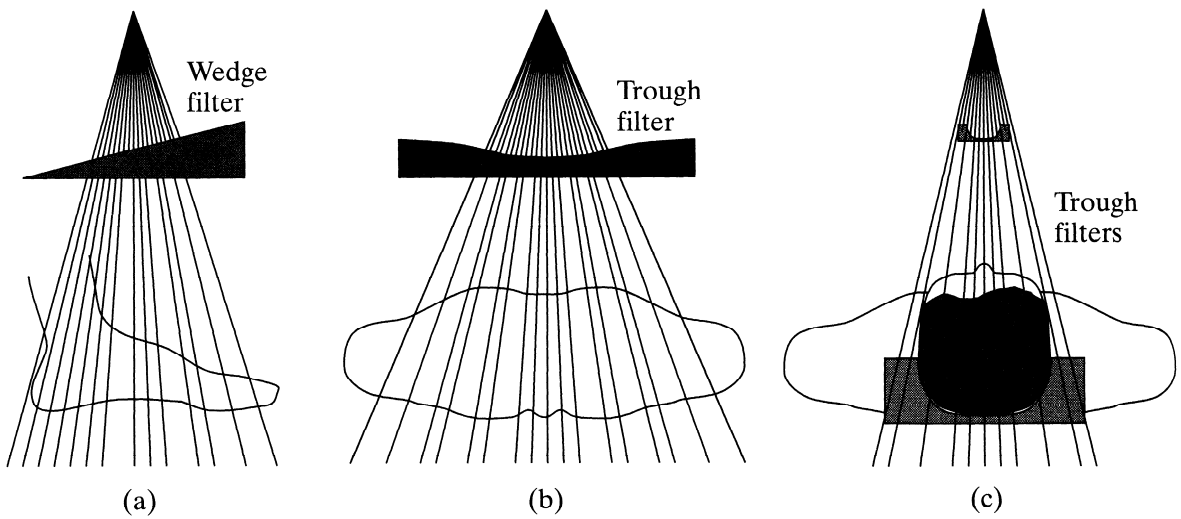


Figure 5.7
Various compensation filters. (Adapted from Carlton and Adler, 2001, and Wolbarst, 1993.)

X-ray Generation – Collimation

- Collimators adjust size and shape of x-ray beam
- Parallel-opposed lead shutters
- Light field mimics x-ray field
- Reduces dose to patient; ALARA: as low as reasonably achievable
- Reduced scatter radiation to image receptor: better image contrast