MAHIDOL UNIVERSITY
Wisdom of the Land

Radiology Instrumentation

ผศ.ดร.นภาพงษ์ พงษ์นภางค์

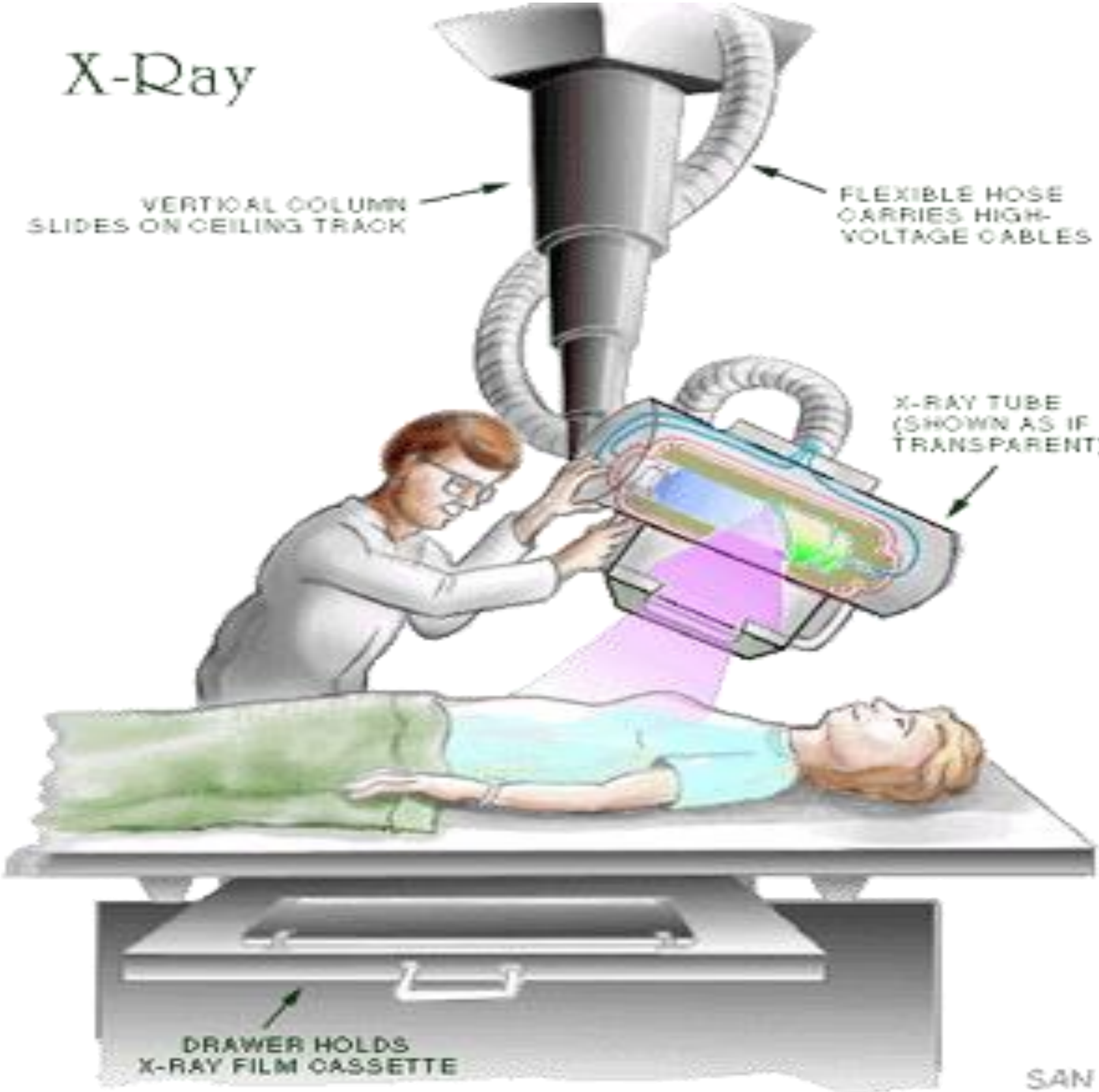
ภาควิชารังสีเทคนิค คณะเทคนิคการแพทย์

มหาวิทยาลัยมหิดล

Radiological Imaging Modalities

- X-Rays production
- Radiography
  - Plane film/screen
  - Digital
- Fluoroscopy
  - Conventional II
  - Digital
- Mammography
- Computed Tomography
- Dental

Plain Radiography

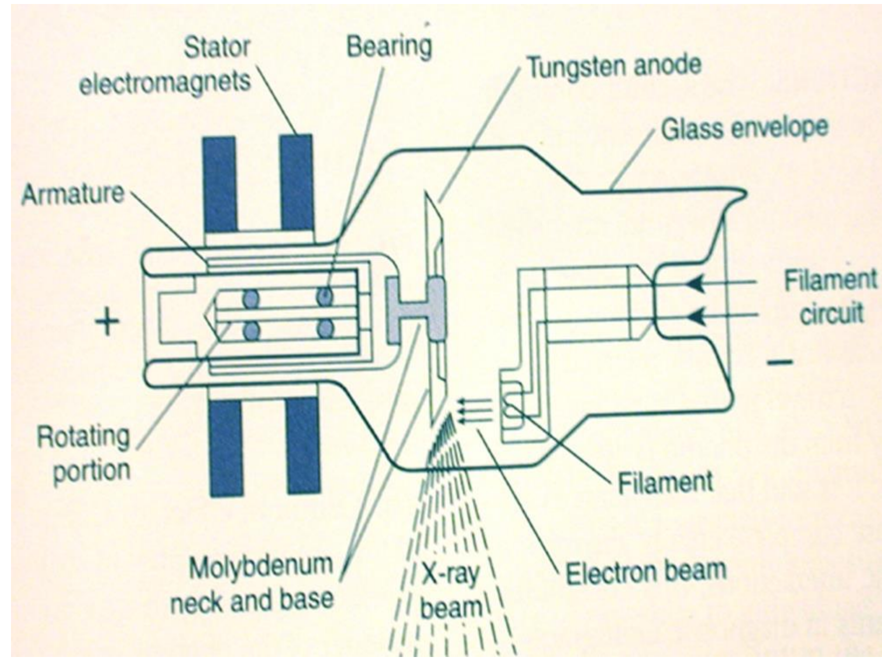

X-Ray
VERTICAL COLUMN SLIDES ON CEILING TRACK
FLEXIBLE HOSE CARRIES HIGH-VOLTAGE CABLES
X-RAY TUBE (SHOWN AS IF TRANSPARENT)
DRAWER HOLDS X-RAY FILM CASSETTE
SAN

X-RAY TUBE

- MADE OF THIN PYREX GLASS OR METAL ENCLOSURE TO WITHSTAND HIGH HEAT LOAD AND MINIMIZE X-RAY ABSORPTON

- IS GAS EVAUCUATED
- so electrons won’t collide with the air molecules in the tube

- The X-Ray tube is the single most important component of the radiographic system. It is the part that produces the X-rays

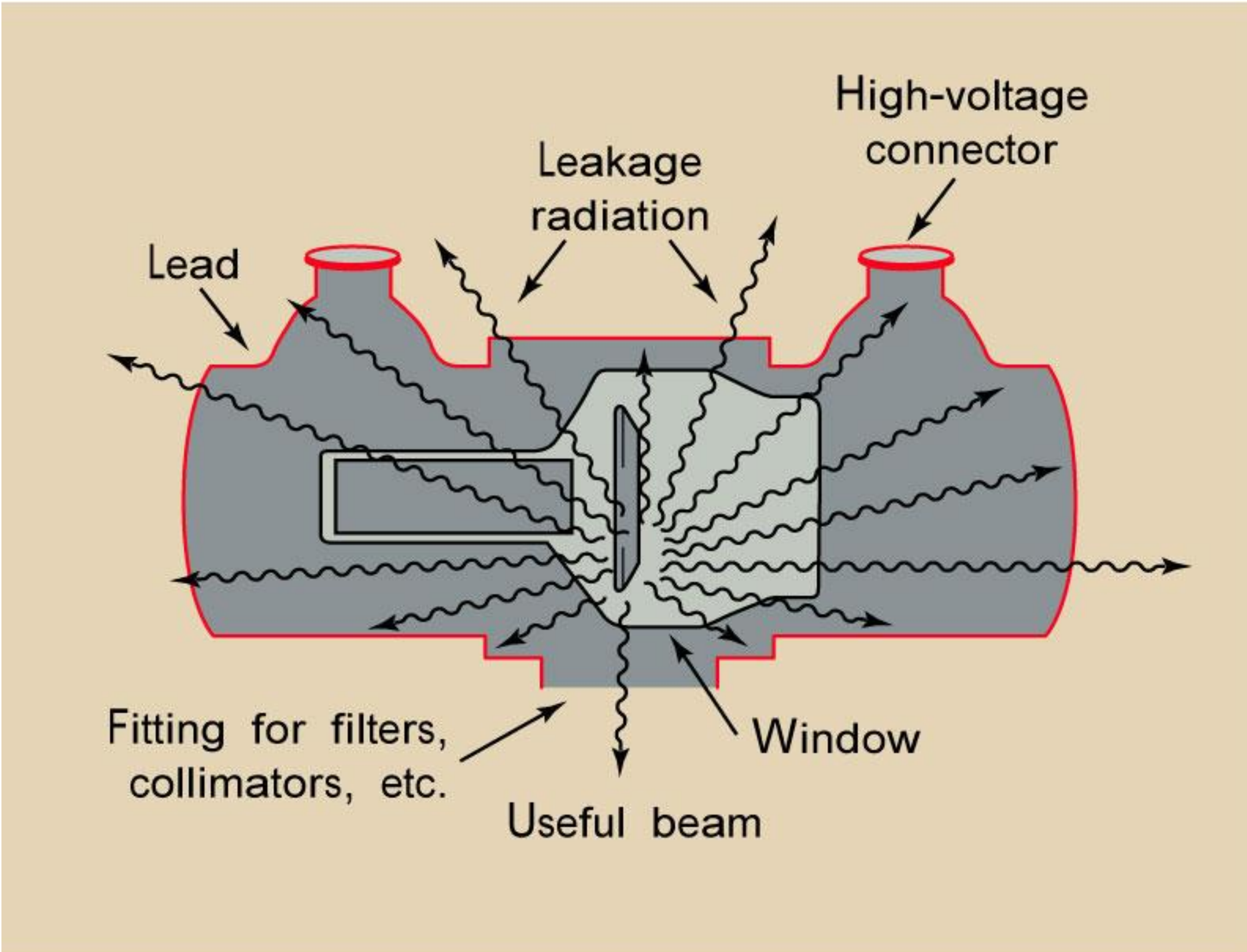


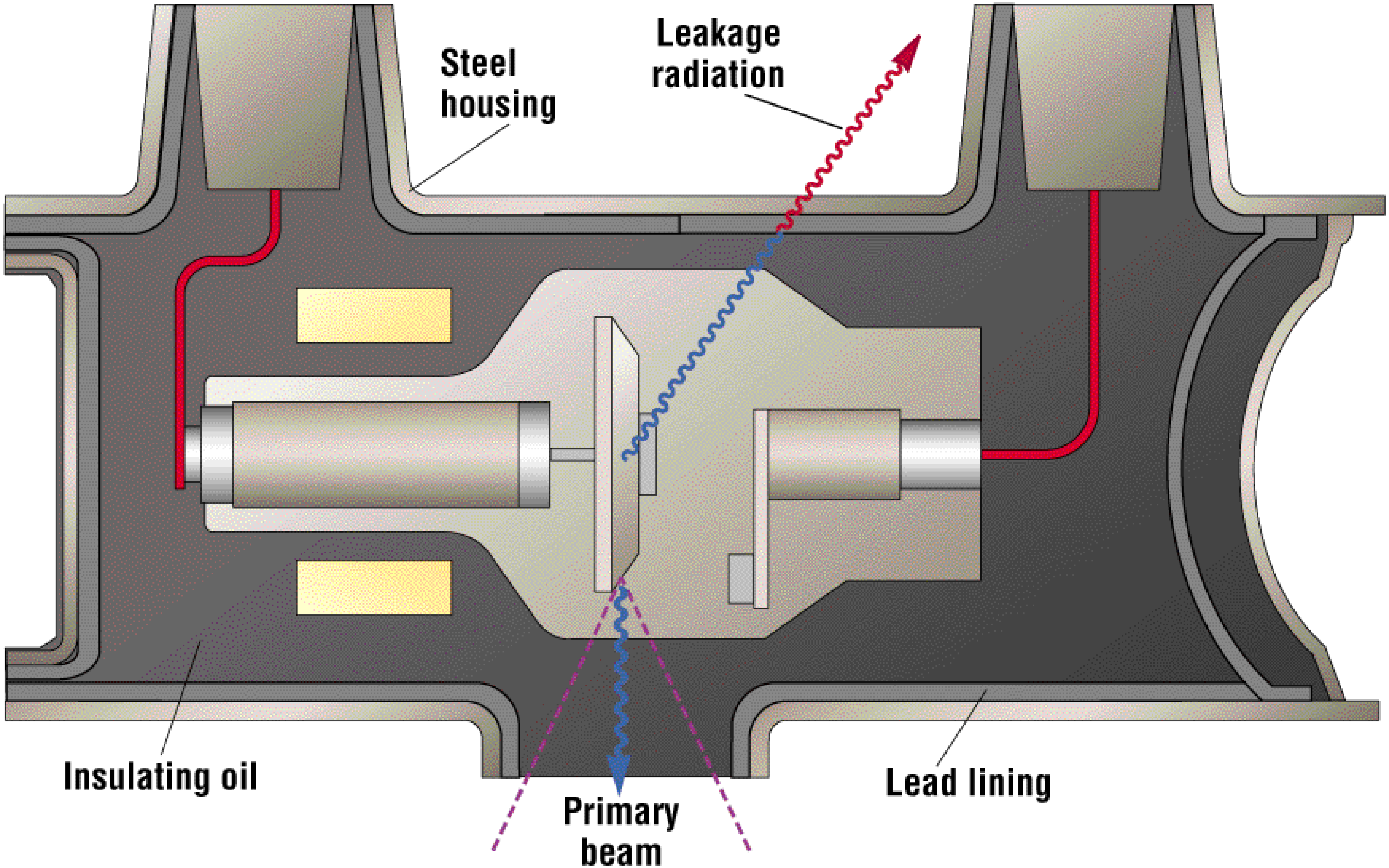
Protective housing

Made of lead & steel

- When x-rays are produced, they are emitted isotropically
  - Equal intensity in all directions

- We only use x-rays emitted through the window or port
  - Called the useful or primary beam


High-voltage connector
Leakage radiation
Lead
Fitting for filters, collimators, etc.
Window
Useful beam


Steel housing
Leakage radiation
Insulating oil
Primary beam
Lead lining

Protective housing

- X-rays that escape through the protective housing are leakage radiation
- Provides mechanical support for the tube and protects from rough handling

Protective housing

- Some tubes contain oil that serves as an insulator against electric shock and as a thermal cushion
  - Dissipate heat

- Some protective housing has cooling fan to air-cool the tube and oil

TUBE HOUSING
MADE OF LEAD & STEEL

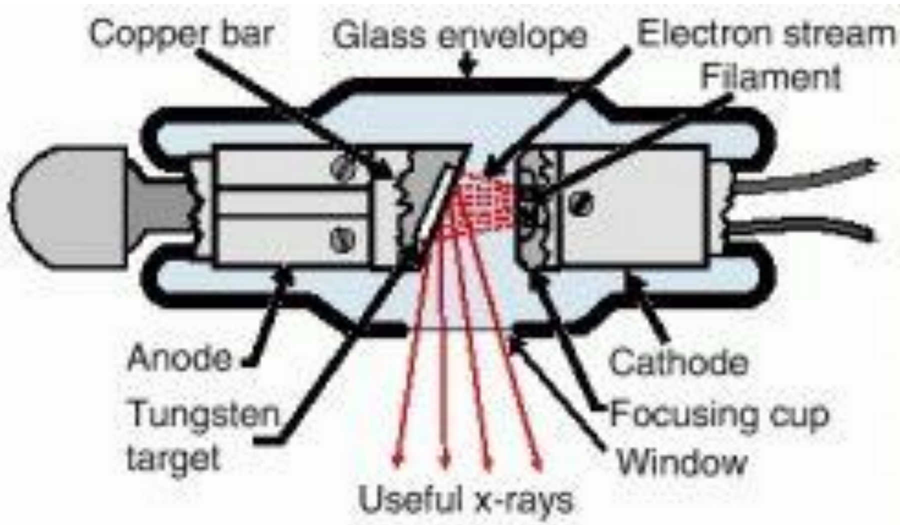
Internal components
Cathode

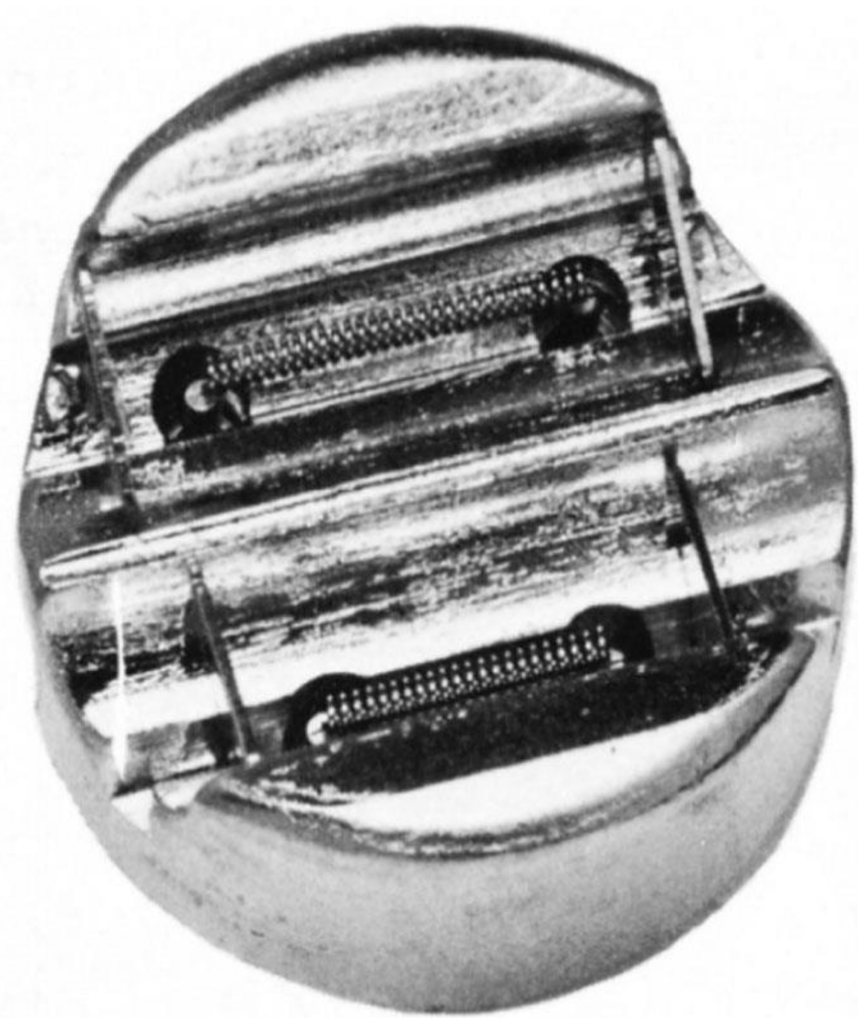
- The negative side of the tube and has two primary parts
  - A filament and focusing cup

- Filament = a coil of wire about 2mm in diameter and 1 or 2 cm long.

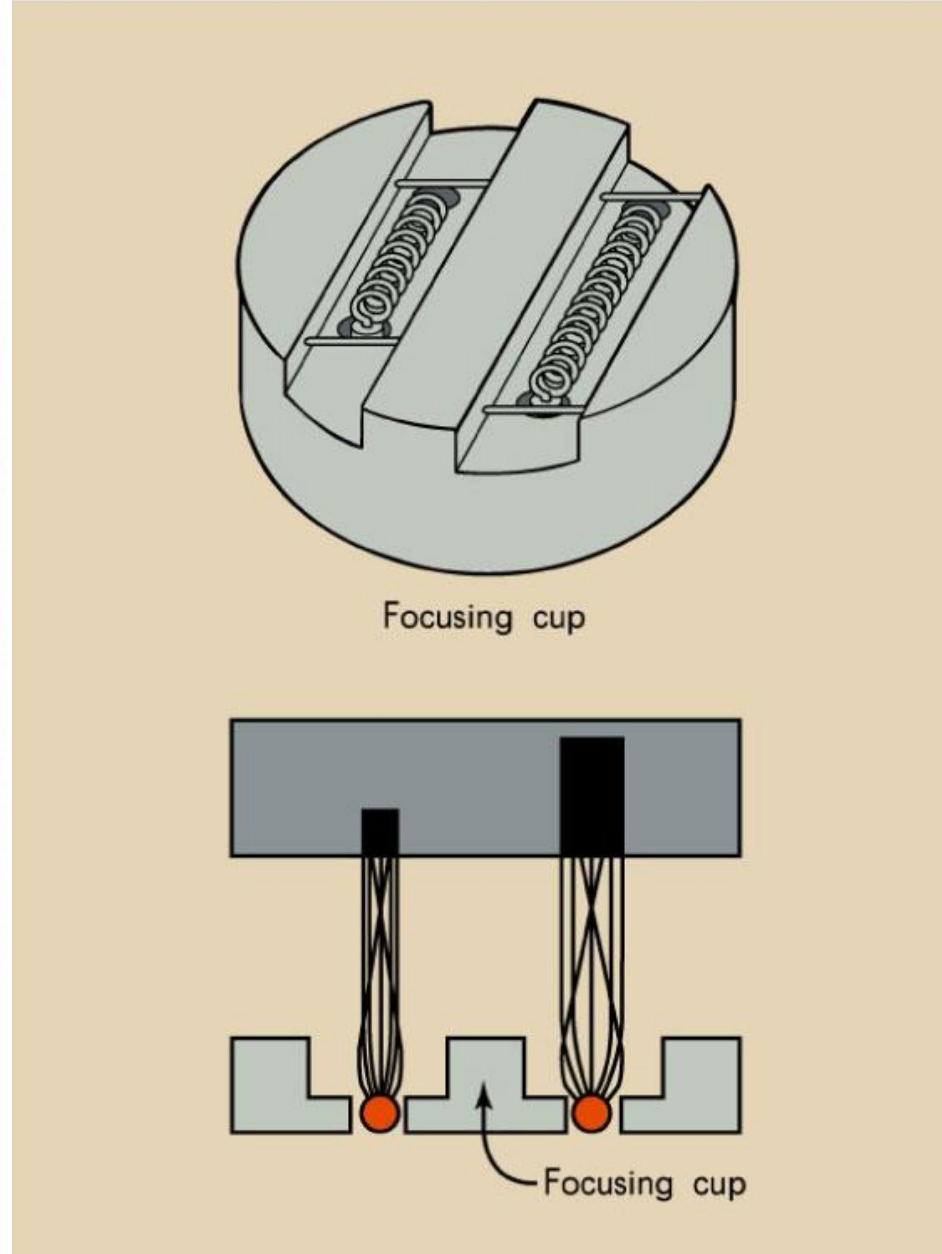
Cathode

- Filament
  - Dual-filament
- Focusing cup
  - Negatively charged



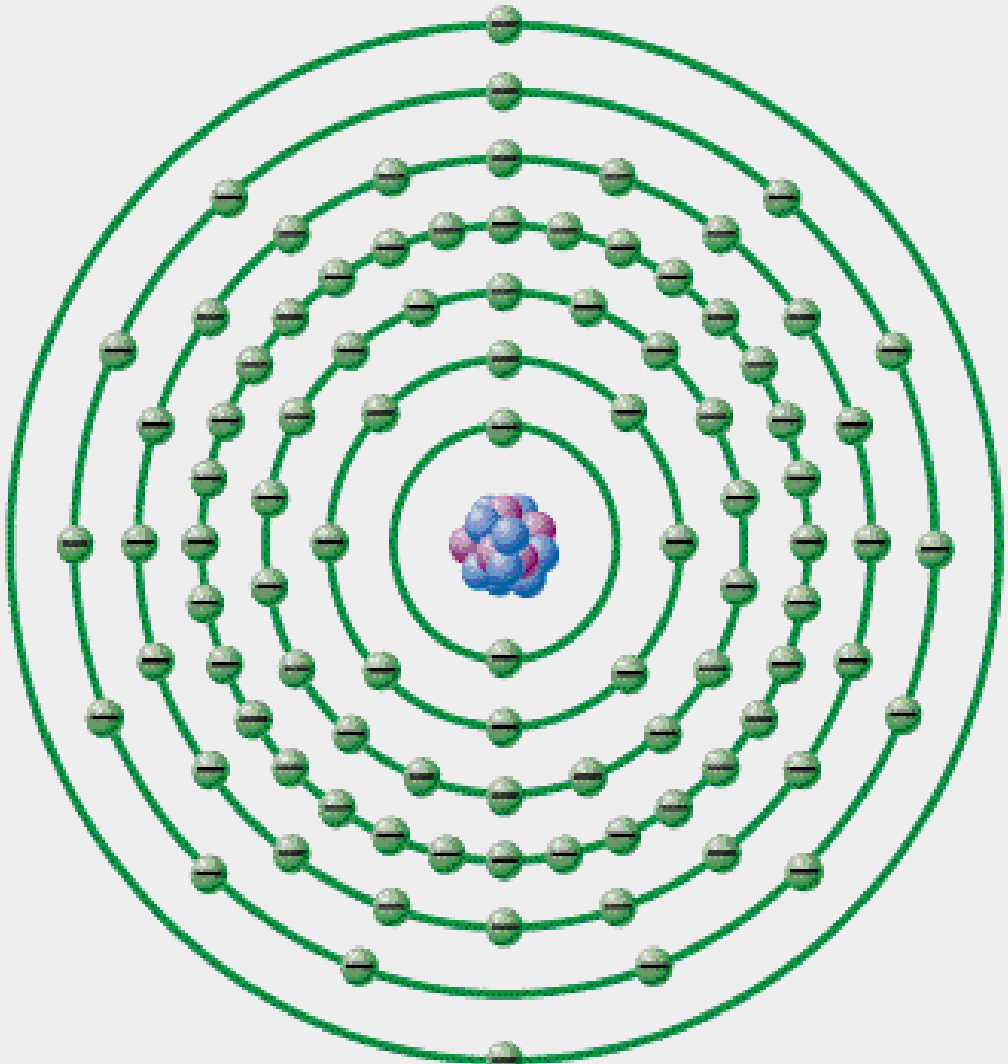


(A, Courtesy The Machlett Laboratories, Inc.)



Tungsten

- Filaments are usually made of tungsten
- Tungsten provides higher thermionic emission than other metals
- Tungsten has a very high melting point

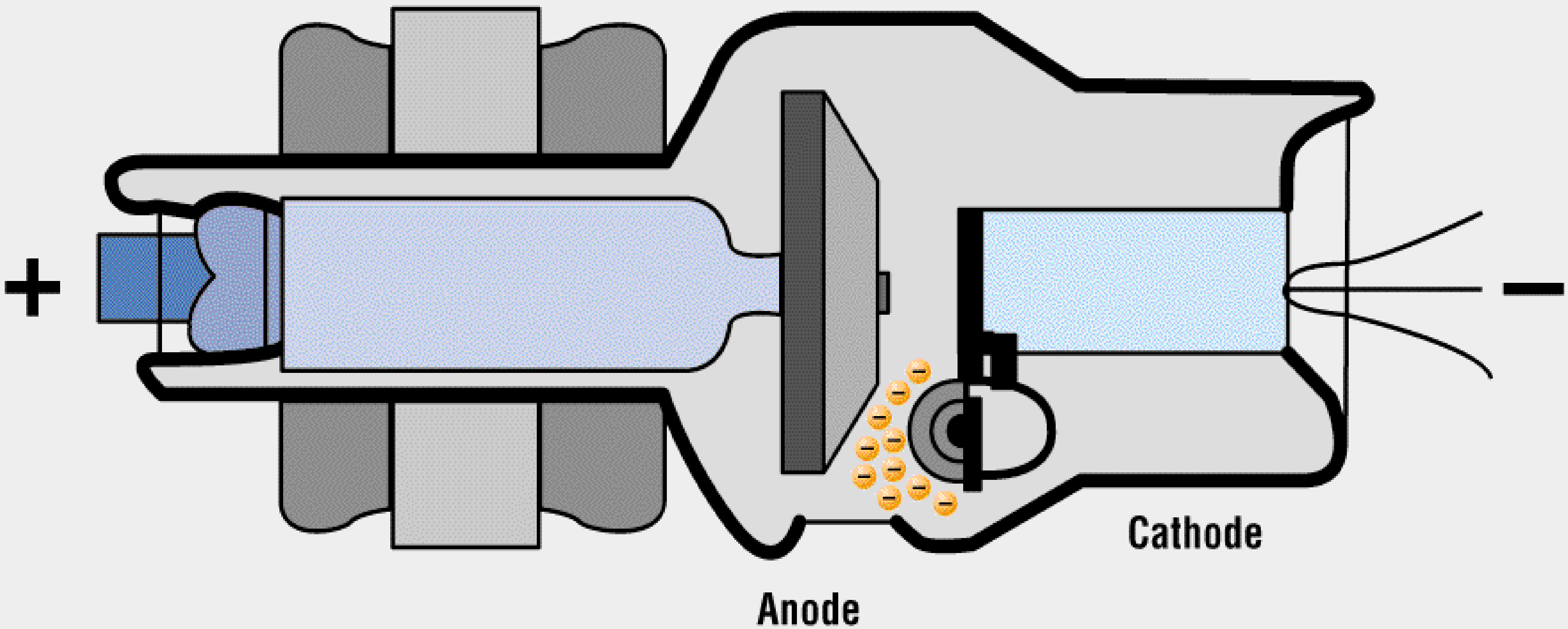


Tungsten Atom

Filament

- When current (mA) is applied to the coil of wire electron are ejected

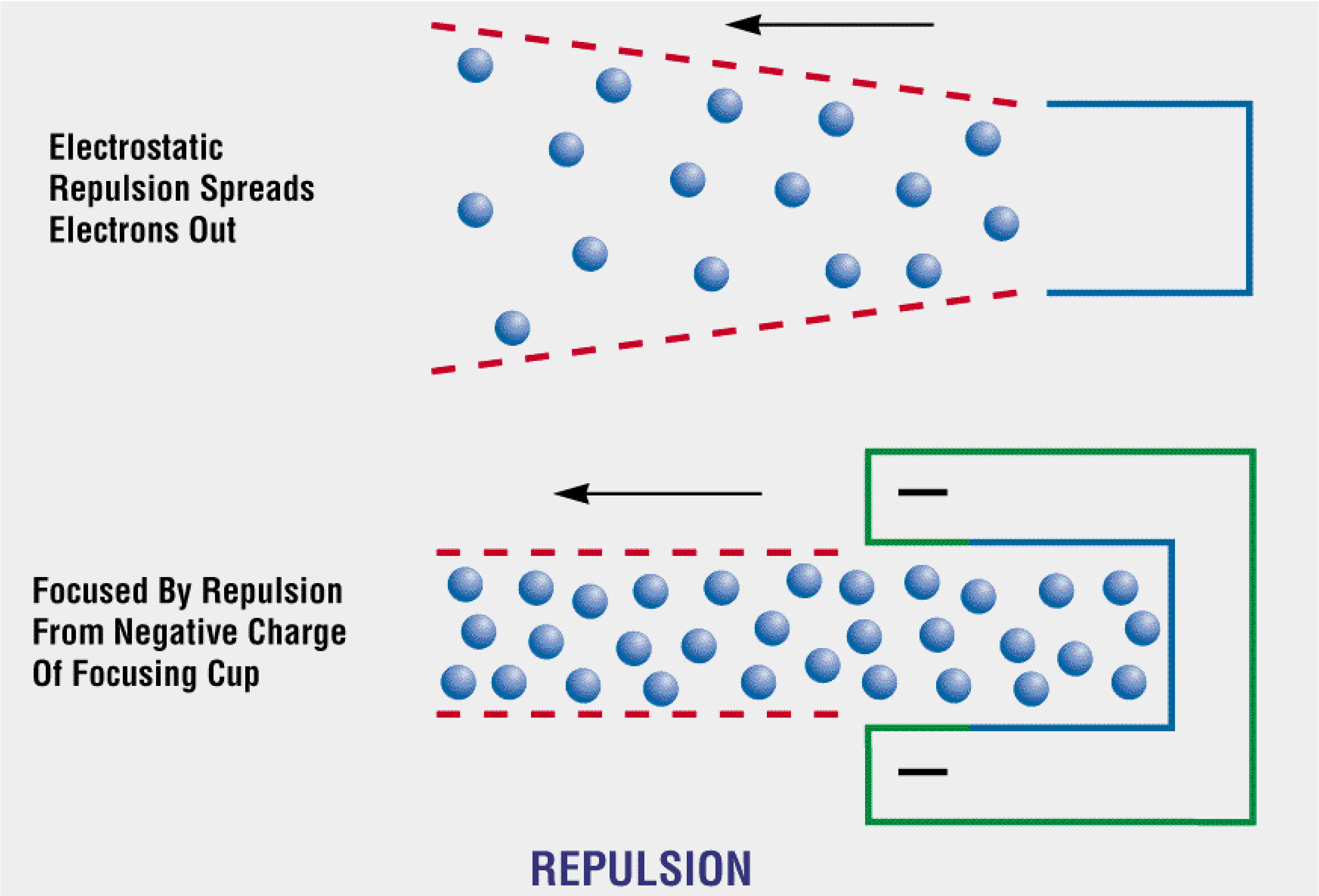
- The outer-shell electrons of the filament atom are “boiled off”.
  - This is known as thermionic emission



THERMIONIC EMISSION

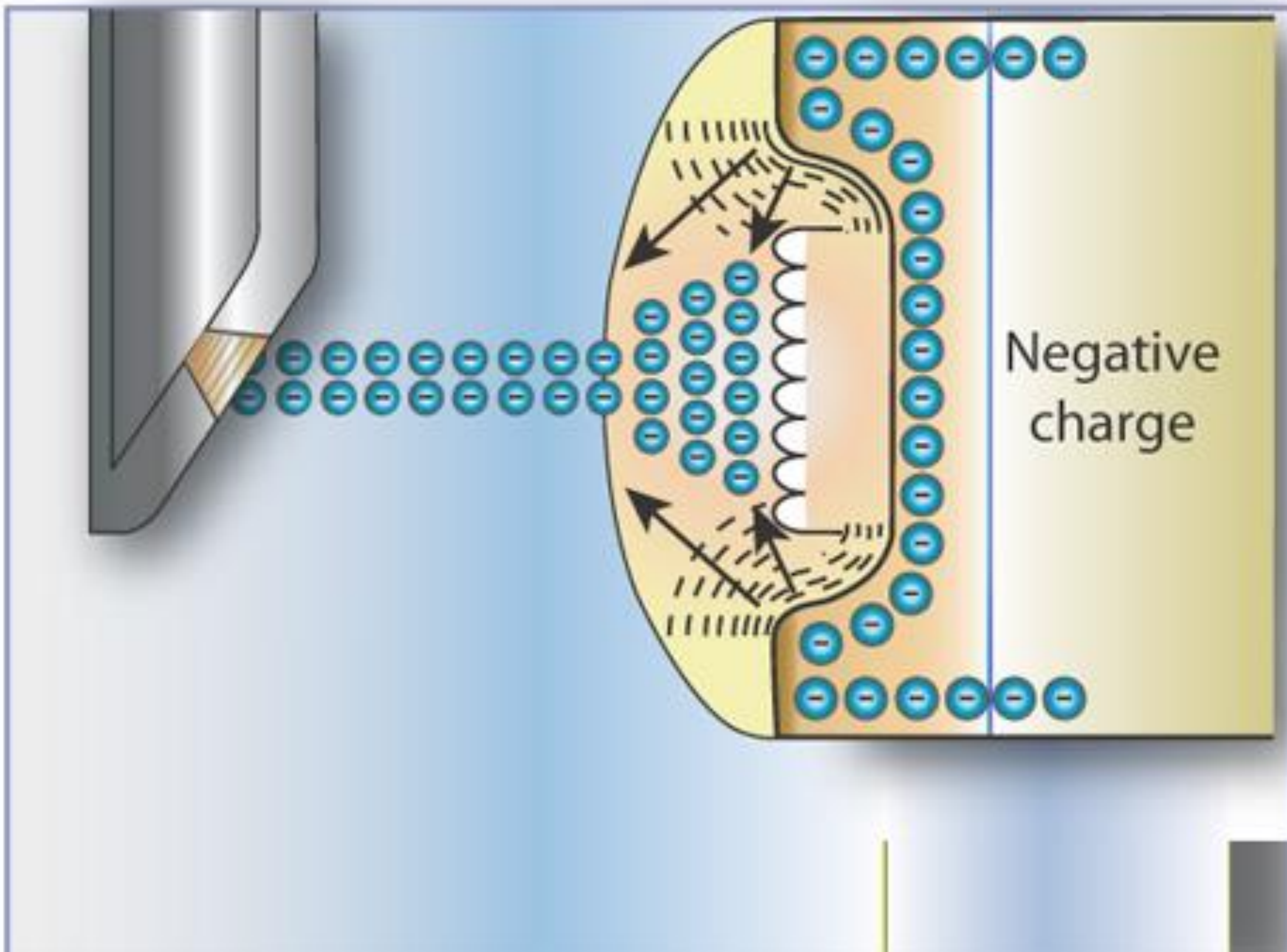
Focusing cup

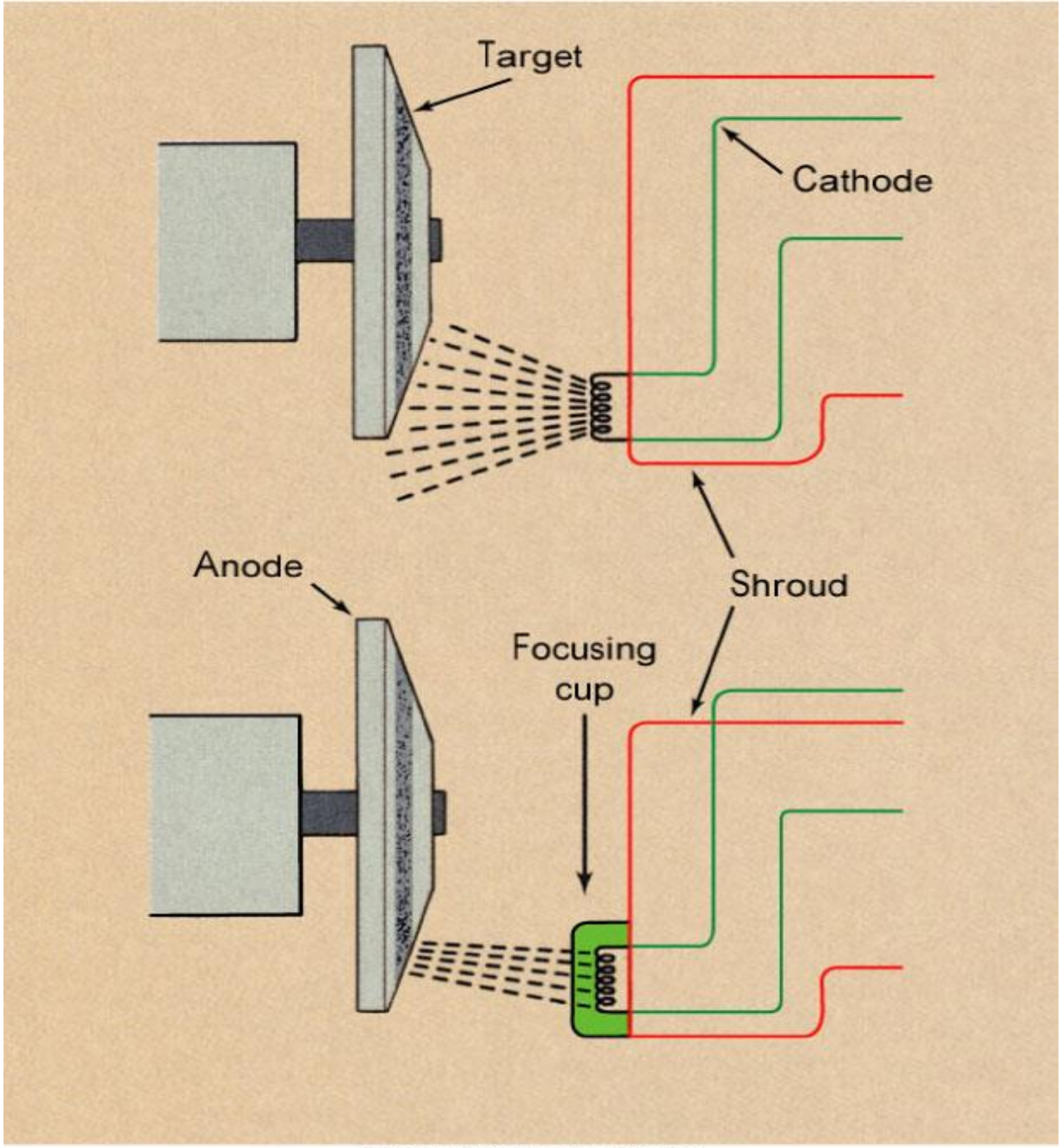
- The filament is embedded in a metal cup that has a negative charge
- Boiled off e- tend to spread out due to electrostatic repulsion. The focusing cup confines the e- cloud to a small area



Electrostatic Repulsion Spreads Electrons Out
Focused By Repulsion From Negative Charge Of Focusing Cup
—
—
REPULSION

kVp = energy   mAs = amount



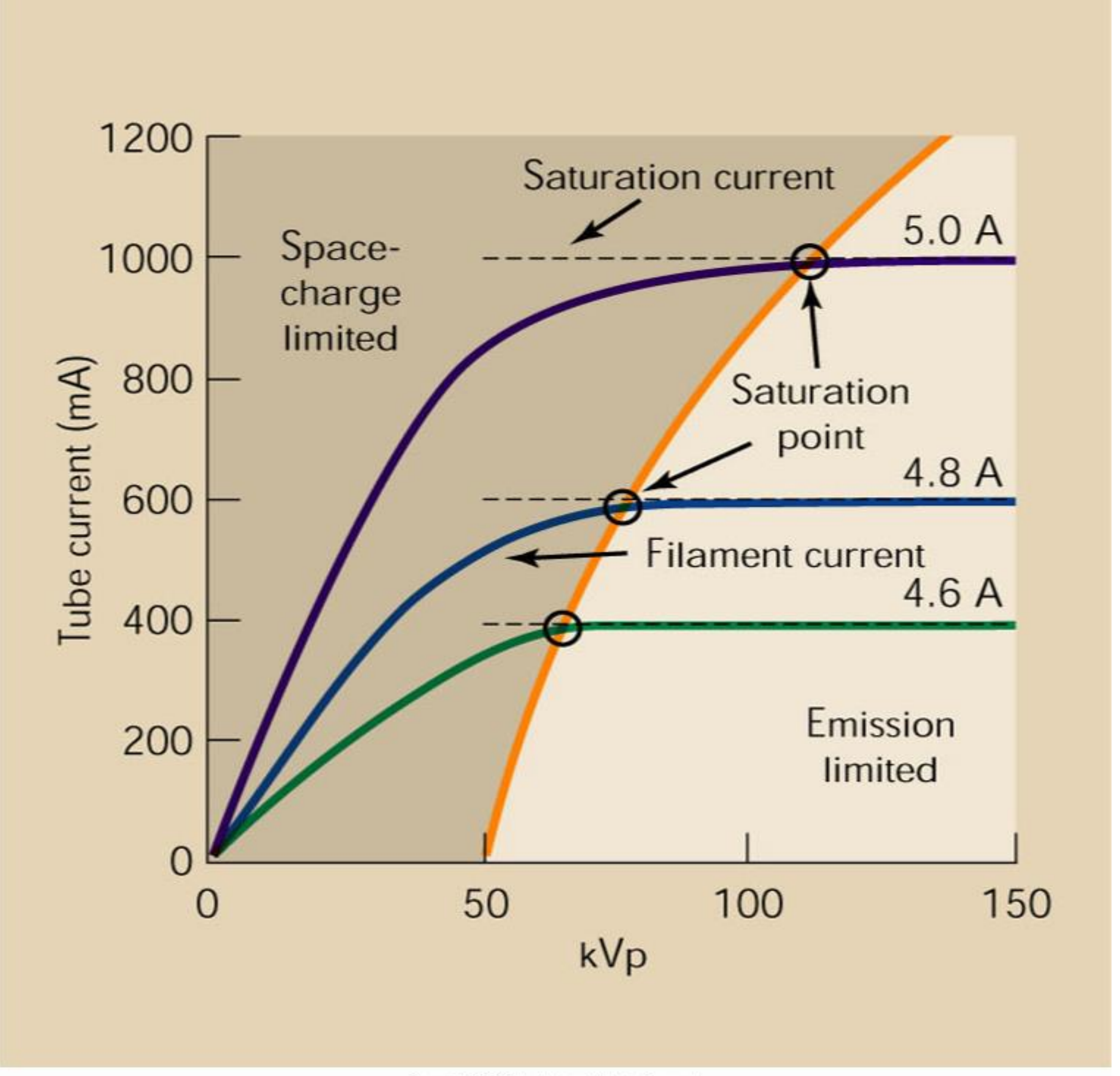

Target
Cathode
Anode
Shroud
Focusing
cup

Filament Current

- When the x-ray imaging system is first turned on, a low current passes through the filament to warm it and prepare it for the thermal jolt necessary for x-ray production
- The current is not enough to energize the tube, just warm the wire of the filament

Space-charge effect

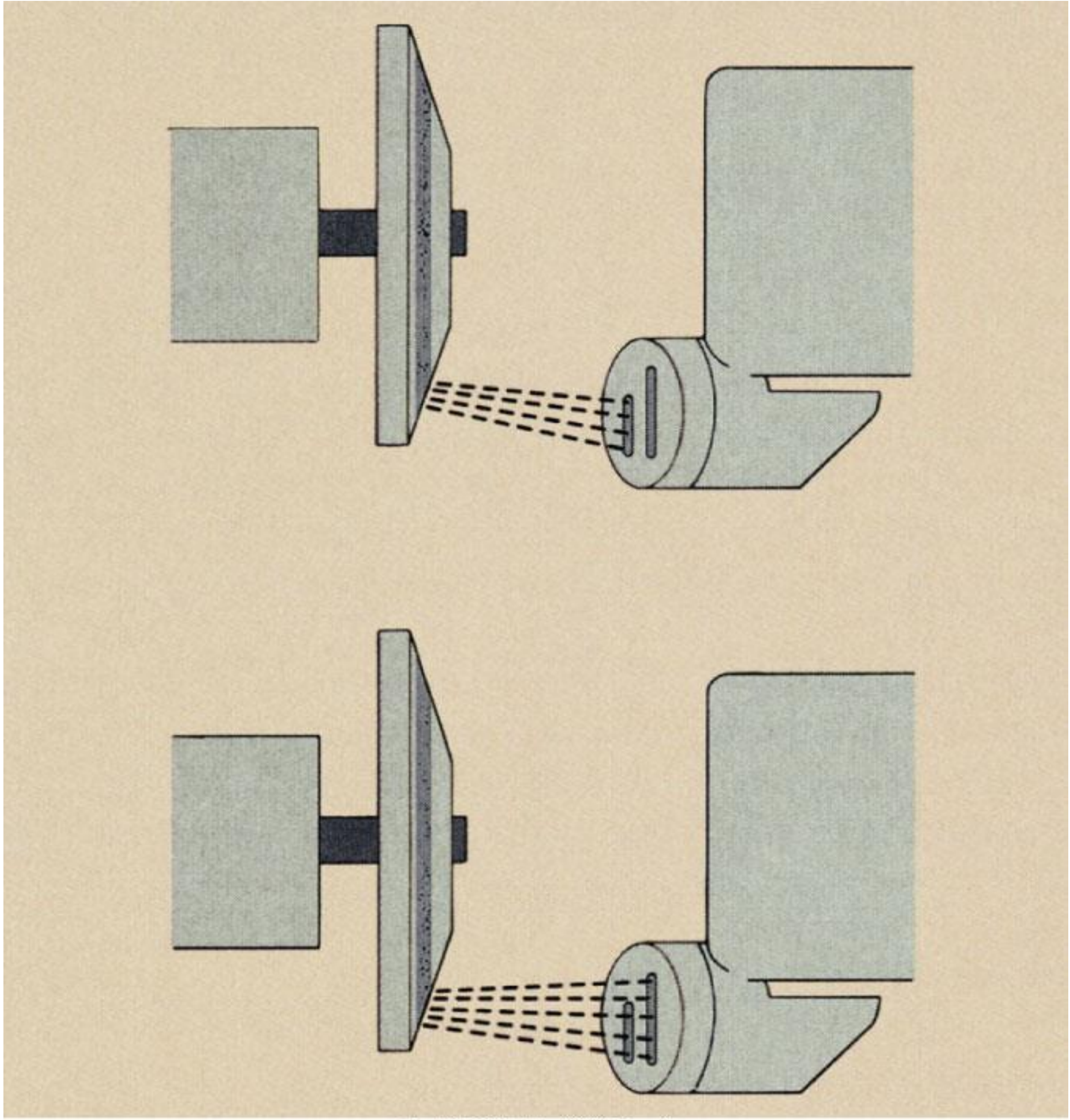
- The cloud of e- = space charge
- As the space charge becomes more negative by the boiling off of more electrons it makes it difficult for more e- to be emitted
  - Electrostatic repulsion
  - Space-charge effect
  - Space-charge limiting at low kVp & high mA



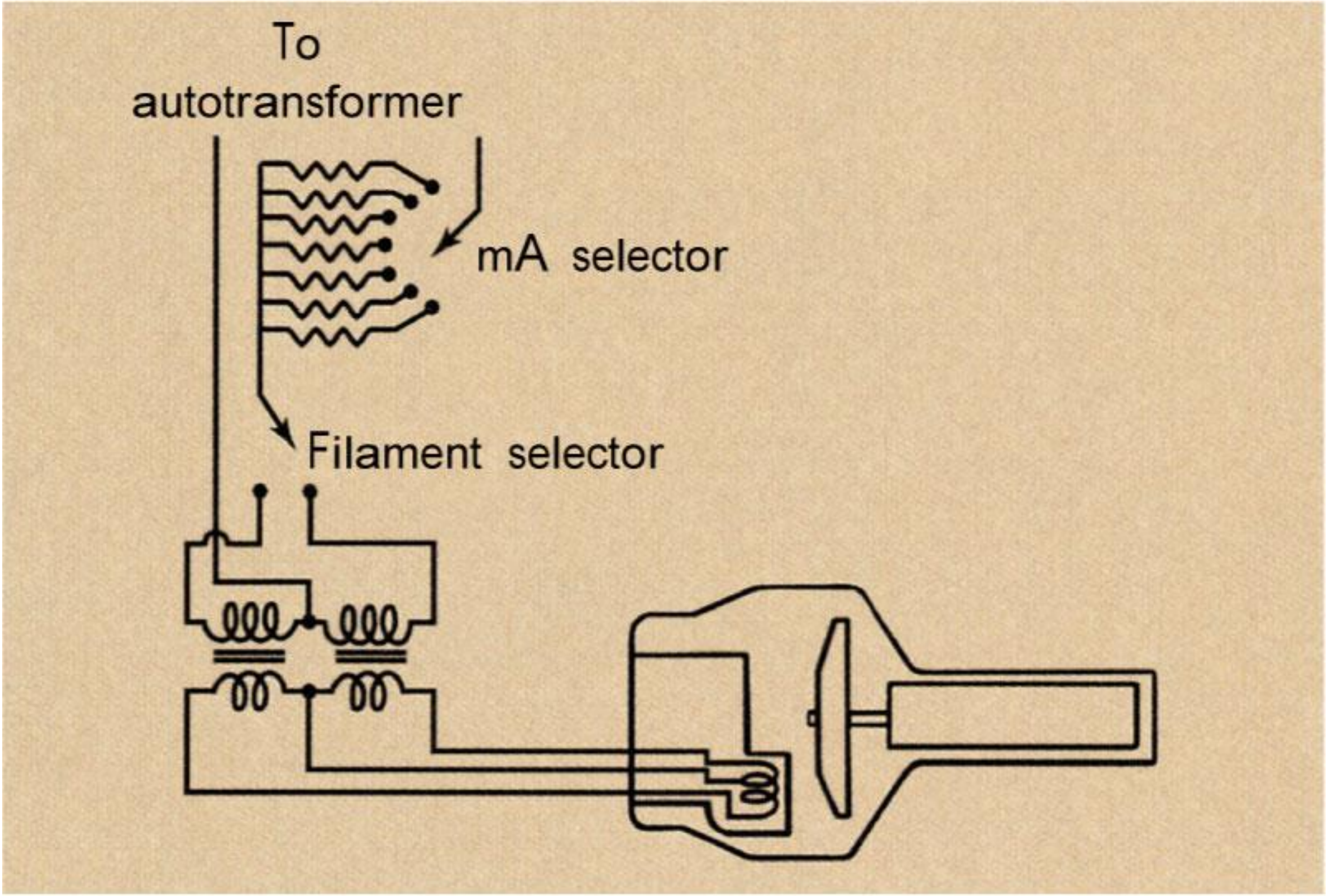
1200
1000
800
600
400
200
0
Tube current (mA)
Saturation current
5.0 A
Space-
charge
limited
Saturation
point
4.8 A
Filament current
4.6 A
Emission
limited
0
50
100
150
kVp

Dual-focus tubes

- Most diagnostic tubes have two focal spots;   large & small
- Large is used when large body parts are imaged
- Small is used when better spatial resolution is desired – better detail
- Filament size



Dual-focus tubes



(A, Courtesy The Machlett Laboratories, Inc.)

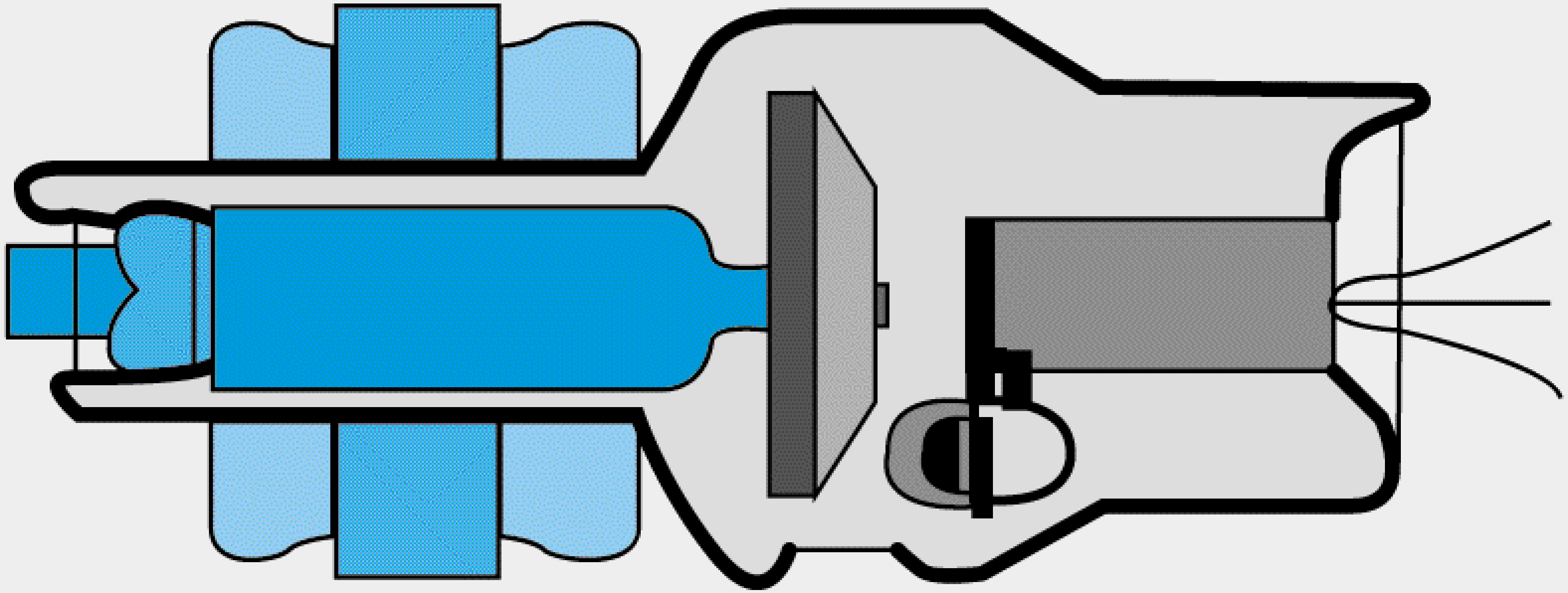
Anode

- Anode is the positive side of the x-ray tube
- The anode conducts electricity, radiates heat and contains the target
- Two types of anodes
  - Stationary & Rotating

Stationary Anode

- Used for dental x-rays, some portable imaging

- Used when high tube current and power are not required because they are not capable of producing high-intensity x-ray beams in a short time



ANODE

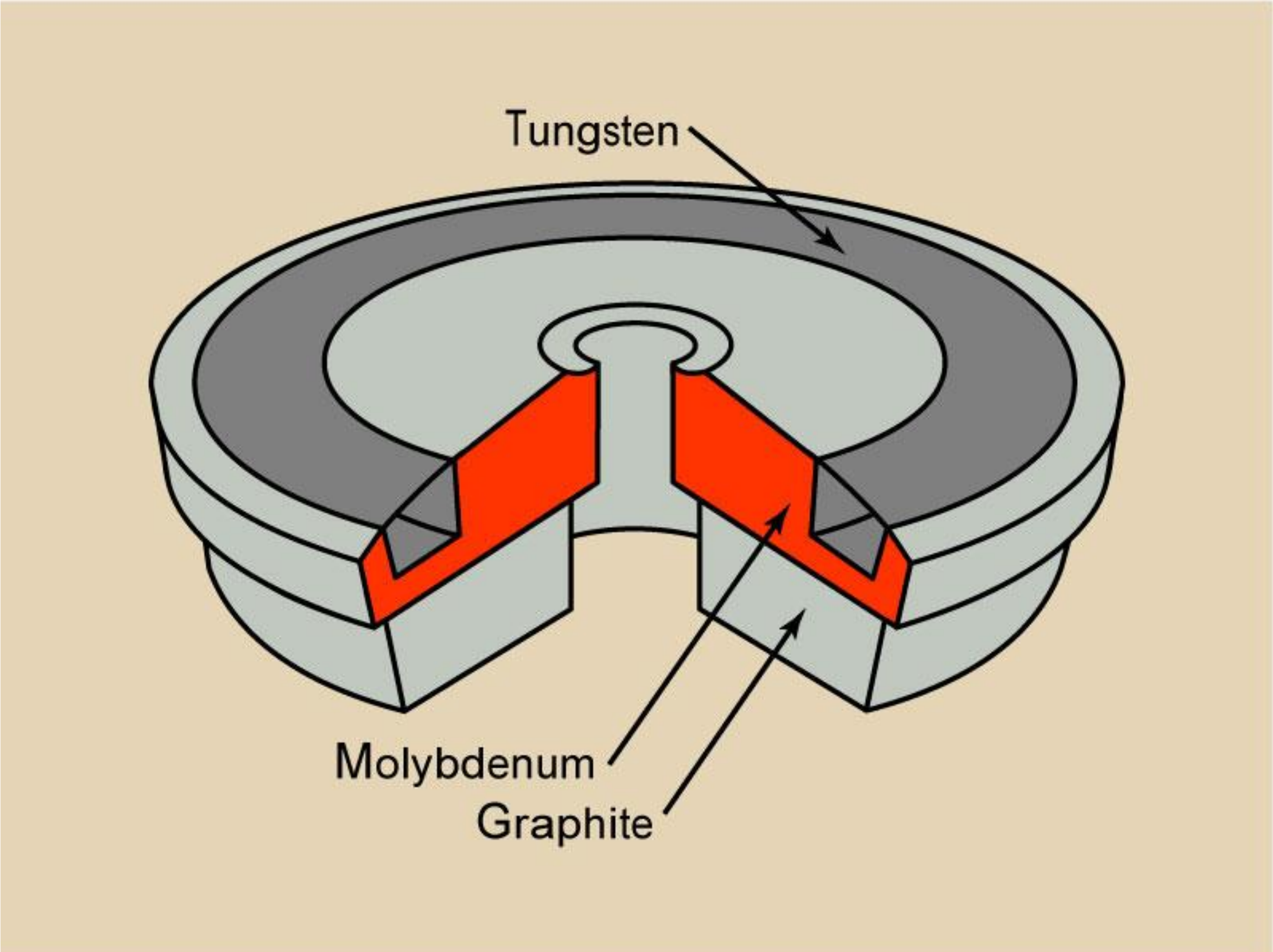
Now let's look at the anode side of the x-ray tube. Remember that the anode is the positively charged side of the tube, where the electrons strike the metal of the anode and produce the

Previous Next Repeat Hide Text Audio On

Anode Function

- Mechanical support for the target
- Dissipates heat
  - 99% of the kinetic energy from the e- is converted into heat; 1% is converted into x-rays
  - Copper, molybdenum and graphite are common anode material

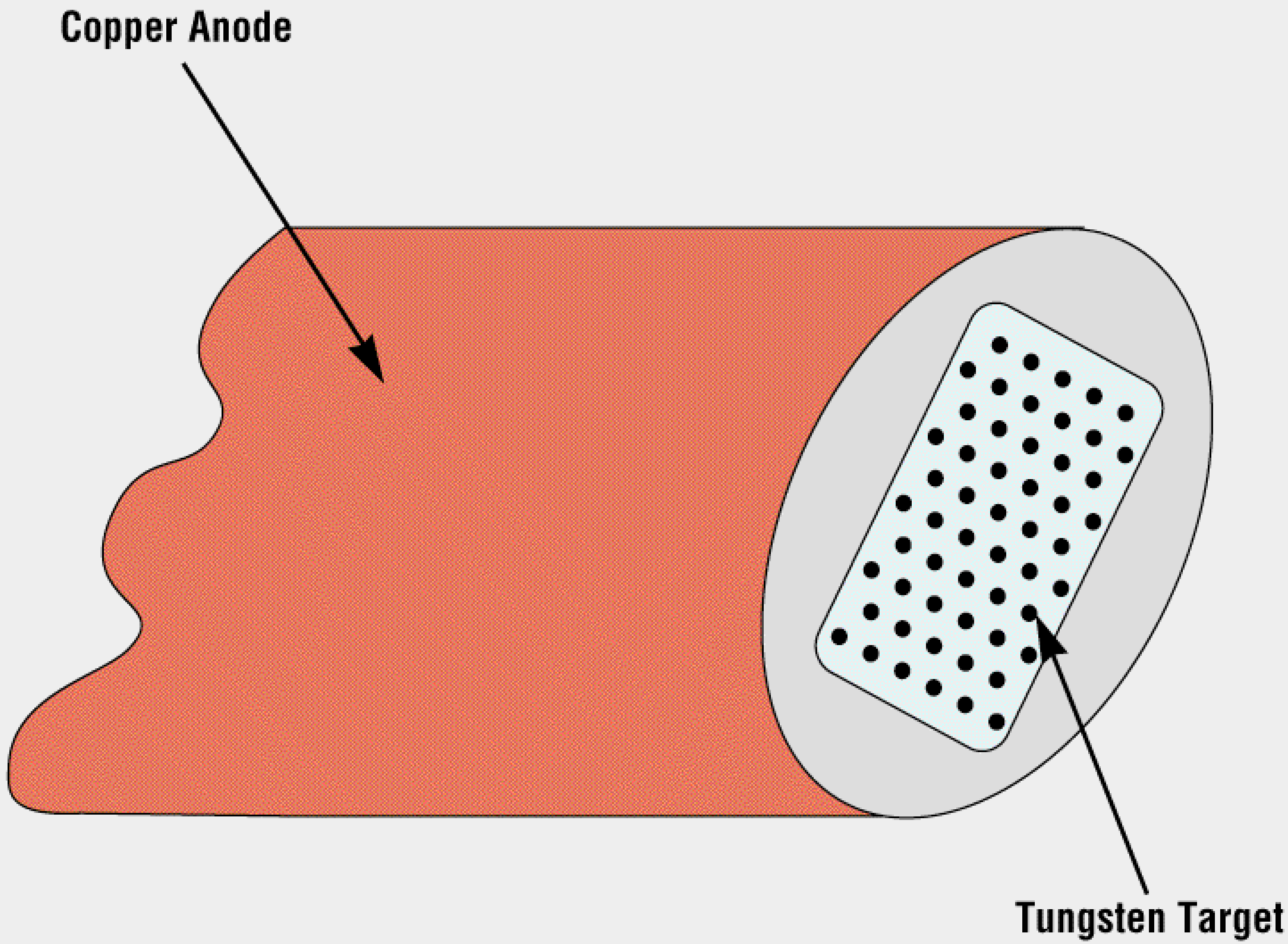
A layered anode increases heat capacity



Target

- Is the area of the anode struck by the e- from the cathode

- Tungsten is the material of choice for the target in general radiography



STATIONARY ANODE

X-ray equipment with stationary anode tubes are used in circumstances when less heat is generated in the anode. This equipment may be used for shorter exposure times or less

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Next
Repeat
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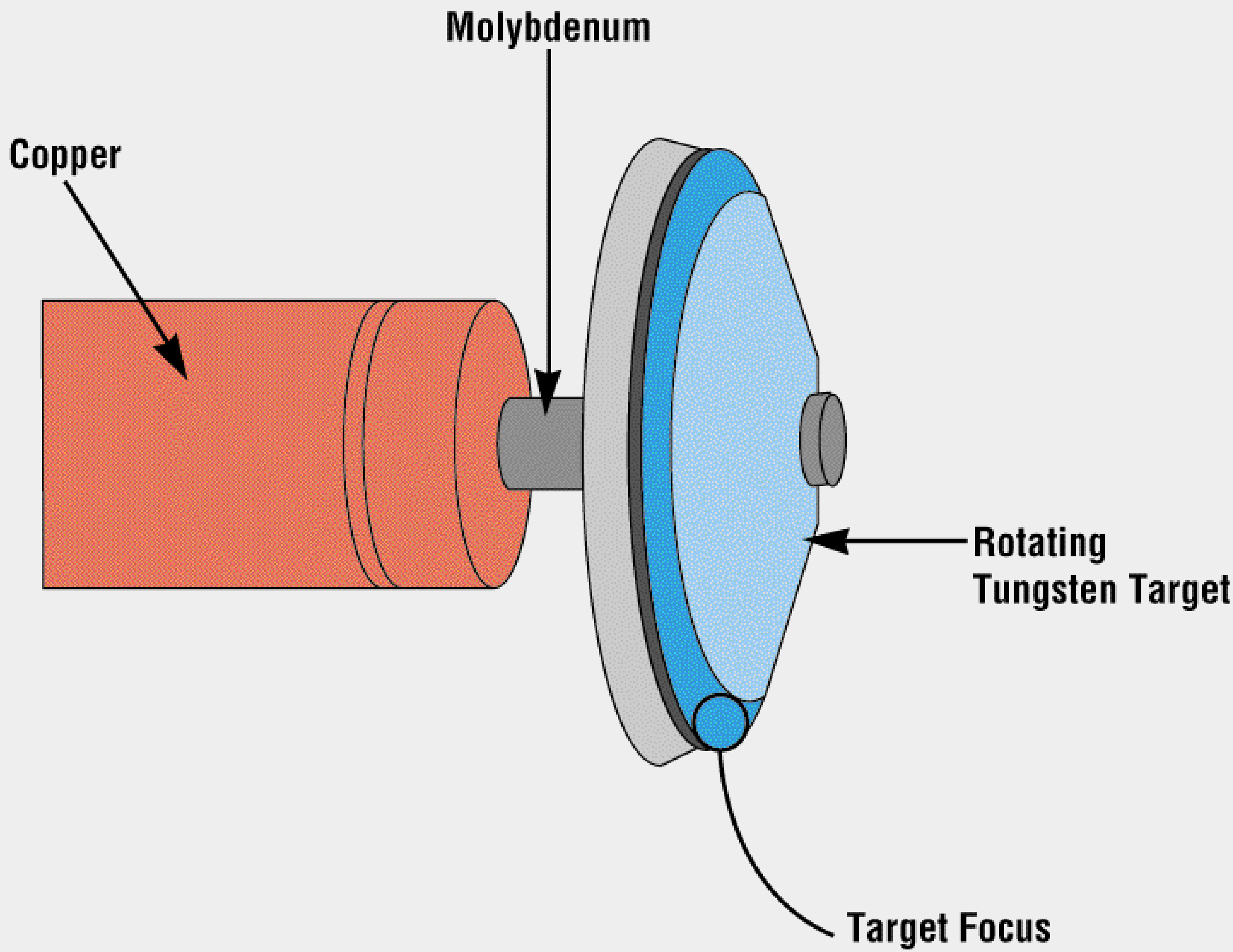


ROTATING ANODE

Rotating Anode

- Is powered by an induction motor
- The stator is on the outside of the glass, consist of a series of electromagnets
- The rotor is a shaft made of bars of copper and soft iron built into one mass

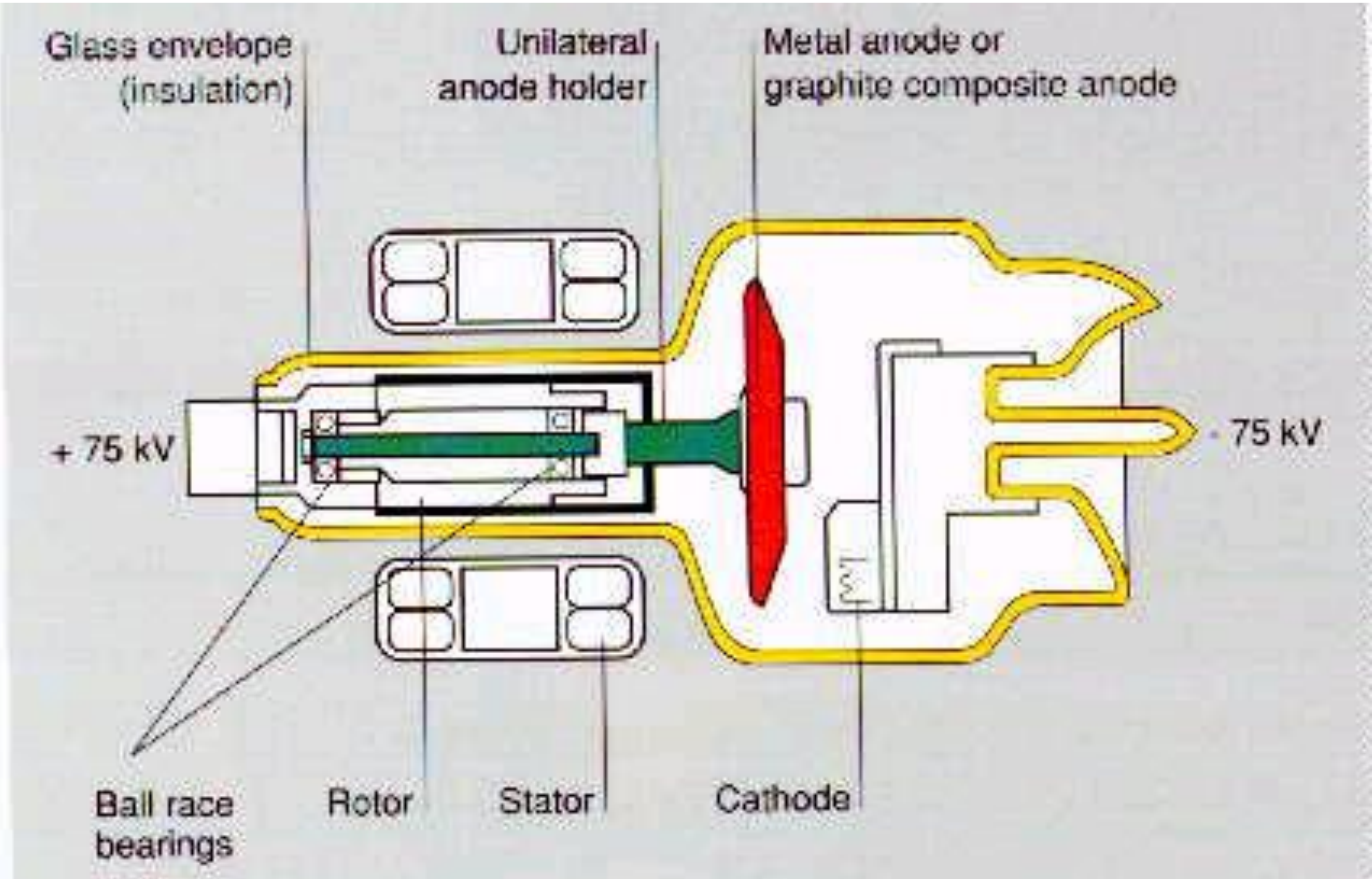
Electromagnetic induction

- As current is applied to the stator sequentially so the magnetic field rotates on the axis of the stator
- This magnetic field interacts with the metal (ferromagnetic rotor) causing it to rotate in unison with the magnetic field of the stator

Electromagnetic Induction Motor

- Anode speed
- Average 3,600 rpm (revolutions per minute)
- High capacity 10,000 rpm

- Anode Cooling Chart
  - Heat Units (HU)


Glass envelope
(insulation)
Unilateral
anode holder
Metal anode or
graphite composite anode
+ 75 kV
- 75 kV
Ball race
bearings
Rotor
Stator
Cathode

Dead-man switch

- Rotor/Prep – applies current (mA) to the tube
  - Allows rotor to accelerate to its designed RPM. Rotor stops about 1 min after exposure
  - Filament current is increased to create e-cloud
- Exposure – applies voltage (kV) to make exposure

Focal spot

- The area of the anode's target where x-rays are emitted
- The smaller the focal spot the better the resolution of the resultant image

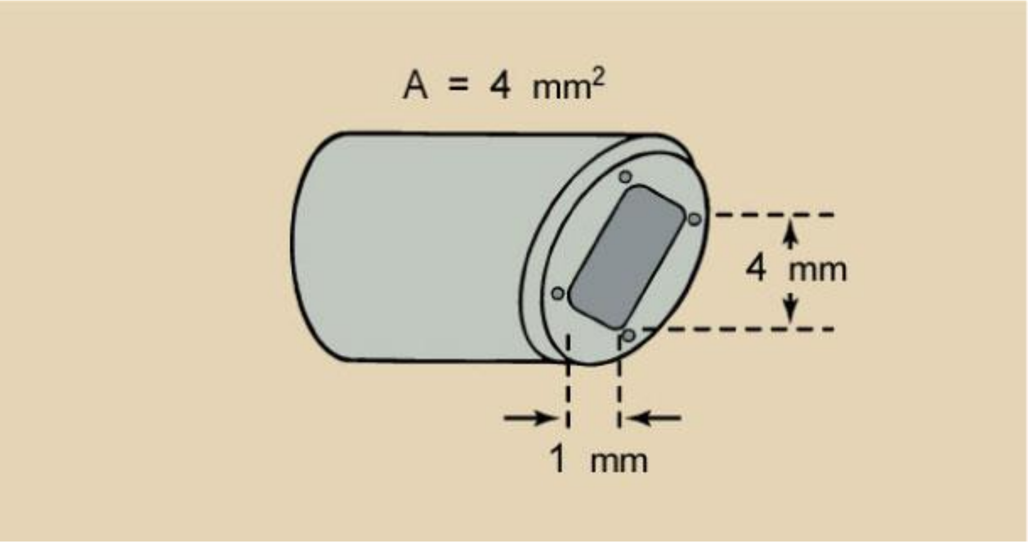
Focal spot

- Unfortunately, as the size of the focal spot decreases, the heat of the target is concentrated into a smaller area

- This is the limiting factor to focal spot size

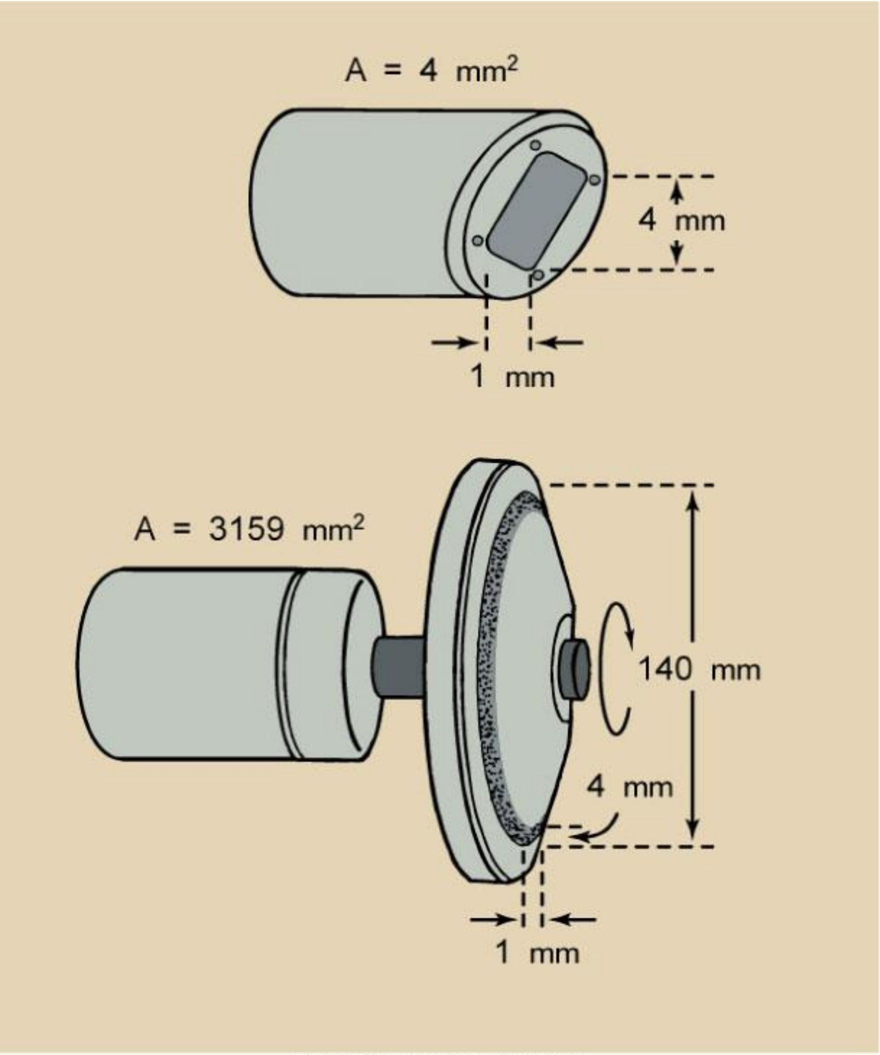
Line-focus principle

- By angling the target, the effective area of the target is much smaller than the actual area of electron interaction



Line-focus principle

- **Effective Focal Spot**

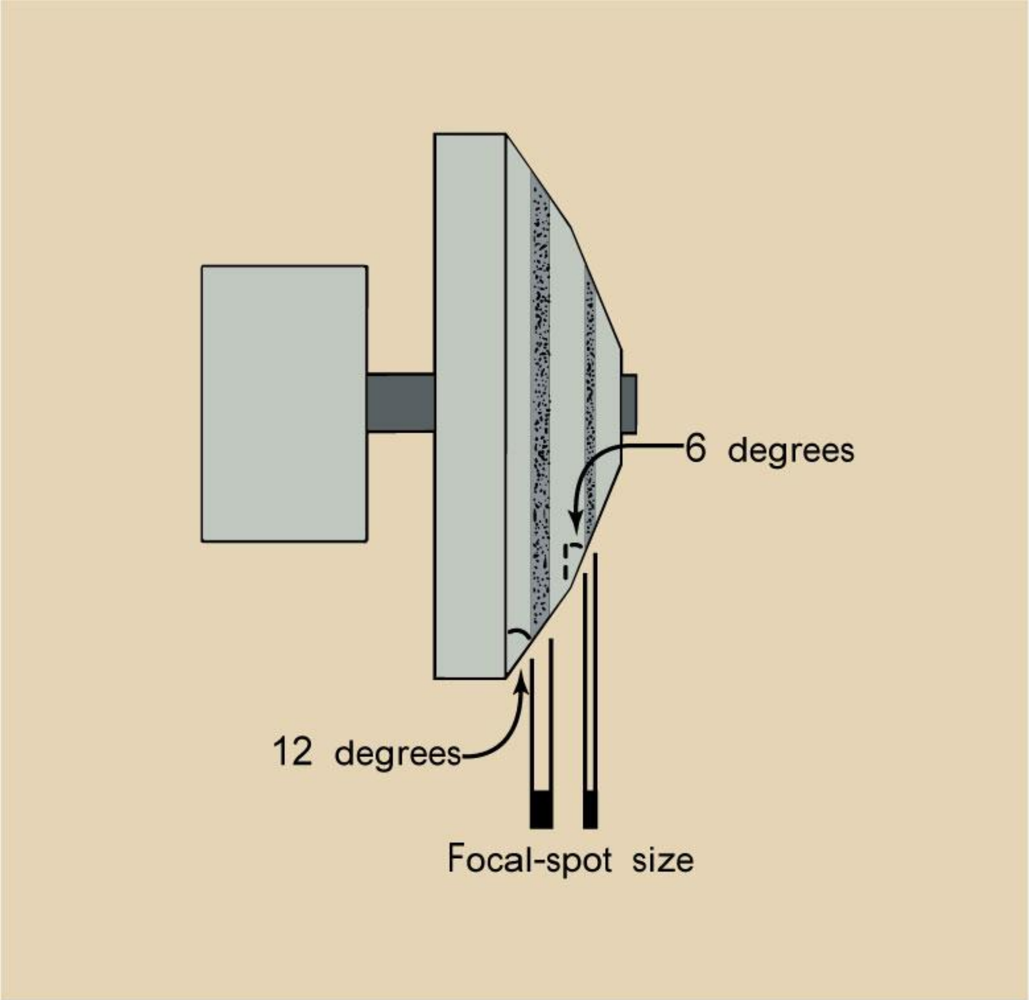


Target angle

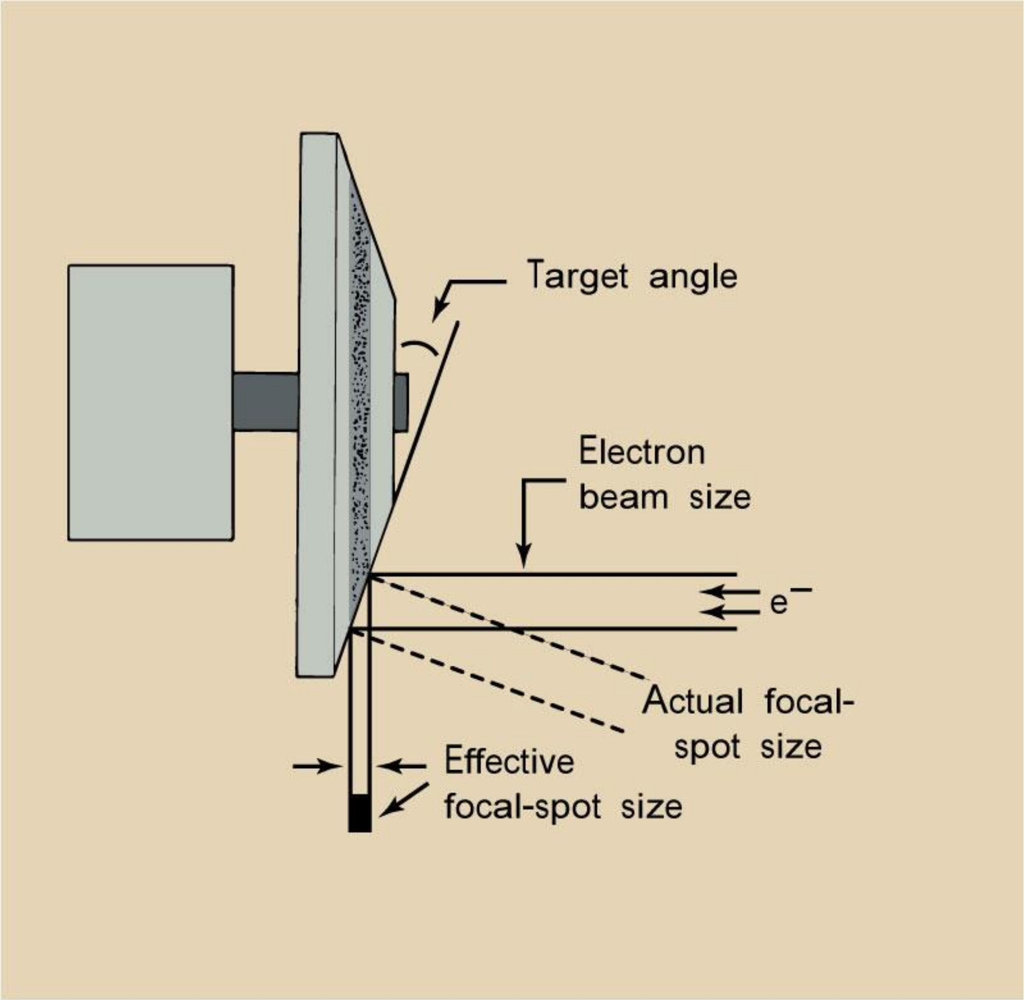
- The smaller the target angle the smaller the effective focal spot
- Angles from 5 degrees to 15 degrees

- Biangular targets are available that produce two focal spot sizes

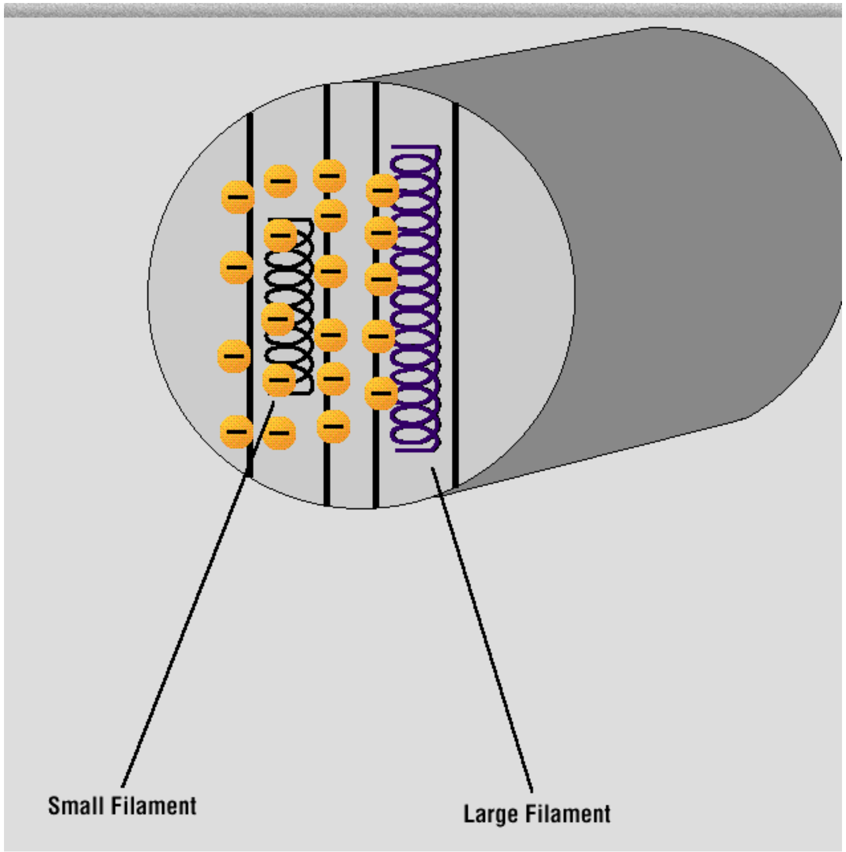
Biangular targets

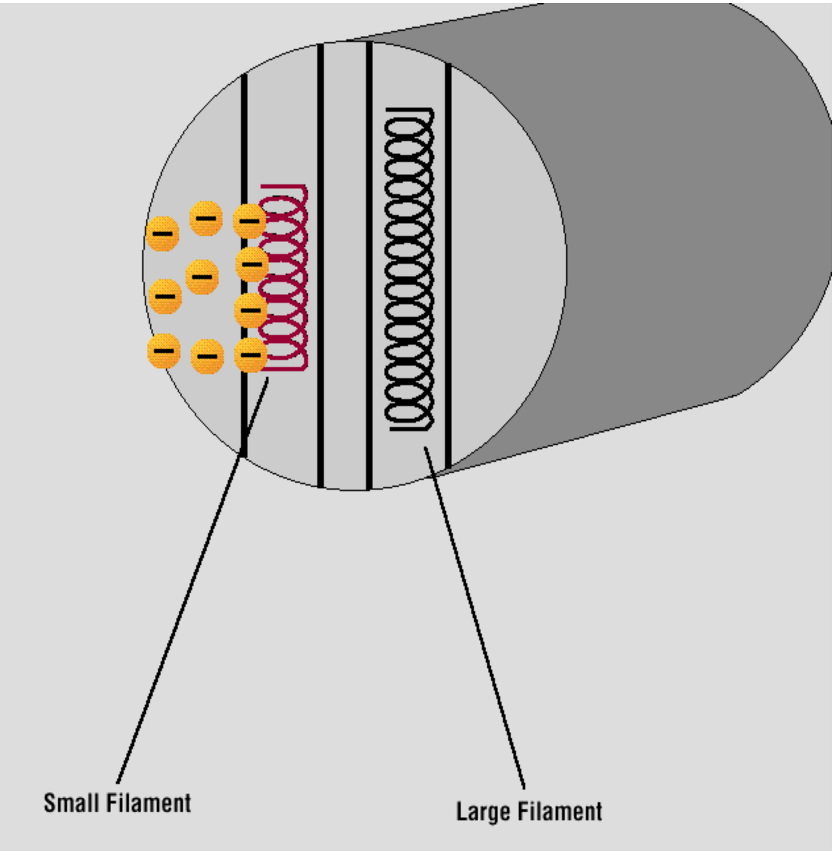


The second factor of effective focal spot is the incoming size of e- stream



Focal spot size of the cathode





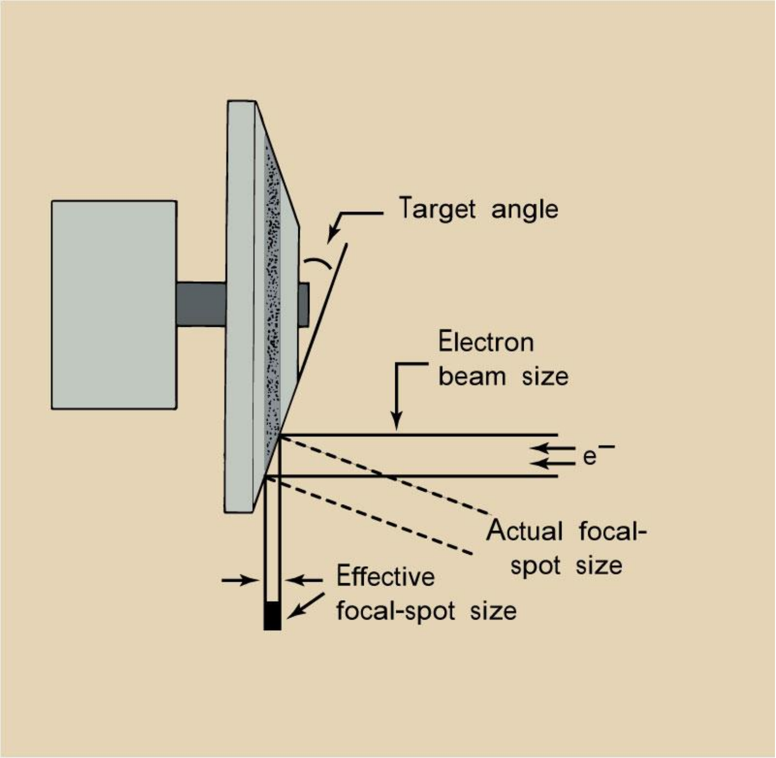
Anode Heel Effect

- Because of the use of line-focus principle the consequence is that the radiation intensity on the cathode side of the x-ray field is higher than that on the anode side
- “Fat Cat”

Heel Effect

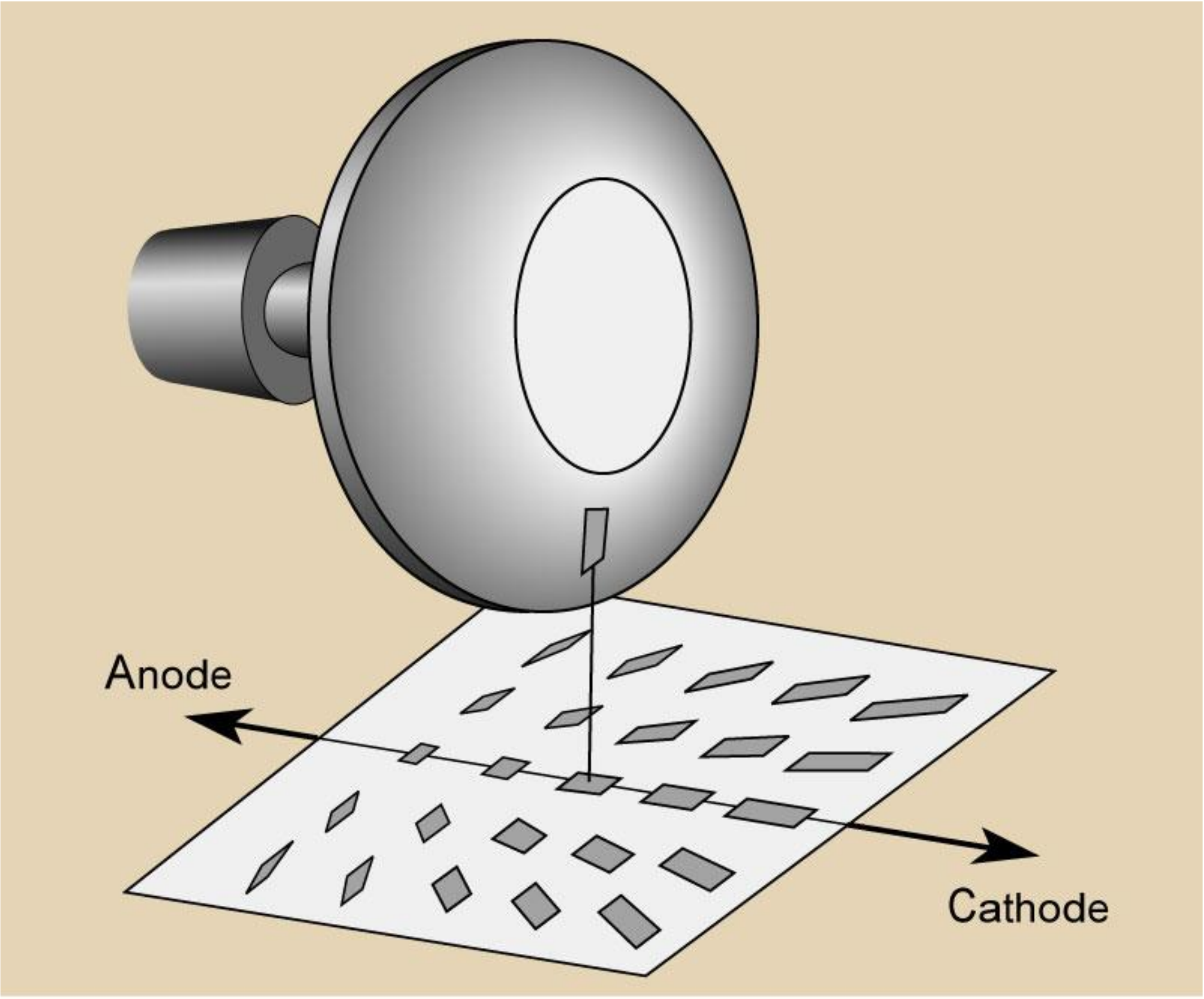
- Because the e- on the anode side must travel further than the e- that are close to the cathode side of the target, the anode side x-rays have slightly lower energy than the cathode side x-rays

Anode Heel Affect



- “Fat Cat”
- The smaller the anode angle, the larger the heel affect

Anode Heel Affect



Extrafocal Radiation

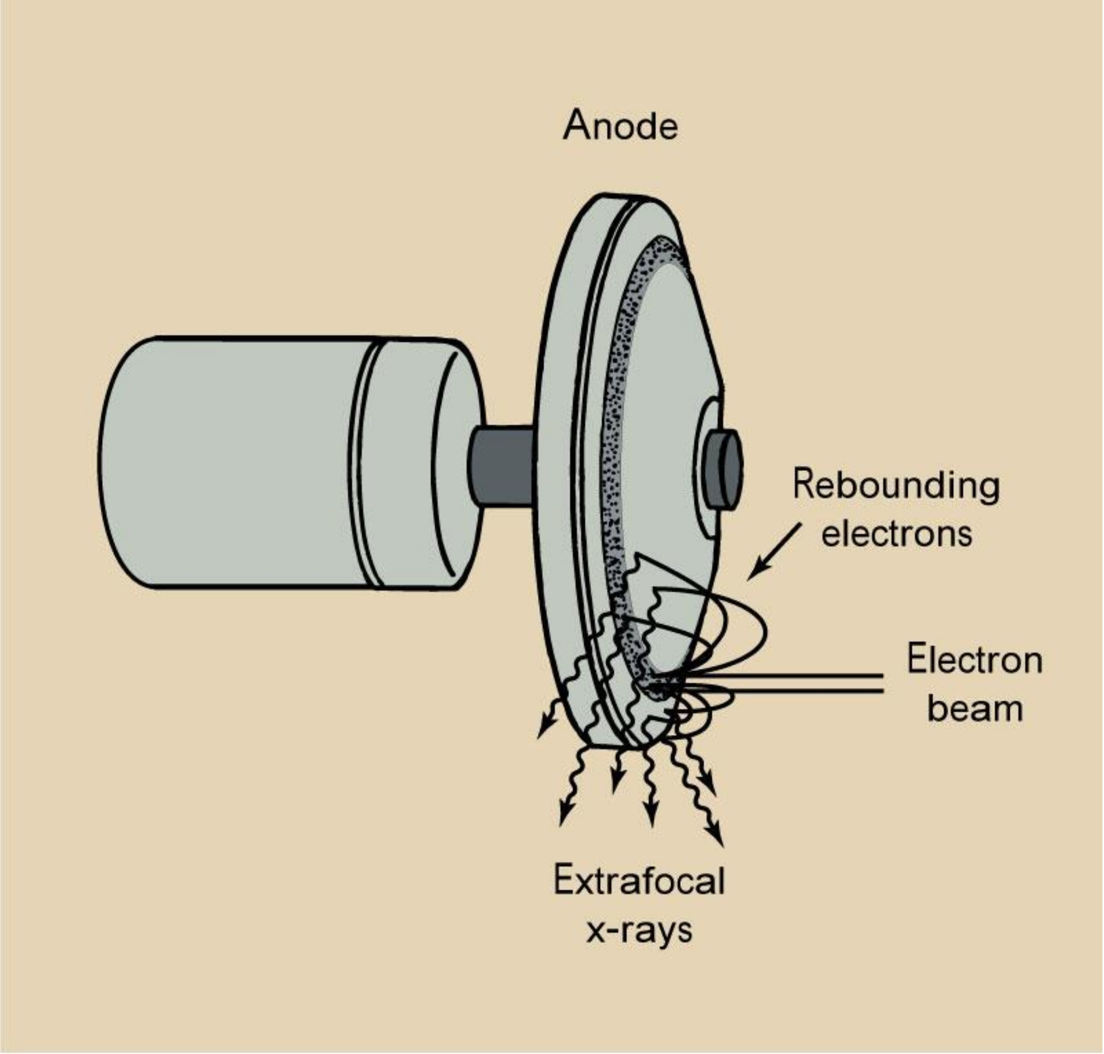
- X-ray tubes are designed so that the projectile e- interacts with the target. However, some of the e- bounce off the target and land on other areas

- This caused x-rays to be produced out side the focal spot

Extrafocal Radiation

- These rays can also be called off-focus radiation
- Extrafocal radiation is undesirable because it extends the size of the focal spot, increases patient skin dose & reduces image contrast

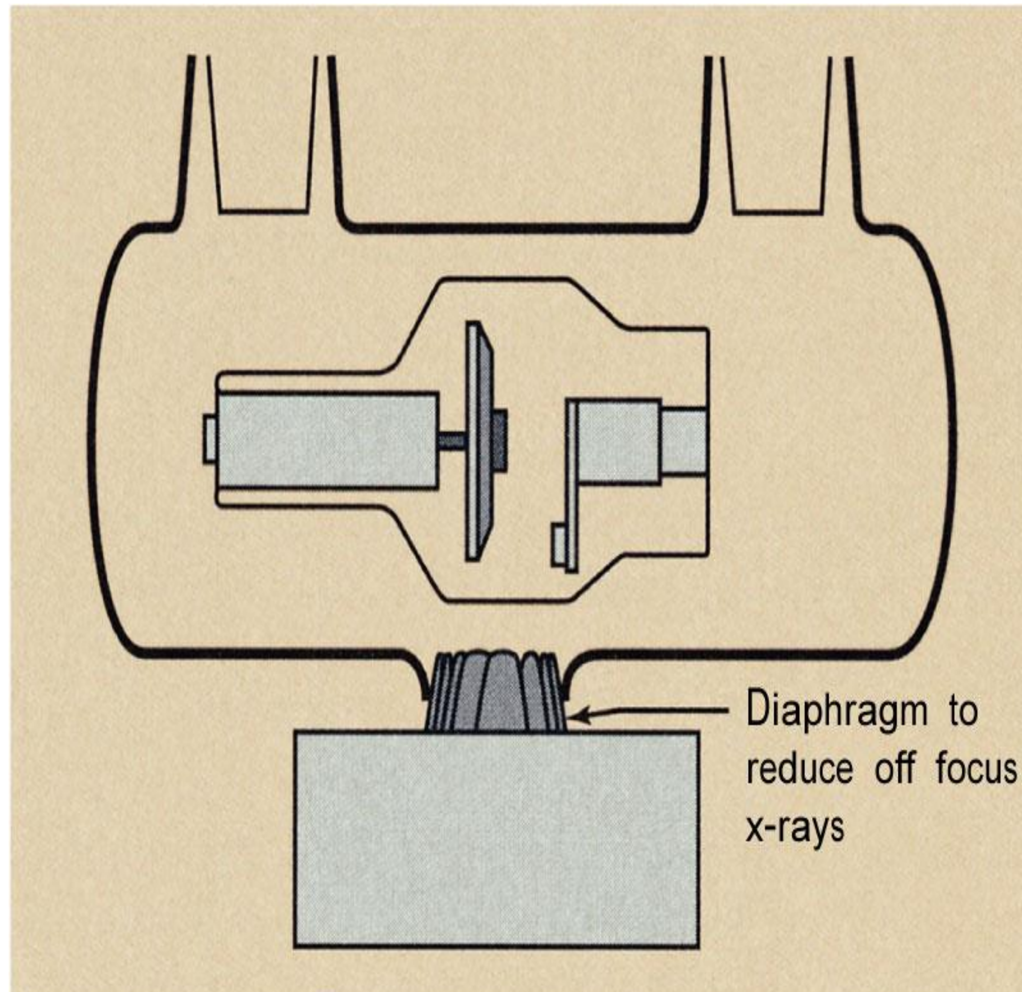
Off-focus radiation

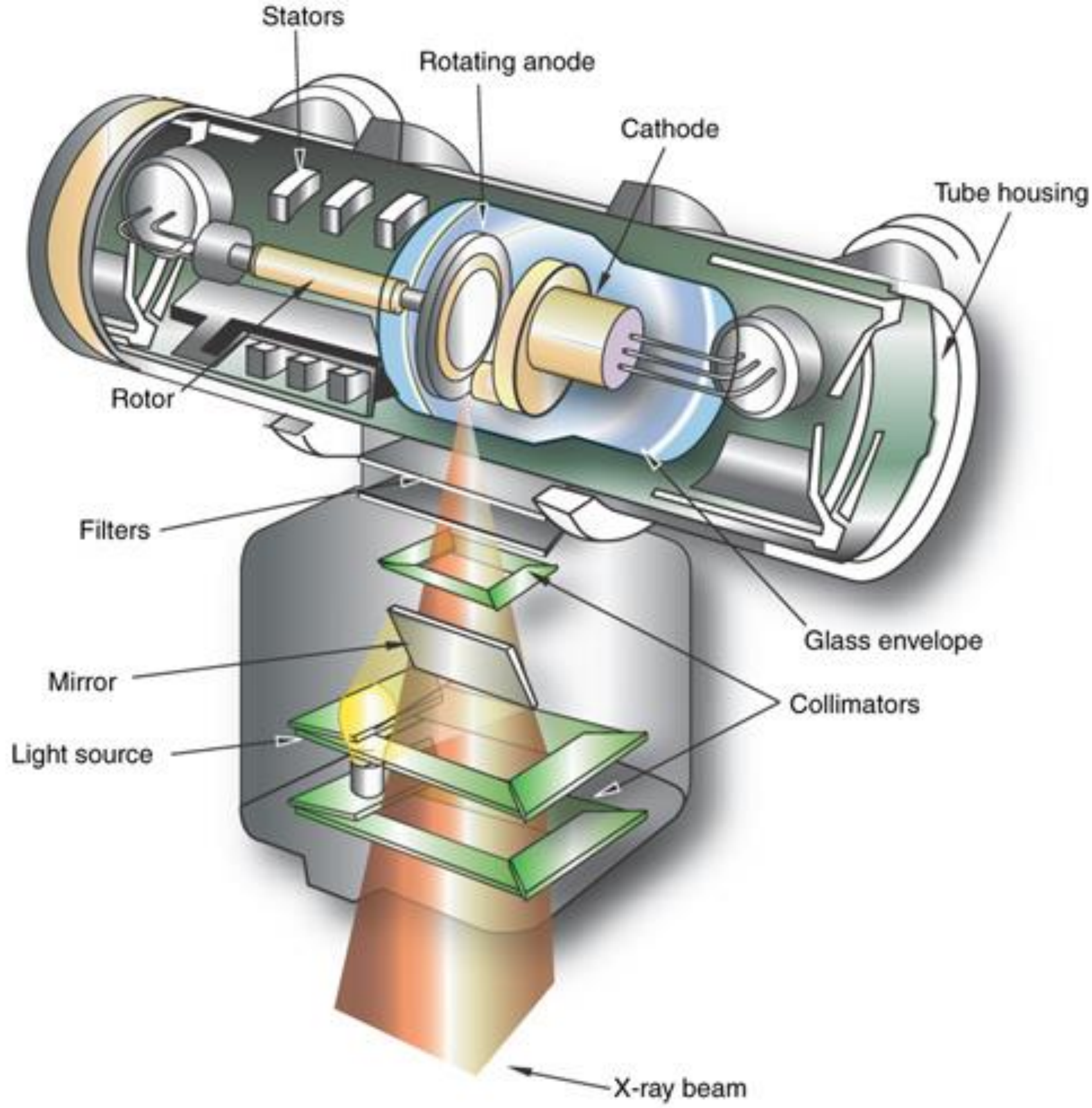


Fixed diaphragm in the tube housing

- Using a grid

does not reduce extrafocal radiation




Stators
Rotating anode
Cathode
Tube housing
Rotor
Filters
Glass envelope
Mirror
Collimators
Light source
X-ray beam

The Control Console

- The control console is device that allows the technologist to set technical factors (mAs & kVp) and to make an exposure.
- Only a legally licensed individual is authorized to energize the console.

Kilovoltage Peak

- kVp
- One kilovolt is = to 1000 volts

- The amount of voltage selected for the
x-ray tube
- Range 45 to 120 kVp (diagnostic range)
- kVp controls contrast

Milliamperage

- mA
- One milliampere is equal to one thousandth of an ampere.
- The amount of current supplied to the

x-ray tube

- Range 10 to 1200 mA

Time

- In seconds
- How long x-rays will be produced
- 0.001 to 6 seconds

mAs

mA X s = mAs

Where does the “POWER” come from?

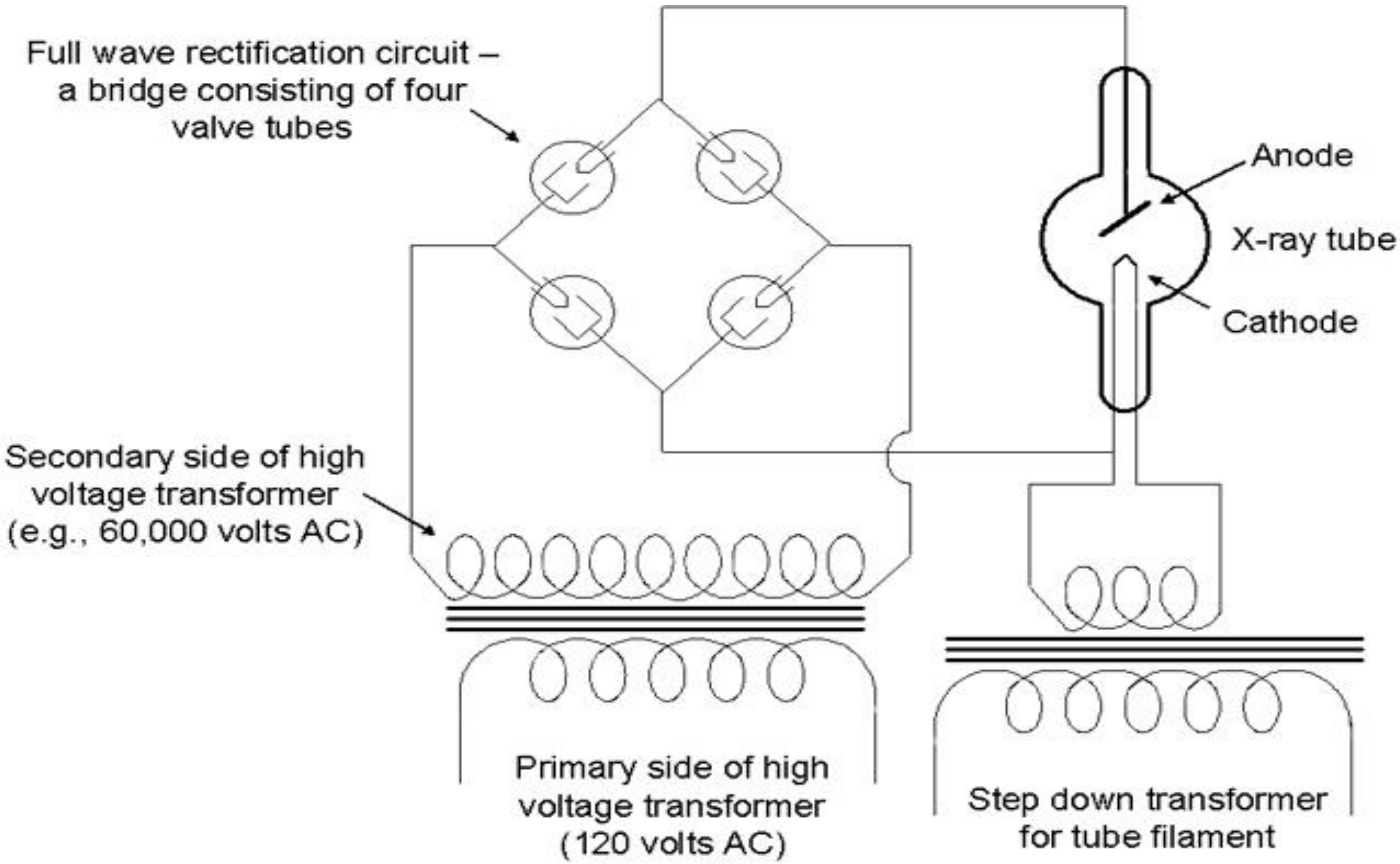
- Circuitry to be covered in detail next year
- Basic Information:
- Transformers are used to boost up the power from the incoming line to the x-ray tube
- 220Volts incoming – up to 120,000 volts (120kVp) to anode side of x-ray tube

Where does the "POWER" come from?

- Voltage current is reduced to milliamps to the filament (cathode) side of the tube.

- The difference in low (-) charge current on the filament side – and the high (+) voltage on the anode side is what helps to attract the electron to charge across the tube

X-Ray Tube Circuit



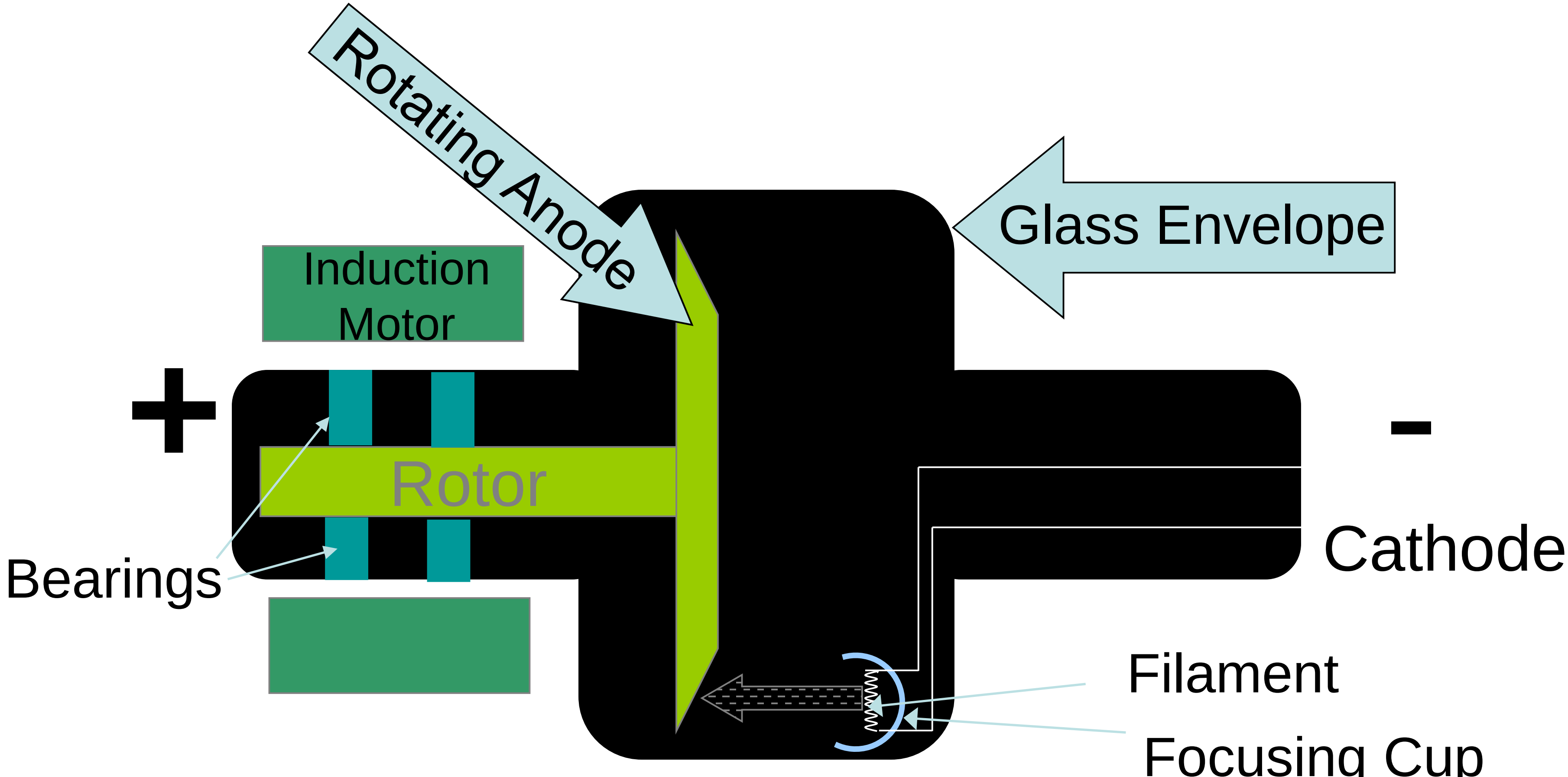
DENSITY & CONTRAST

- KVP = CONTROLS CONTRAST
- (DIFFERENCES FROM BLACK TO WHITE

- MAS – DENSITY
- AMOUNT OF BLACK ON THE FILM

"I can assure you our x-ray procedures follow very strict health and safety guidelines."

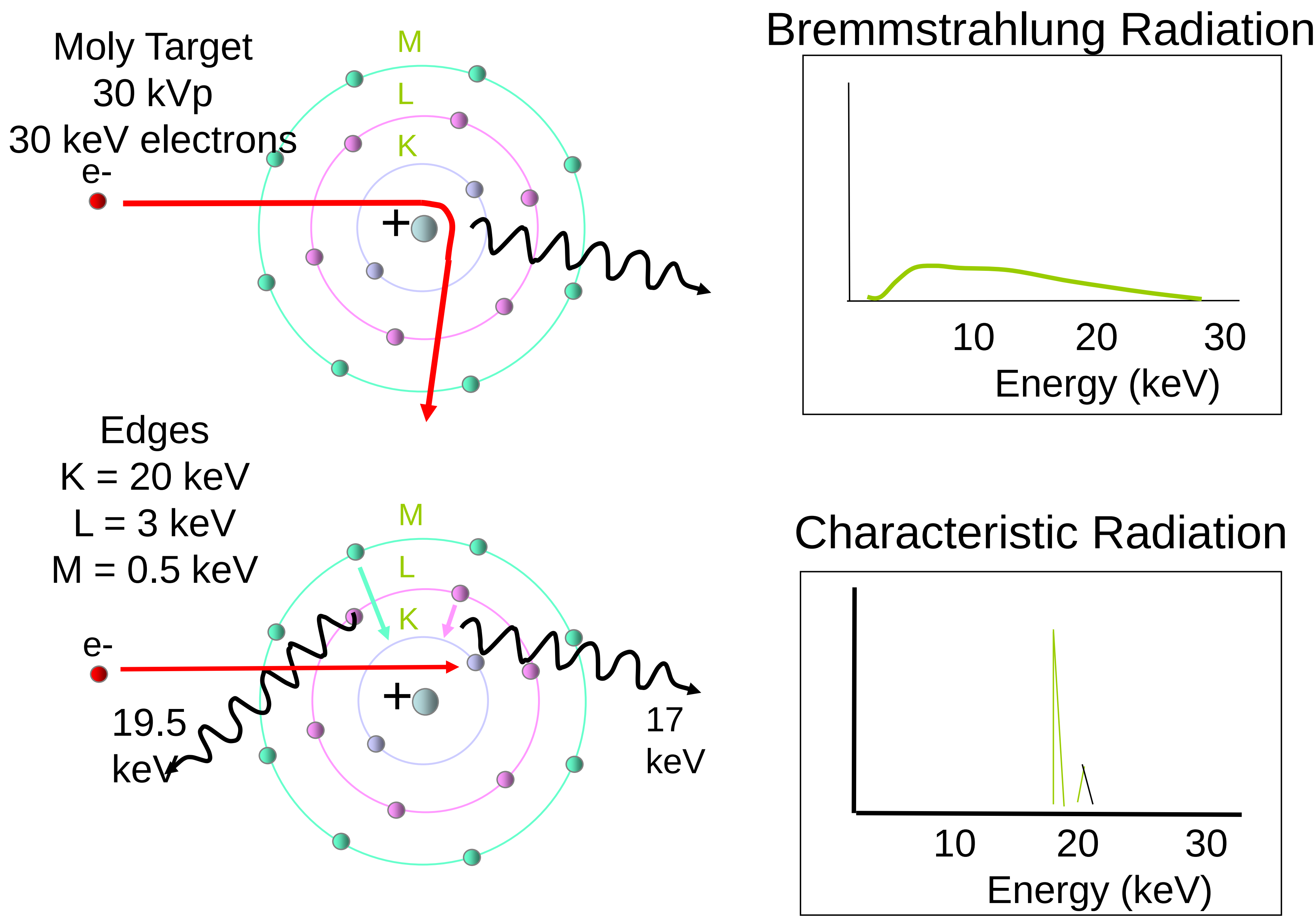
Modern X-ray Tube



X-ray Production

X-rays are produced when electrons give up kinetic energy.

- Changing direction because of the attraction of the nucleus requires the electron to give up some of it's kinetic energy.
  - **Bremsstrahlung**
- The electron knocks an atom's bound electron out. A "hole" is created. When a new electron fills the "hole" it gives up energy.
  - **Characteristic**



Moly Target
30 kVp
30 keV electrons
e-
M
L
K
Bremmstrahlung Radiation
10
20
30
Energy (keV)
Edges
K = 20 keV
L = 3 keV
M = 0.5 keV
e-
19.5
keV
17
keV
Characteristic Radiation
10
20
30
Energy (keV)

X-Ray Beam Descriptors

- Quantity (Intensity)
  - Number of x-ray photons produced
- Quality (Energy Distribution) –
  - A measure of the penetrating power of the x-ray beam
  - Shape of the x-ray spectrum
  - Maximum photon energy

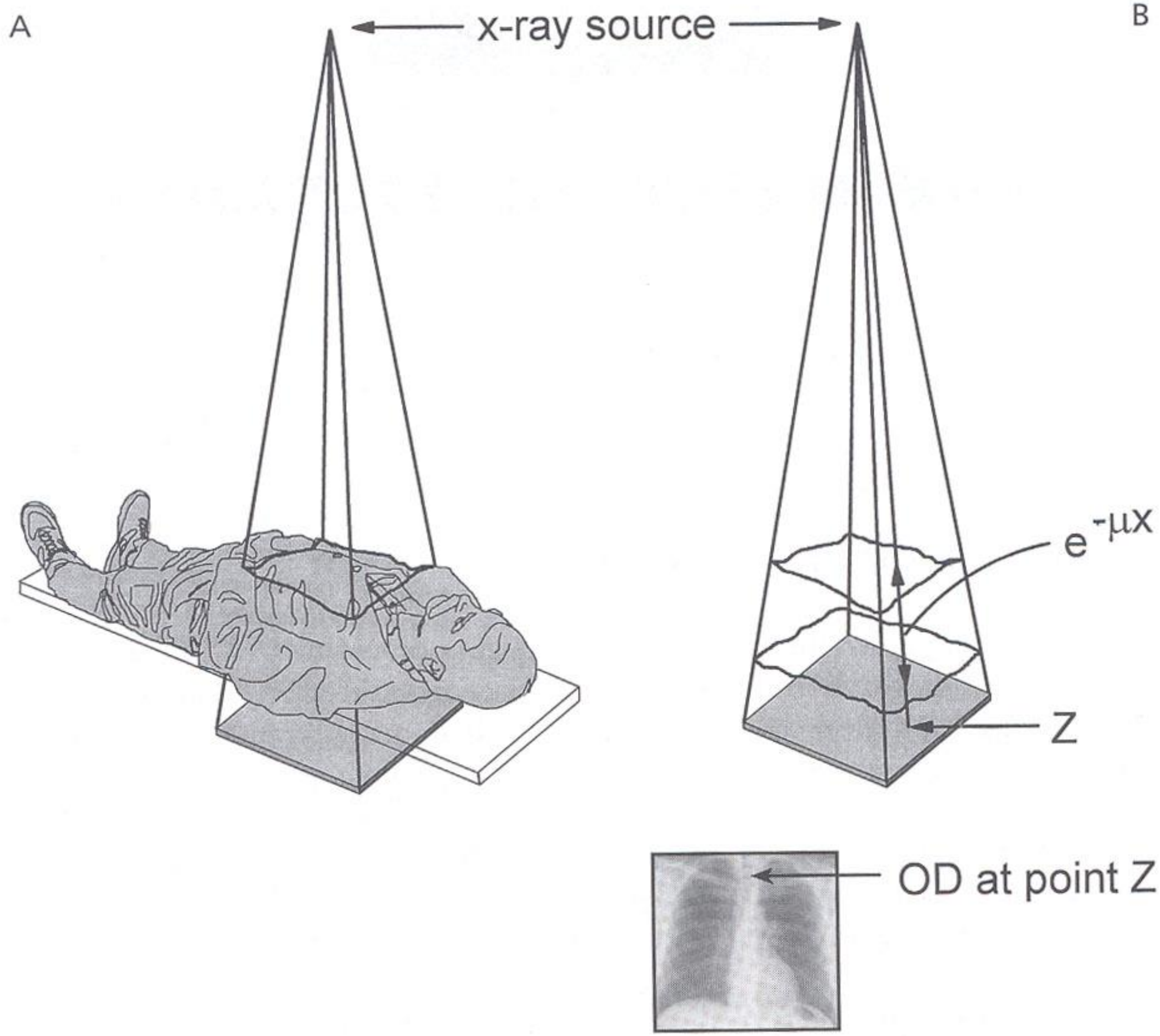
Primary Technique Factors

- mA – tube current
- s – duration of exposure
- mAs – controls total number of x-rays producing image
- kVp – maximum electron kinetic energy
- SSD – source to skin distance
- SID – source to image receptor distance

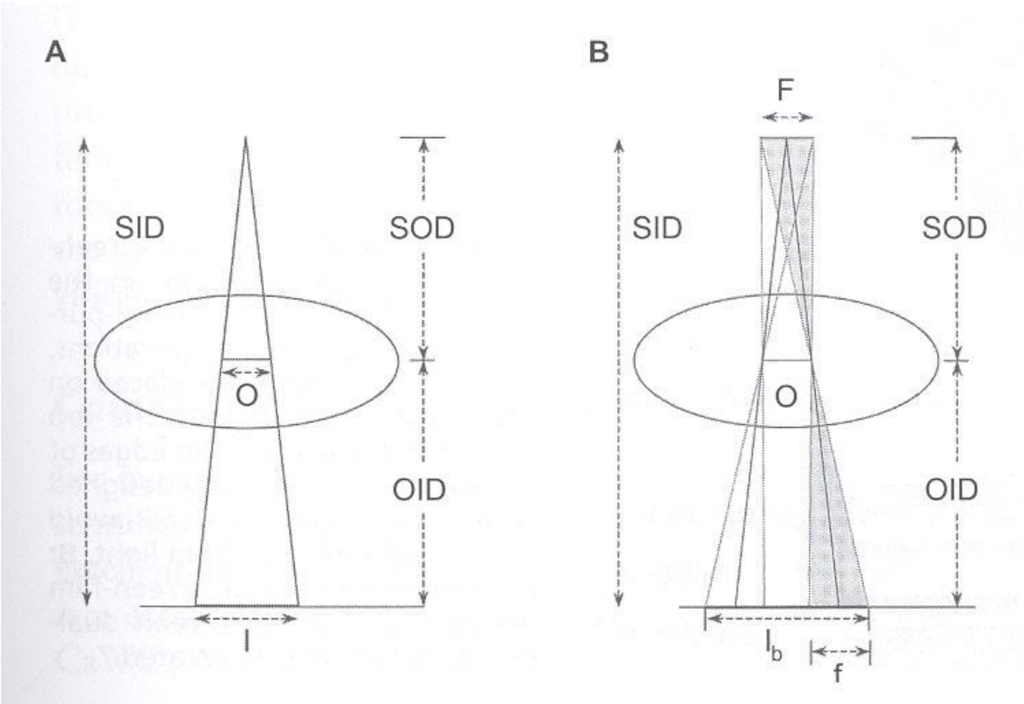
Heat Unit

- The Heat Unit was defined for Single Phase Equipment.
  - HU = kVp * Time (sec) * Tube Current (mA)
  - For Constant Potential Generators the average voltage is higher so to get “apples to apples” we apply a fudge factor of 1.4.
    - HU = kVp * 1.4 * sec * mA

Projection Radiography



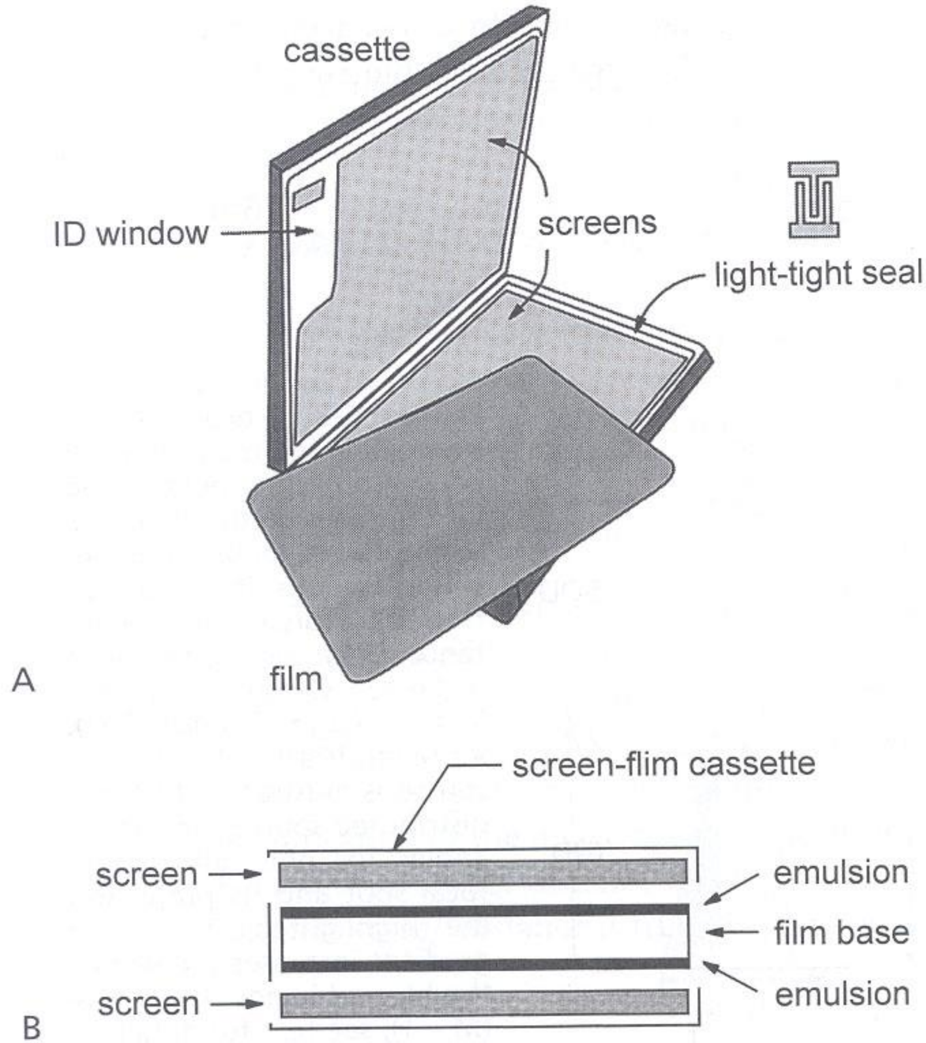
Geometric Principles



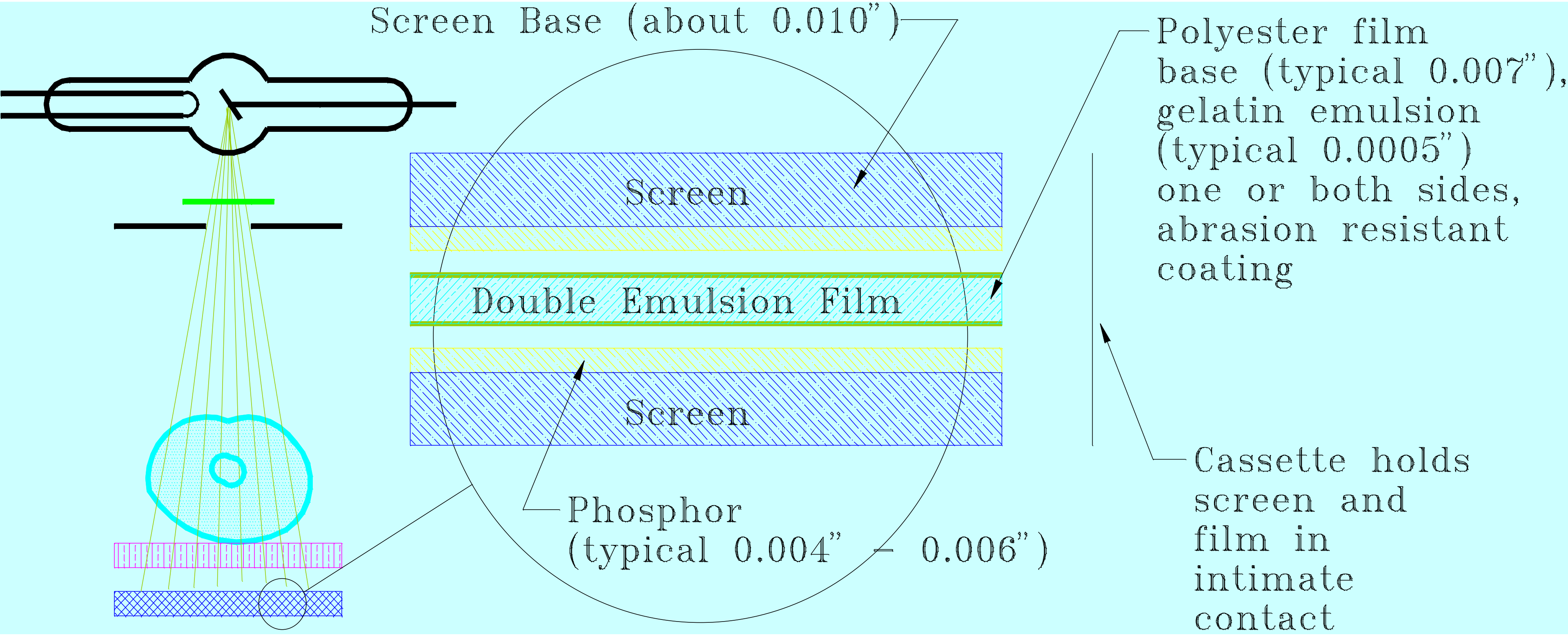
$$M = \frac{I}{O} = \frac{SID}{SOD}$$

$$\frac{f}{F} = \frac{SID}{SOD} - 1 = M - 1$$

Screen-Film Detector System



Typical Screen/Film Combination



X-ray Intensifying Screens

Screens convert x-rays to visible light, which darkens the film more than direct x-ray interaction in the emulsion.

The active mechanism in the screen is x-ray induced fluorescence.

Fluorescence is the prompt (10^{-8} s or quicker) emission of light from an excited atom, molecule, or crystal.

Phosphorescence is the delayed emission

($>10^{-8}$ s after excitation) of light.

Why We Use Screens

To increase the amount of film darkening obtained from a given exposure (increases detection efficiency)

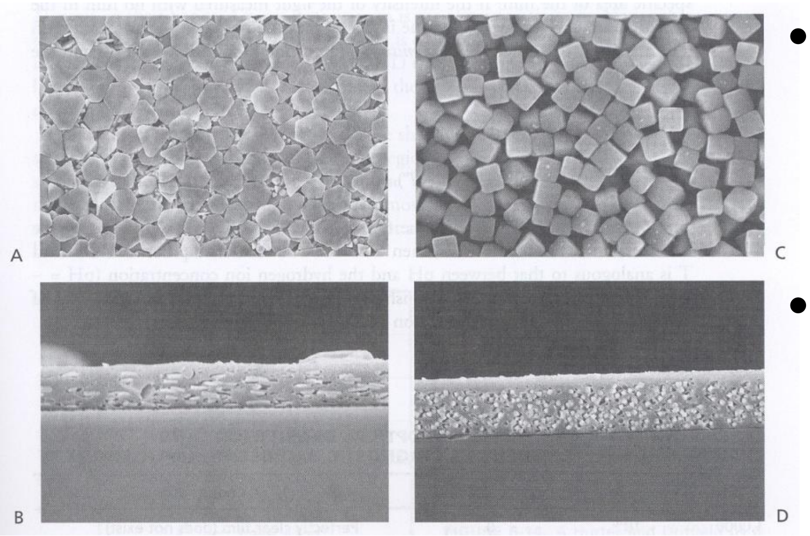
RESULTS:

1) Reduces patient exposure
2) Smaller x-ray generators required
3) Shorter exposure for fixed mA
4) Changes image characteristics

   As, decreases image spatial resolution

Film

Modern Older

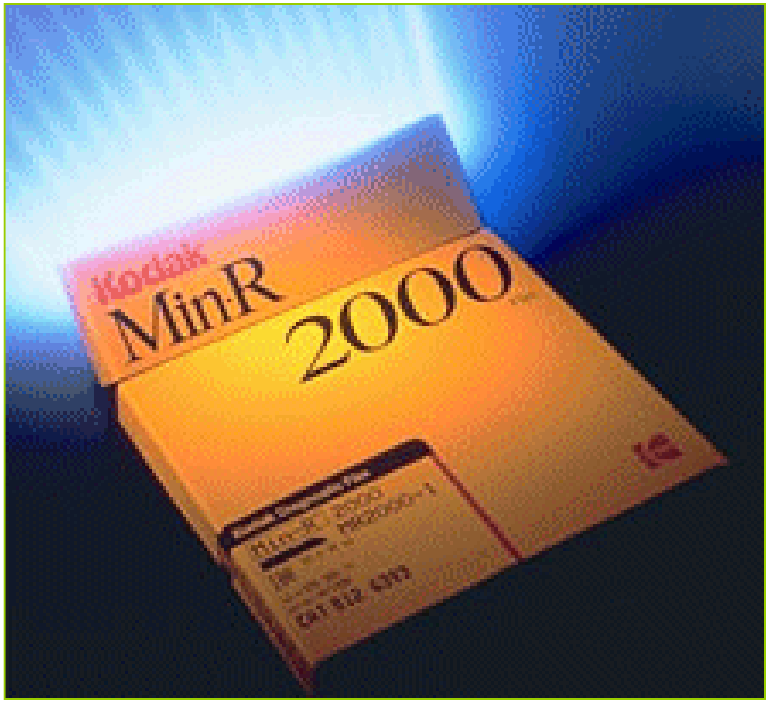


Radiographic films

- A film consists of film emulsion coated onto a type of plastic, mylar.
- The grains of silver halide with a gelatin base comprise the film emulsion.
- The composition, size and size distribution of the grains determine the film speed, film contrast.
- Bigger size grains requires less exposure to get certain optical density. Uniform grain size makes higher film contrast.

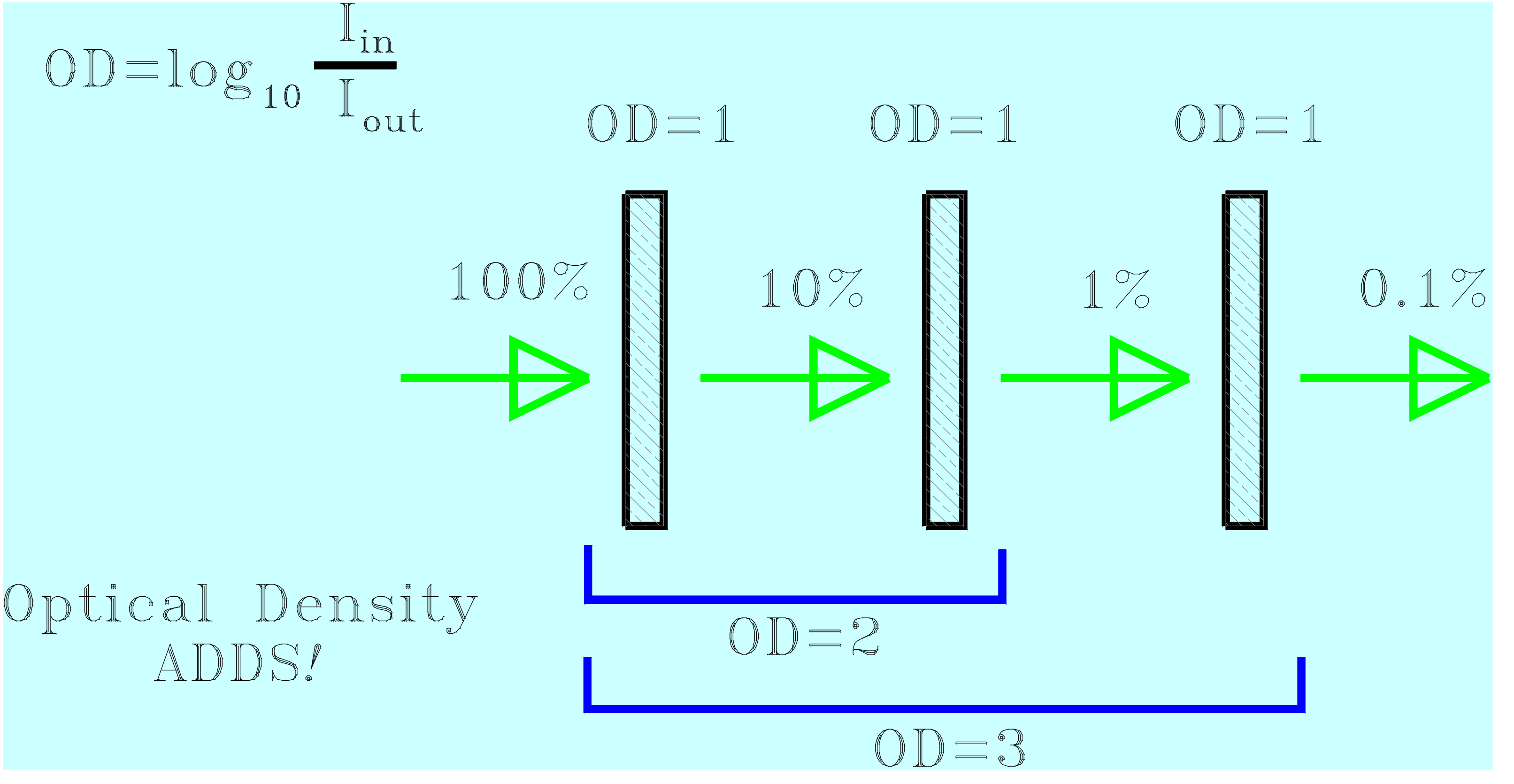
Type of Films

- Blue sensitive film (non-orthochromatic) sensitivity runs from UV to between 4000 and 5000 Å
- Orthochromatic film -- sensitivity extended to green for use with green emitting rare earth screens

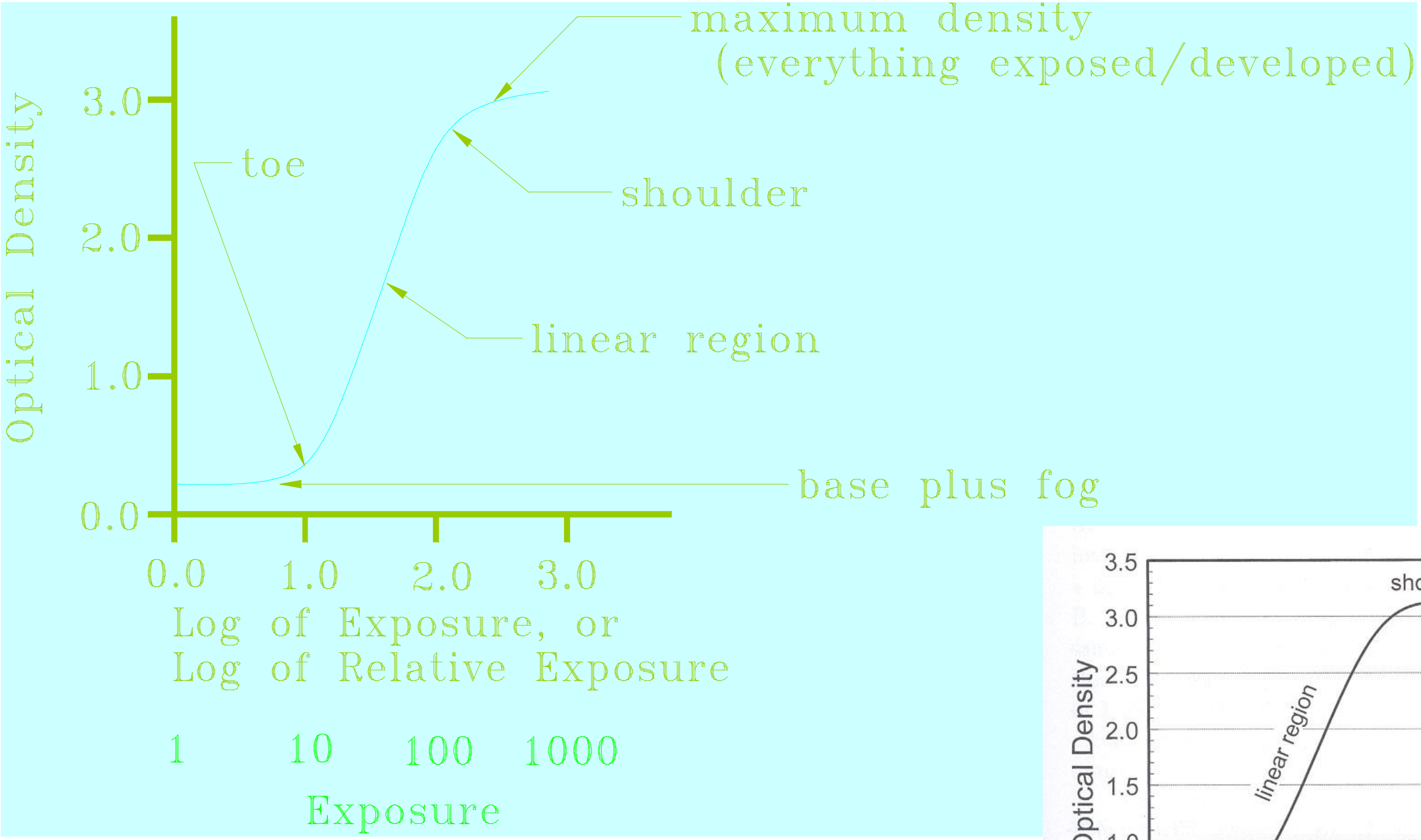


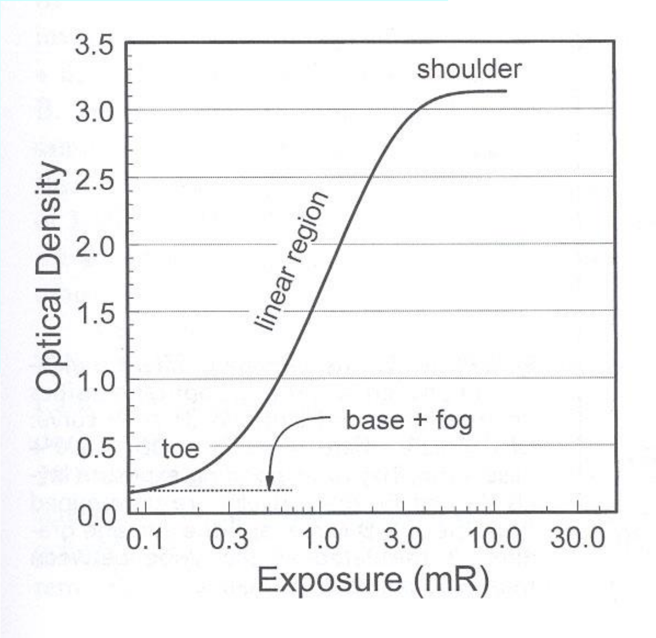
- Panchromatic film -- sensitivity extended to red

Optical Density: A Measure of Film Response

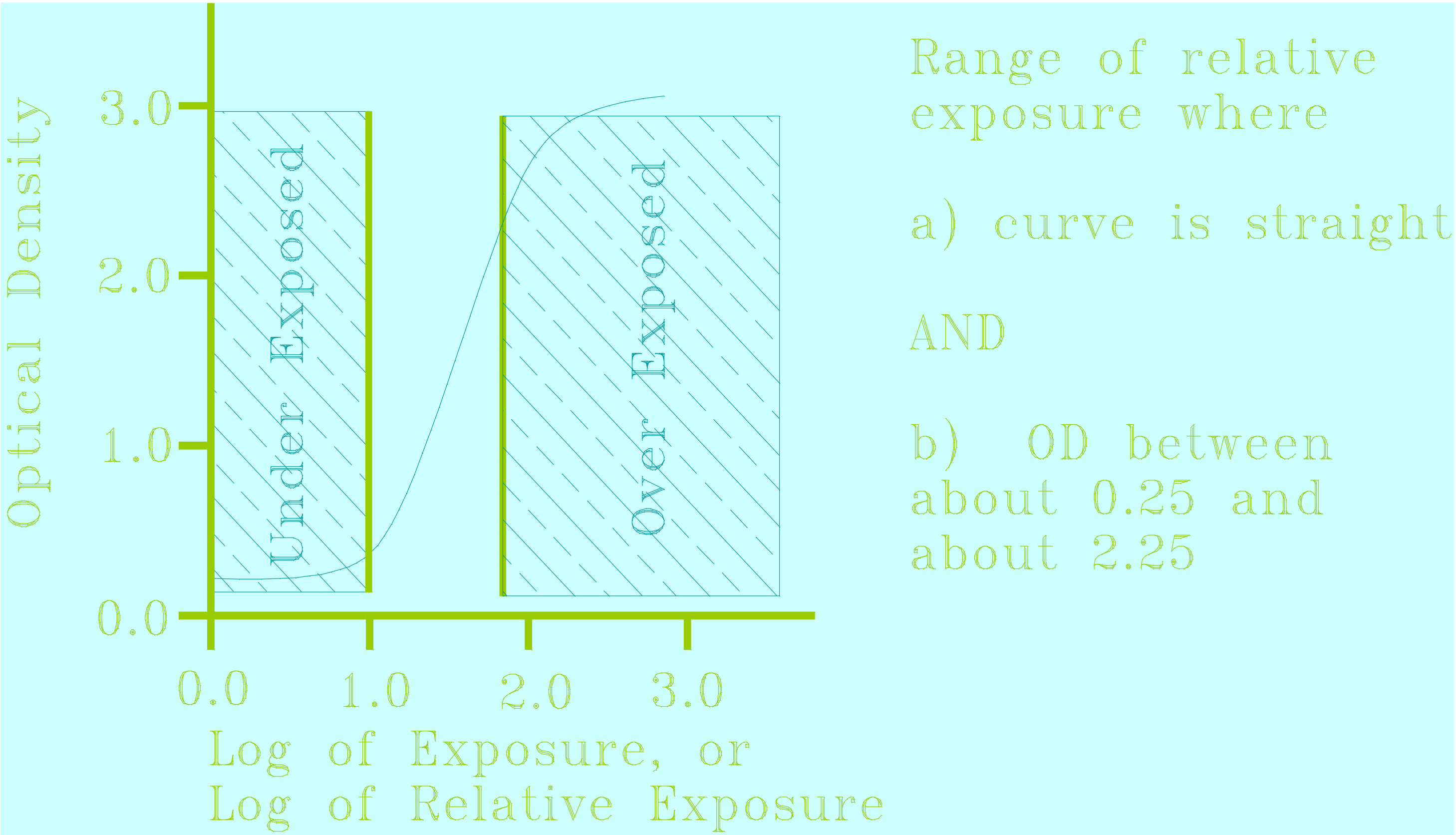


What's What on the H and D





Latitude: The Range of OK Exposure



A good image means the region which is important to the diagnosis of the diseases should be in the range of OK exposure for the film.

Current Technology in Digital Imaging

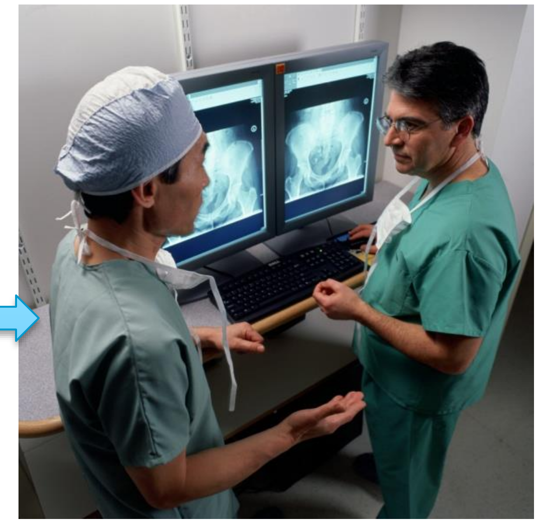
Why do we go for “Digital”?

- Limitations of film-screen characteristics
- Limitation of physical storage space in hospital
- Need for electronic type imaging (PACS)
- Workflow convenience
- Image Quality
- Dose efficiency

Benefit of PACS: Real-time clinical consultation

Radiology

PACS



Clinician

Digital Radiography

From screen/film to DR

- Images from analog to digital…
  - Image viewing, transferring and archiving
  - Digital image processing
- Factors affecting techniques:
  - Dynamic range
  - Detector energy response
  - Detector efficiency

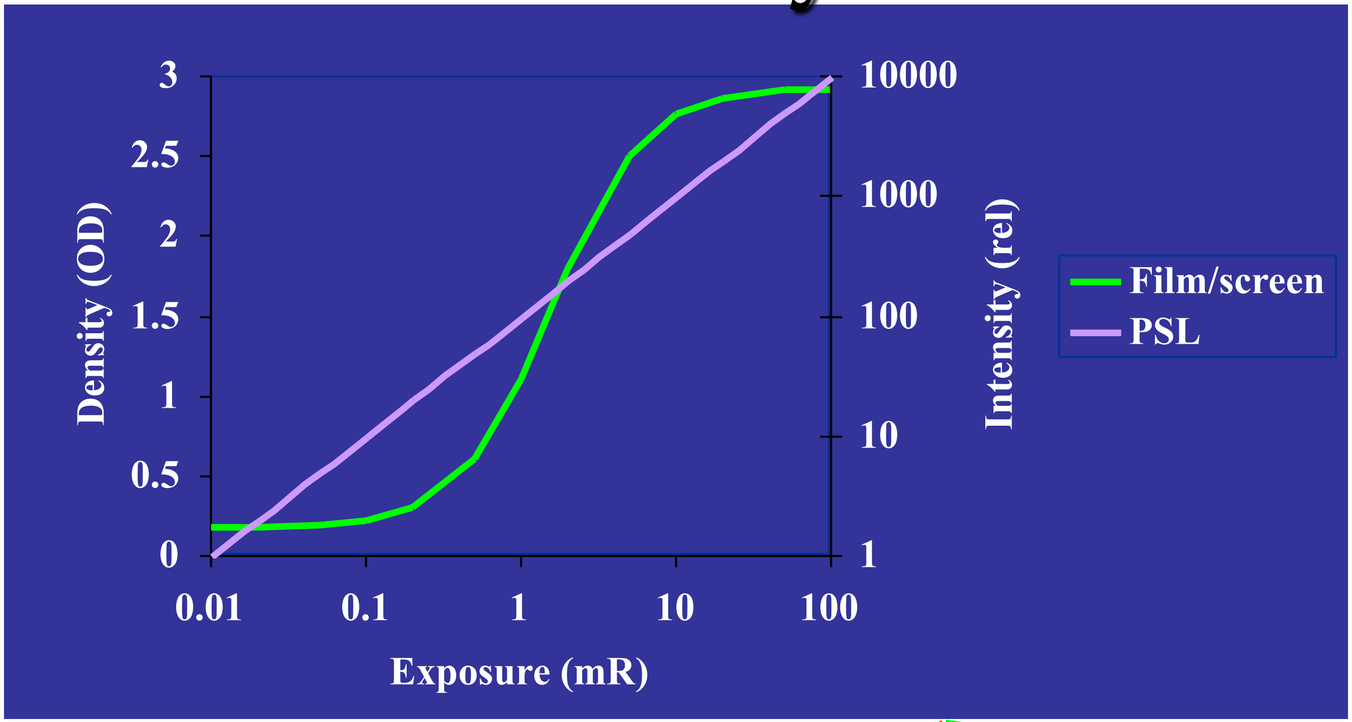
Interesting Points

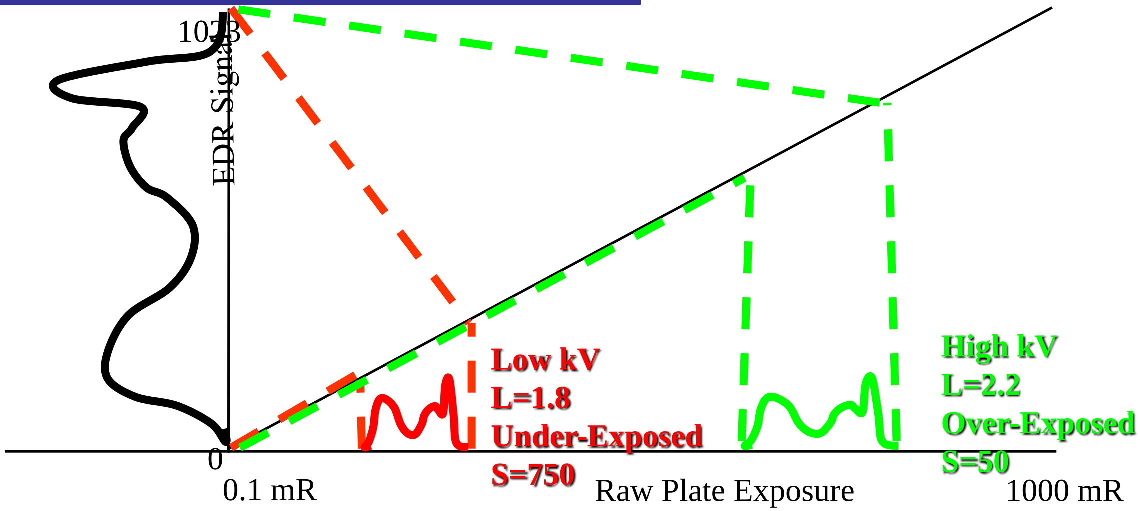
- Screen-film characteristics are different from digital radiography
- Among types of digital radiography, they are different
- Technical settings by Technologist are different for conventional vs. digital imaging
- QC procedures/Film Reporting for each type of these modalities are different

DR: Acquisition Technology

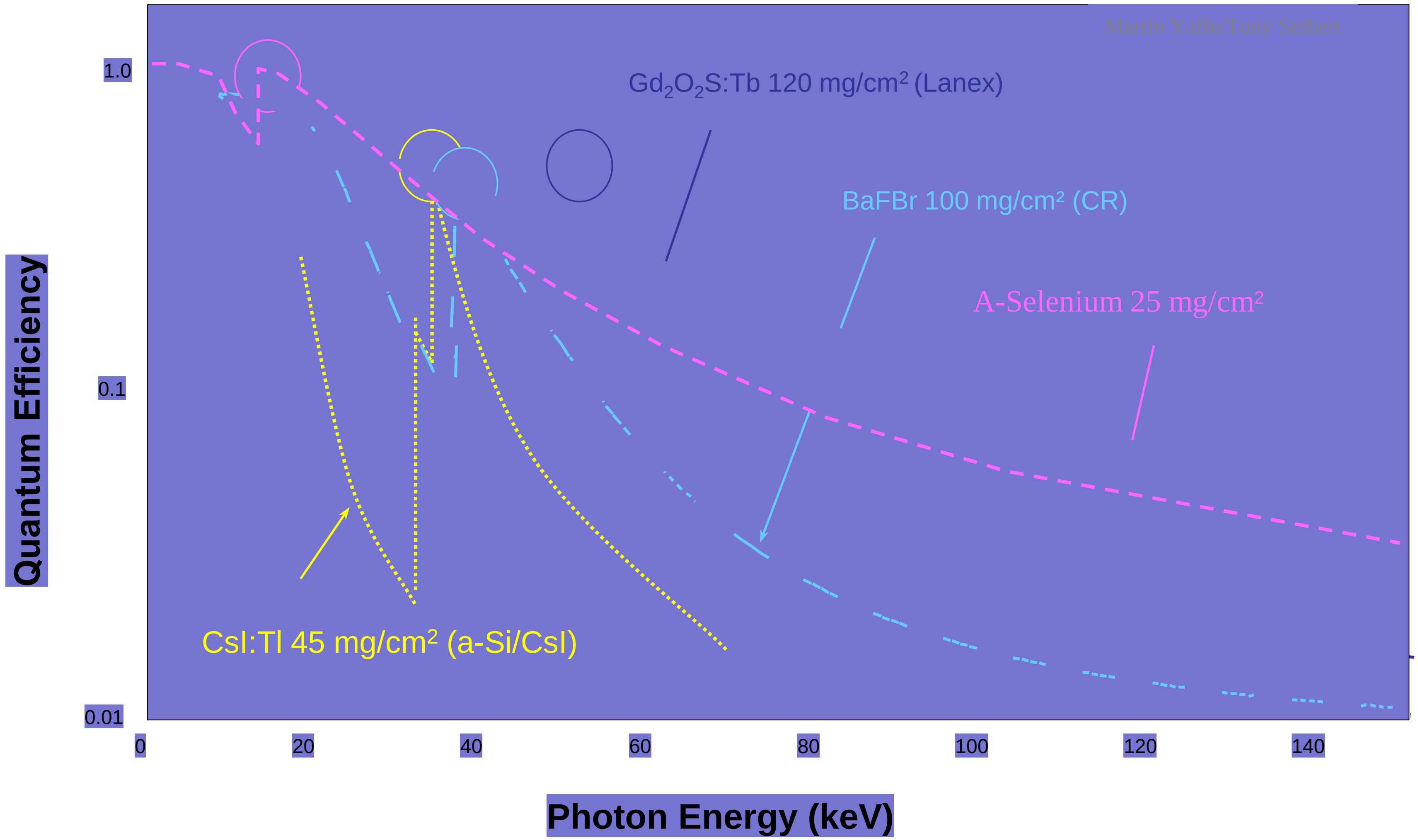
- Photostimulable Phosphors ("CR" or "PSP")
  - Photostimulable phosphor plates
- Flat-panel Detectors
  - Direct DR (DDR): Amorphous Selenium Detector – matrix of transistors, without photon conversion layer
  - Indirect DR (IDR): Amorphous Silicon TFT or CCD with CsI conversion layer

Wide Dynamic Range





Detector energy sensitivity



Wide dynamic range

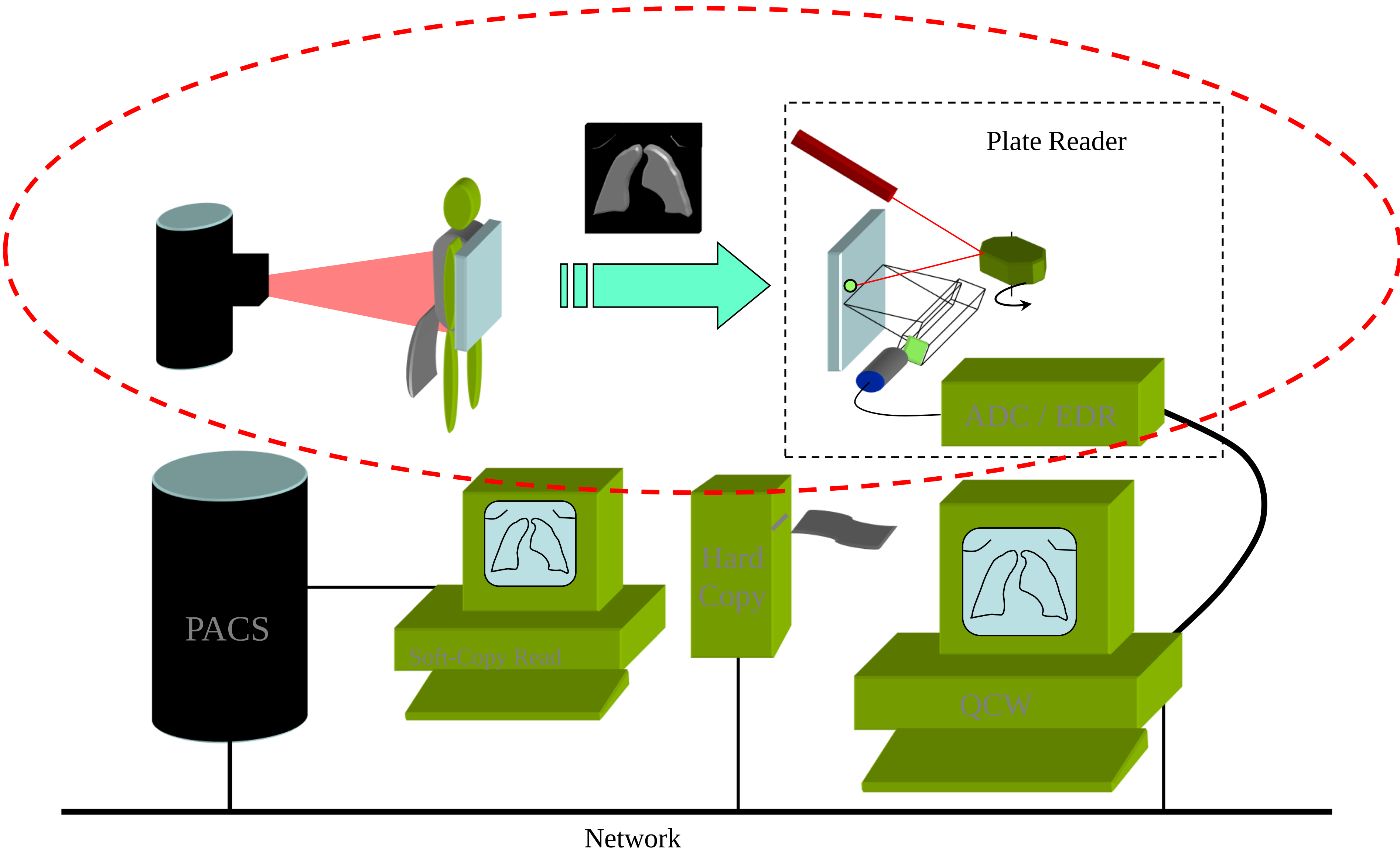
- **Under- and Over- exposure**
  - **Fewer photons – More noise**
    - **Obscures low-contrast details**
  - **More photons = More signal strength (signal-to-noise ratio improves)**
    - **Beautiful images!**
    - ***High patient dose!***
- **Wide dynamic range can lead to higher patient dose**

Detector energy response and efficiency

- **Optimal beam quality could be different**
  - **kVp**
  - **Filtration**
  - **Also consider contrast and patient dose**

- **Optimal beam quantity (mAs) could be different**
  - **AEC calibration or manual techniques**
  - **Patient dose (kV dependant)**

Workflow: Computed Radiography (CR)



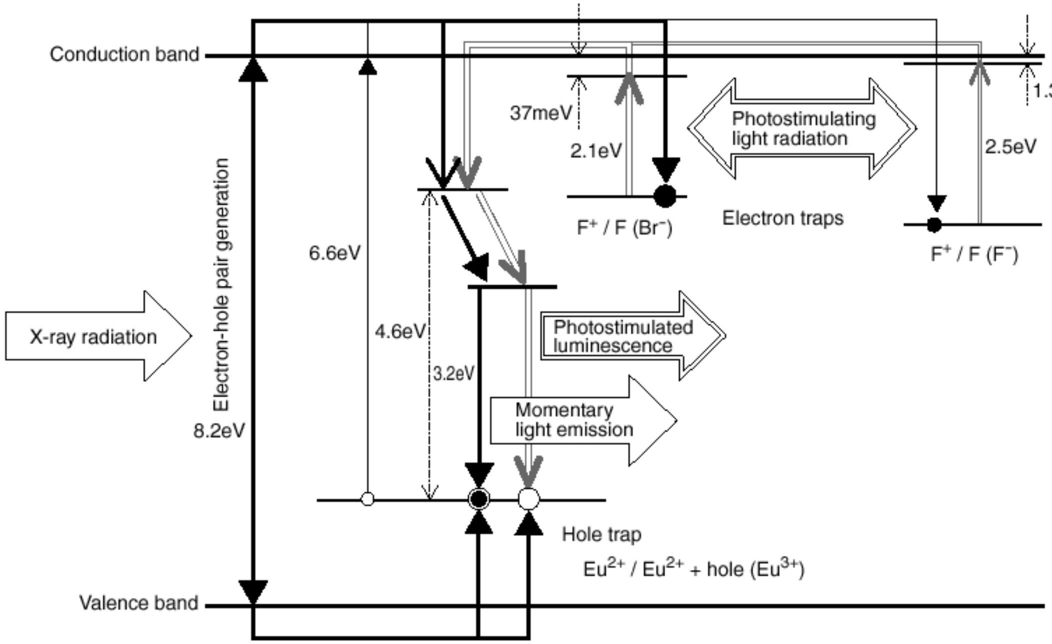
CR is based on the physical process of photostimulable luminescence (PSL)

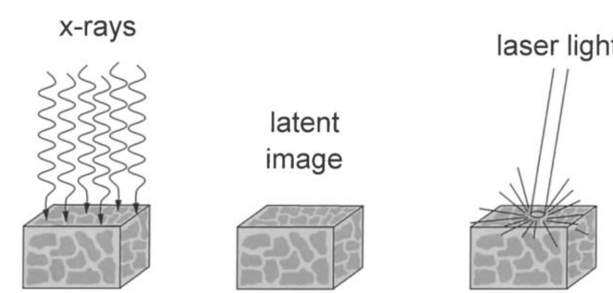
- **X-rays contribute energy to the electrons by the photoelectric effect**
- **Electrons can give up energy (violet light)…**
  - ***by emitting light immediately (fluorescence)***
  - ***by emitting light slowly (phosphorescence)***
- **Some electrons can retain (*store*) their energy**
  - ***crystal defects can "trap" excited electrons***
  - ***electrons can escape the traps when exposed to the proper wavelength (red) light (photo-stimulated luminescence)***
  - ***electrons can also escape by thermal mechanisms***

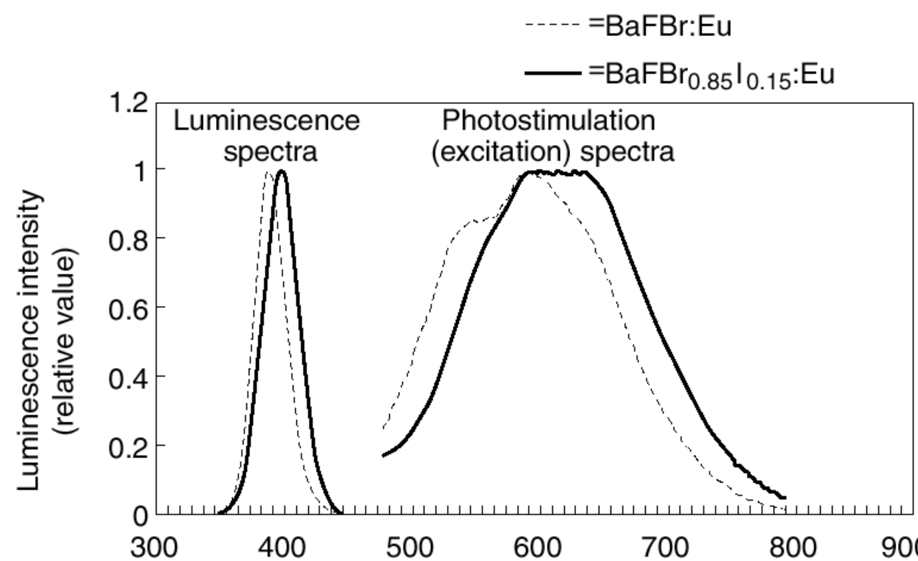
Materials that exhibit PSL are called photostimulable phosphors (PSP)

- **PSPs currently in use for CR are crystals of alkaline earth and halides "doped" with Eu**
  - **$BaFBr:Eu^{+2}$ =>** ***Fuji ST – STIIIA, Kodak?***
  - **$BaFBr_{0.85}I_{0.15}:Eu^{+2}$ =>** ***Fuji STV - STVI***
  - **$Ba_{0.86}Sr_{0.14}F_{1.10}Br_{0.84}I_{0.06}:Eu^{+2}$ =>** ***Agfa***
  - **$BaFBr_{0.8}I_{0.2}:Eu^{+2}$ =>** ***Konica (early)***
  - **$BaFI:Eu^{+2}$ =>** ***Konica (current)***
  - **RuBr:Tl =>** ***Konica (ancient)***

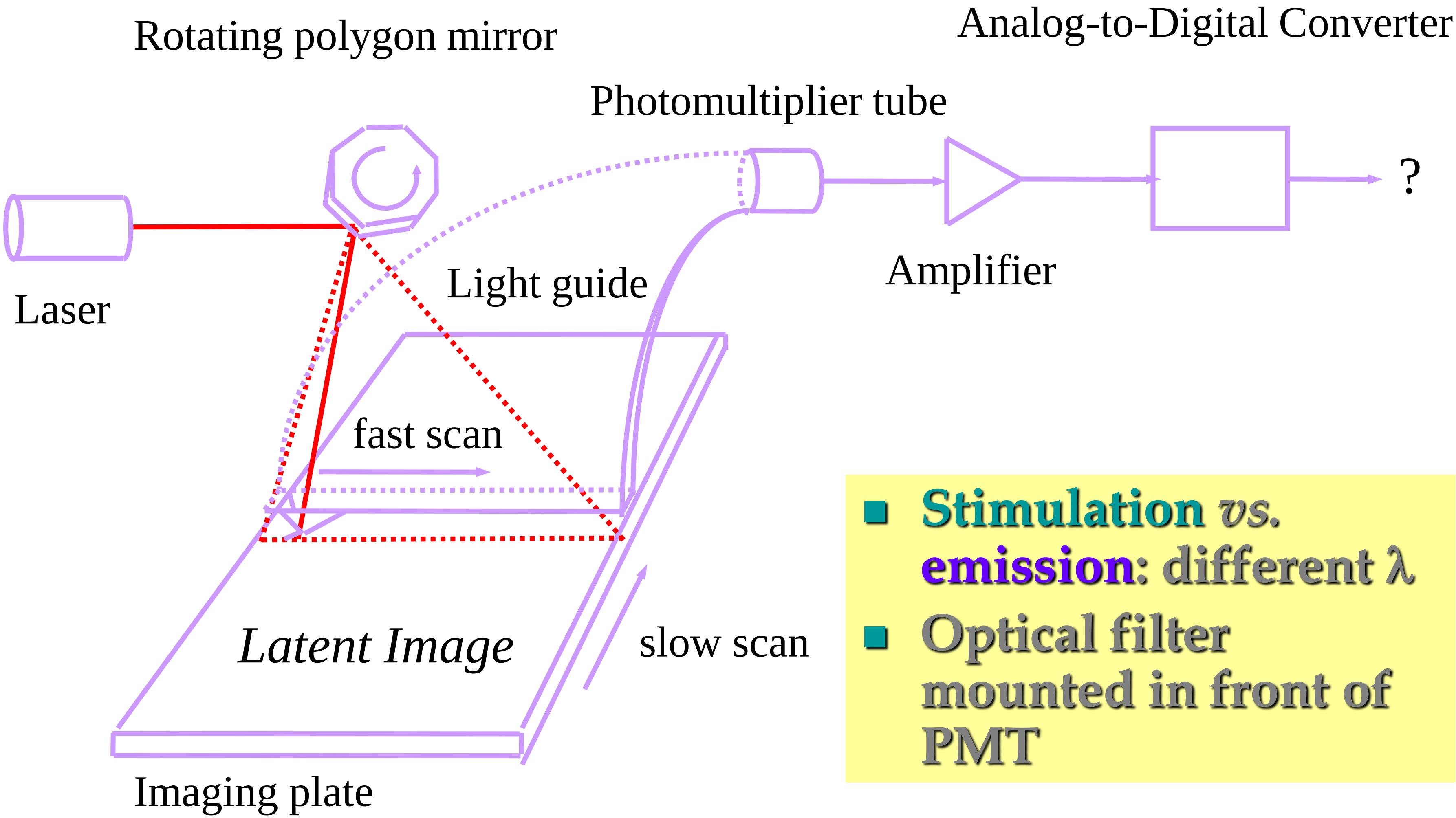
Photostimulable Luminescence





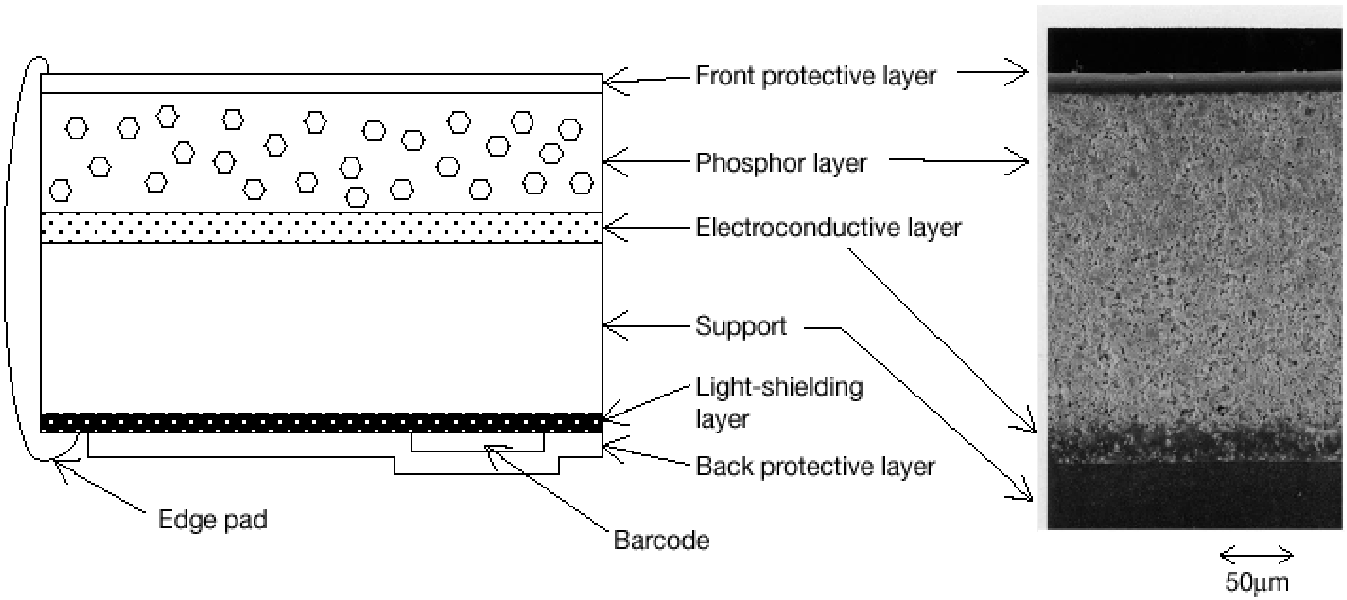


Development and Digitization of the CR latent image

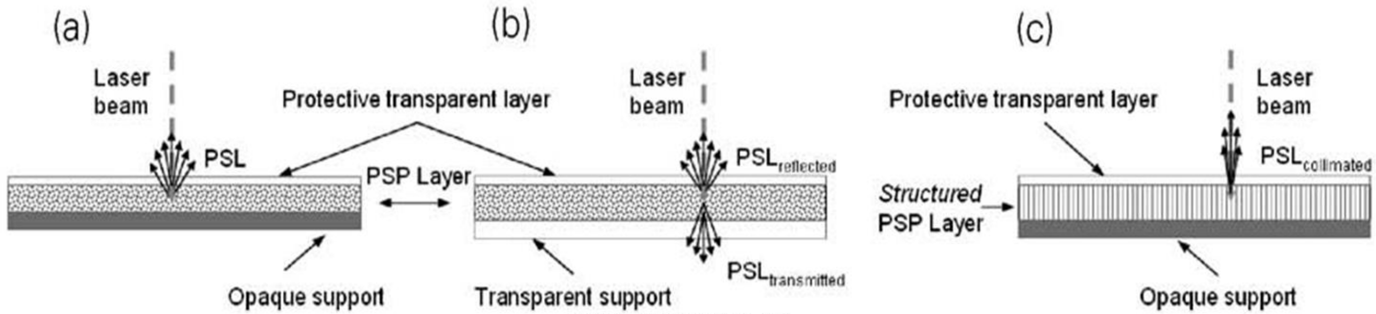


- **Stimulation** ***vs.*** **emission: different λ**
- **Optical filter mounted in front of PMT**

Plate Structure



Available computed radiography technology



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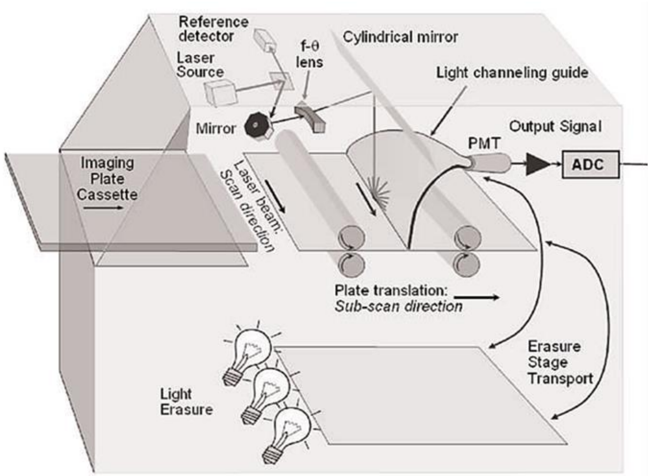
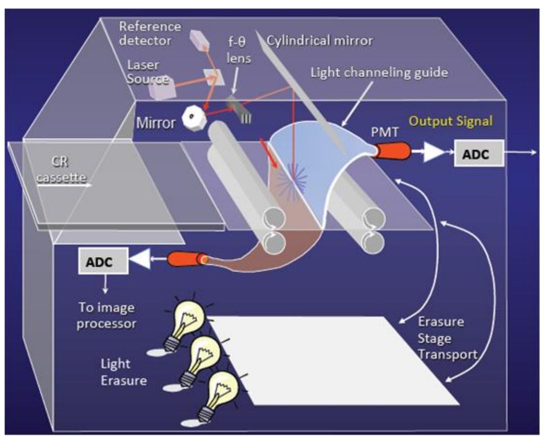
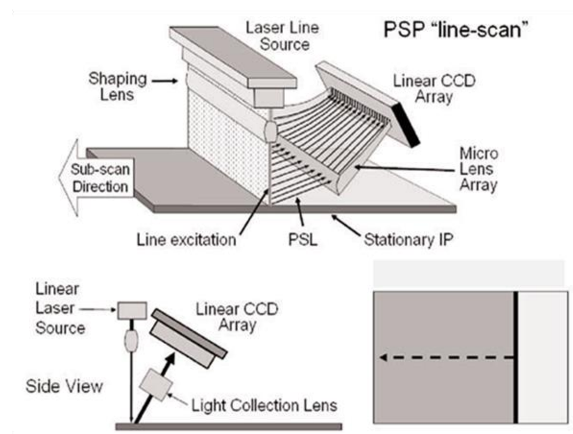


Point-scan Laser Readout



Dual-side Laser Readout



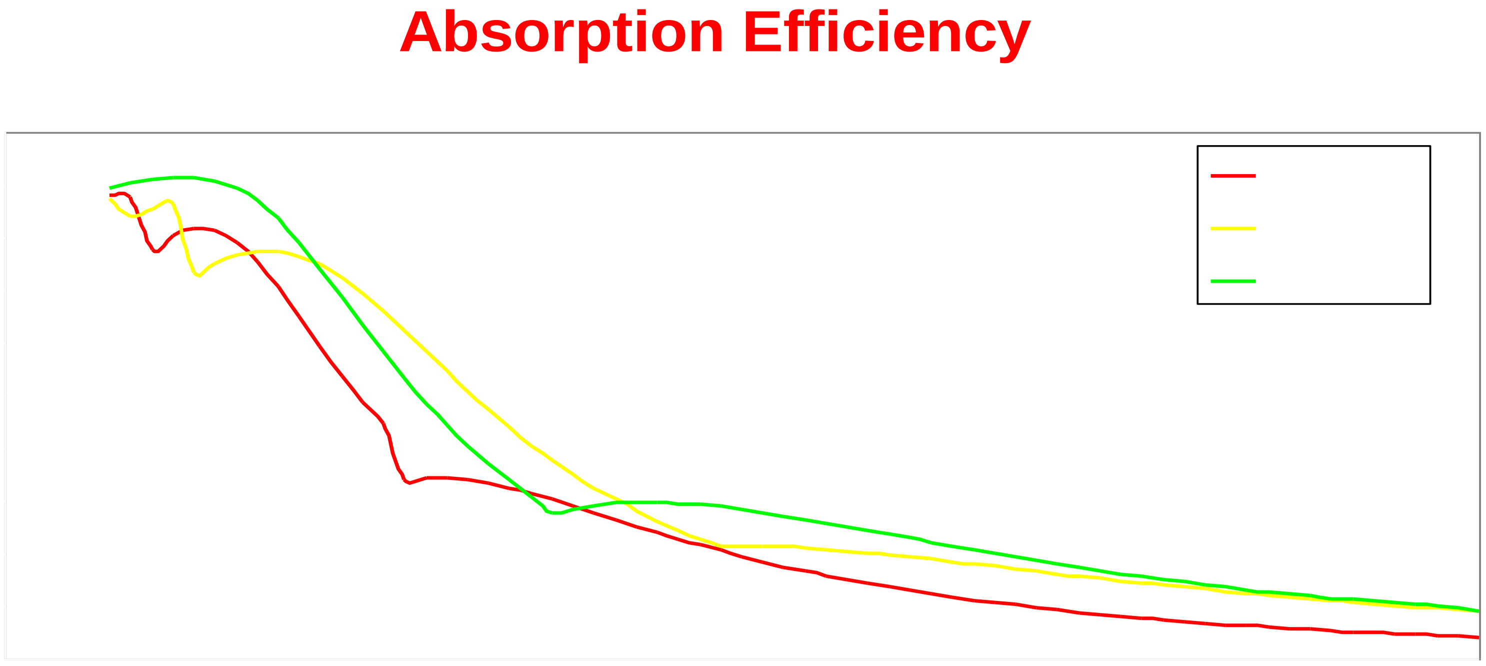
Line scan Laser Readout

J. Anthony Seibert,Ph.D, UCDAVIS University of California Davis Health System Sacramento, California USA

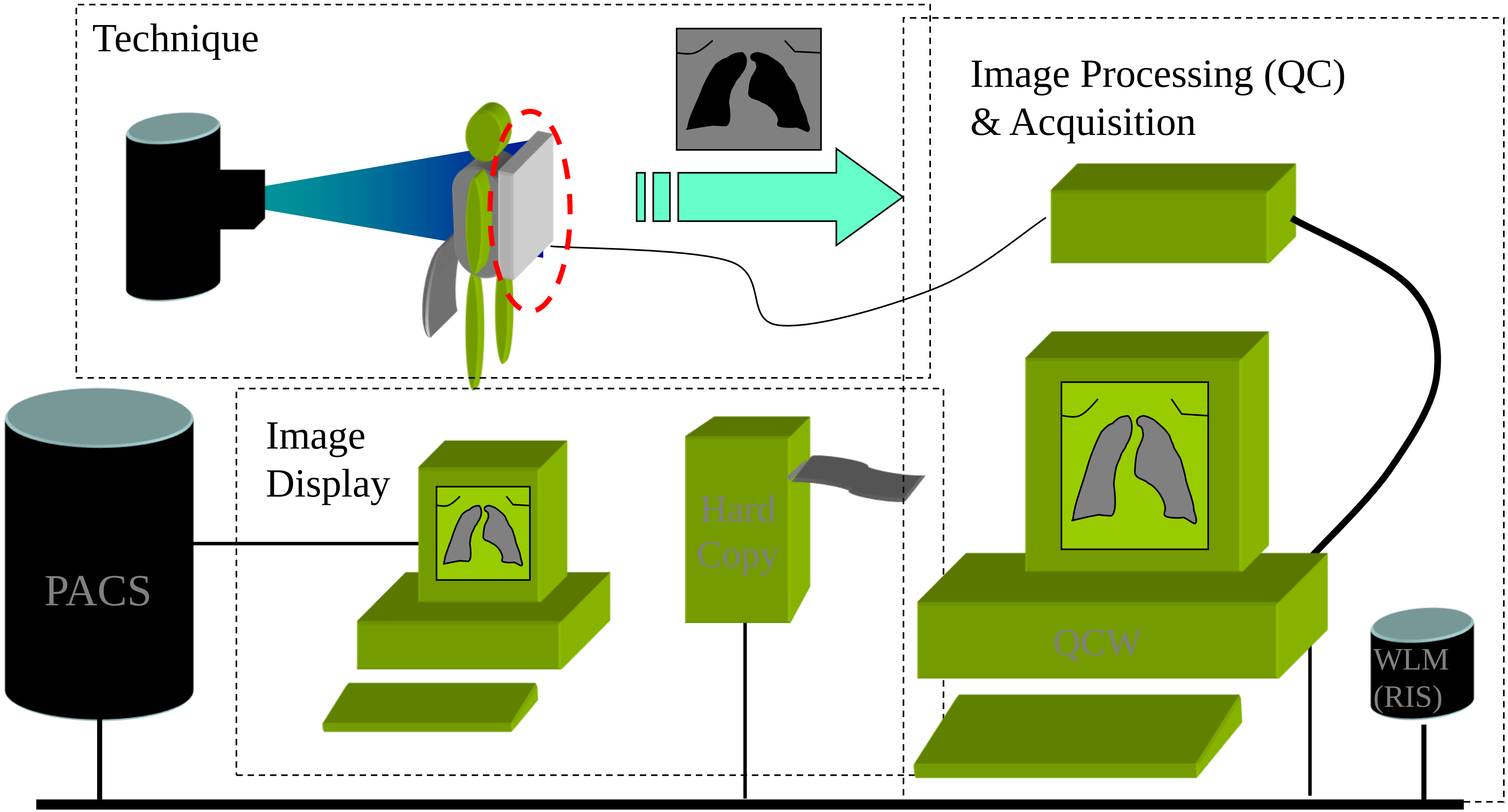
CR Reader/Digitizer

Various capabilities, sizes, throughput

X-ray absorption by PSP is different from most intensification screens



Workflow: Flat Panel System



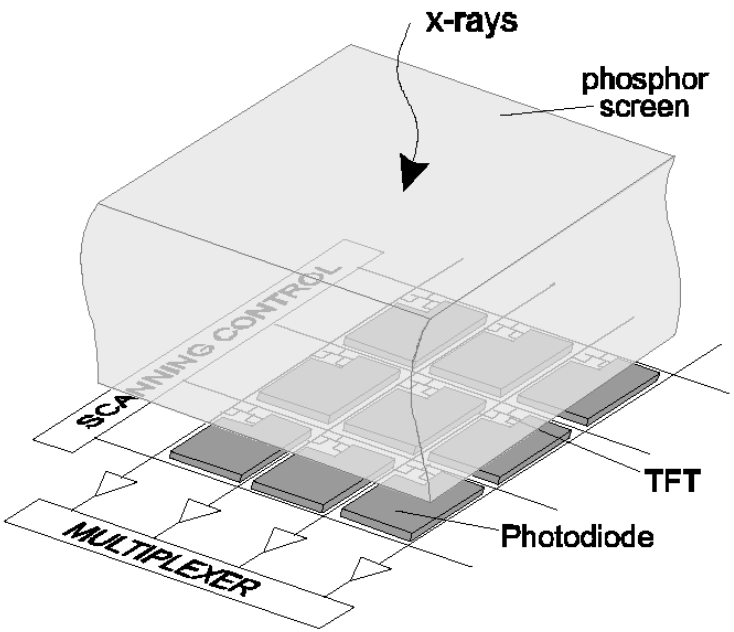
Flat Panel System

- **Direct x-ray detection (DDR)**
  - **Photoconductor (a-Se) on top of TFT array**
  - **X-rays interact with photon sensors directly**
  - **After exposure, e's are generated and migrated through the Se layer (+) to the TFT layer for readout**

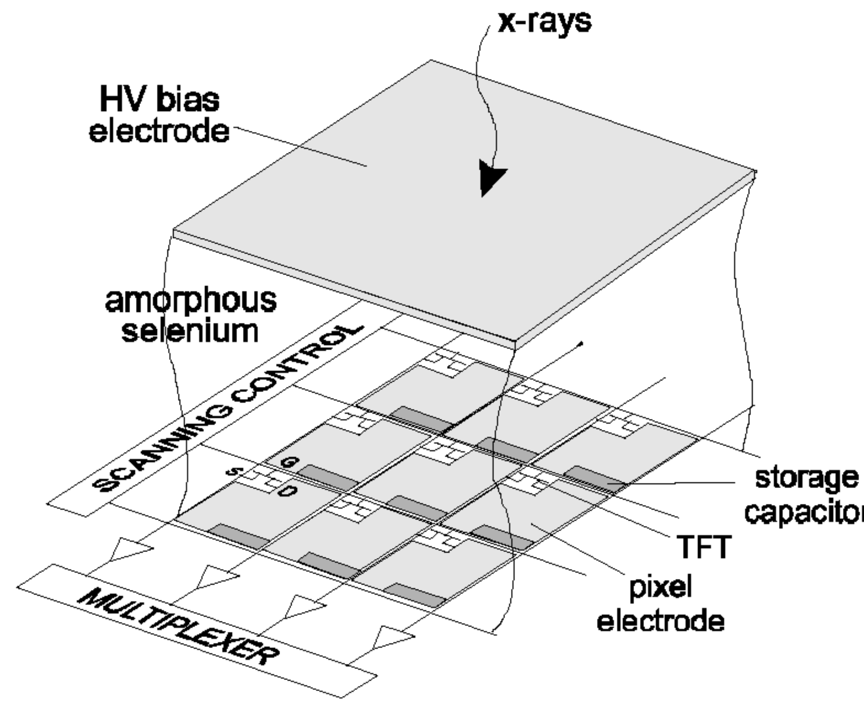
- **Indirect x-ray detection (IDR)**
  - **X-rays interact with an intensifying screen, and secondary photons interact with sensors (a-Si TFT or CCD)**
  - **The screen causes more blurring**
  - **CsI is more commonly used to improve spatial resolution.**
  - **GOS is also used as x-ray to light conversion**

Unlike CR, no mechanical readout process is involved (self-reading)

DR Detector Configurations



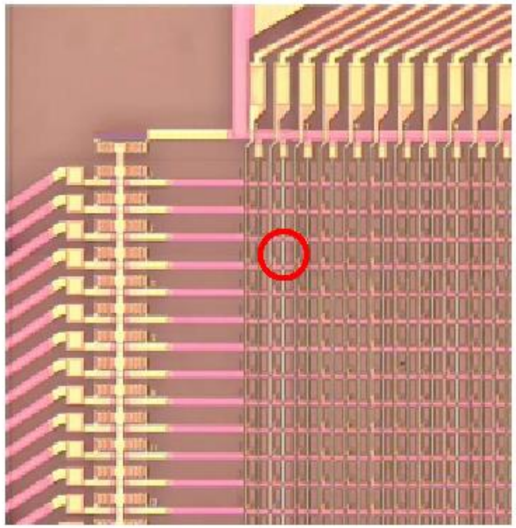
Indirect DR



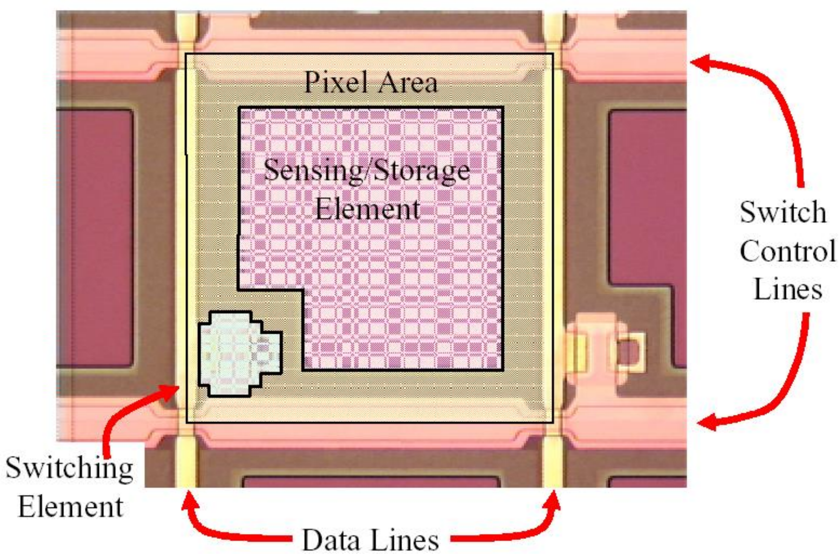
Direct DR

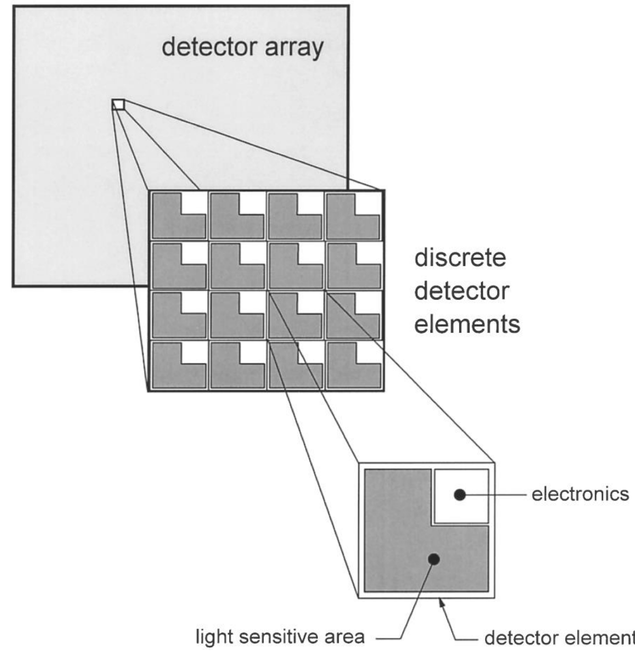
Pixel Construction

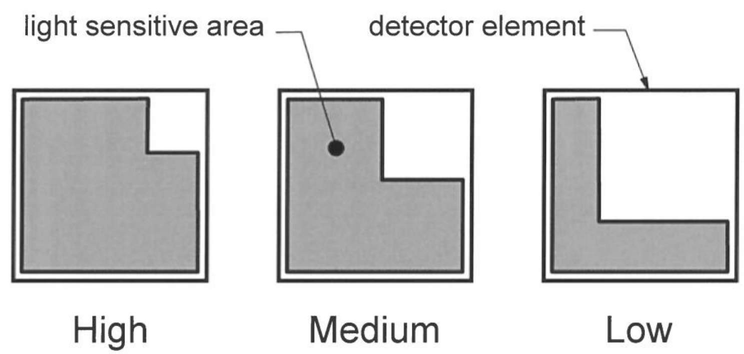
Large Area Substrate



Corner of Substrate

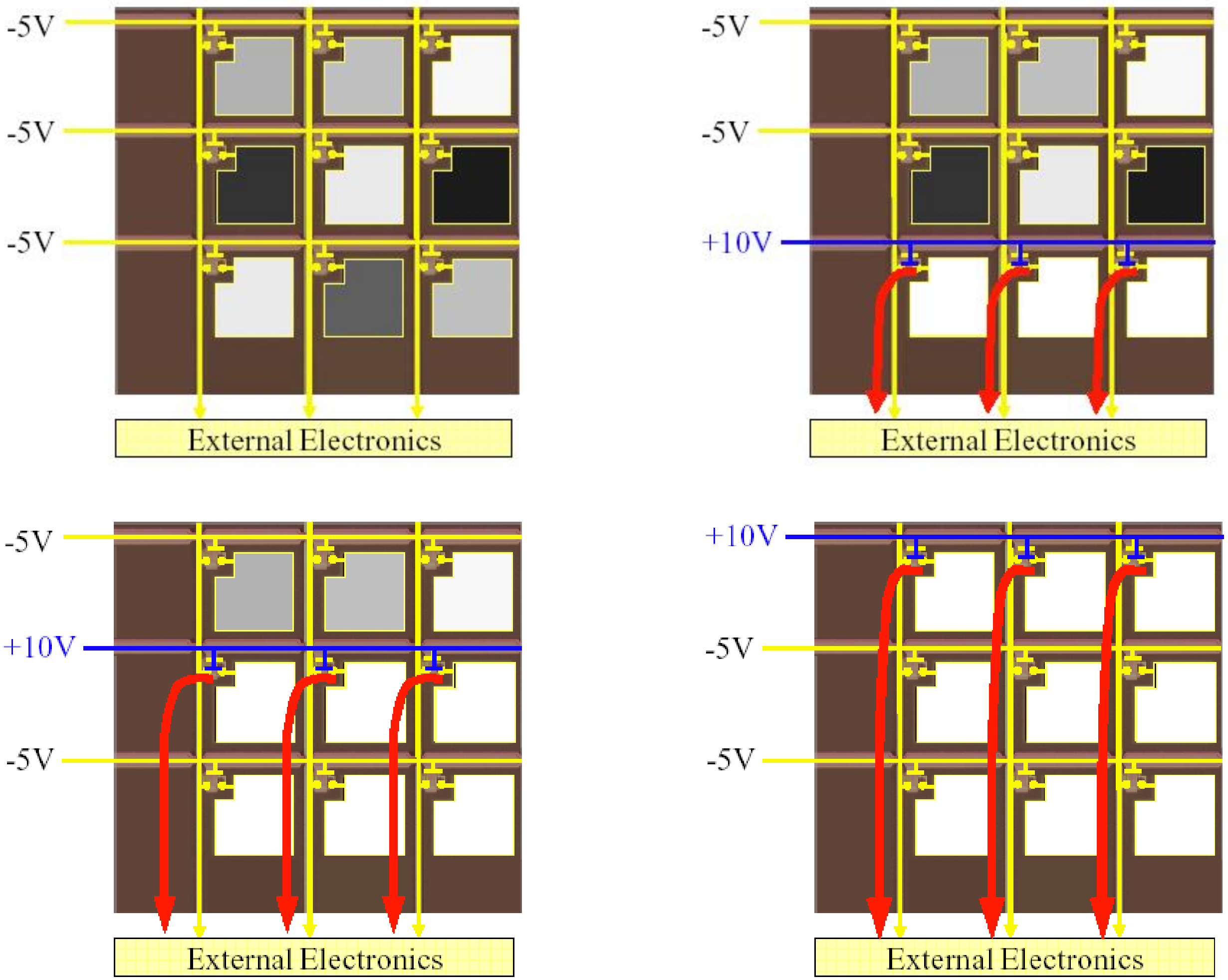






$$\text{Fill Factor} = \frac{\text{light sensitive area}}{\text{area of detector element}}$$

TFT Readout



Gate lines

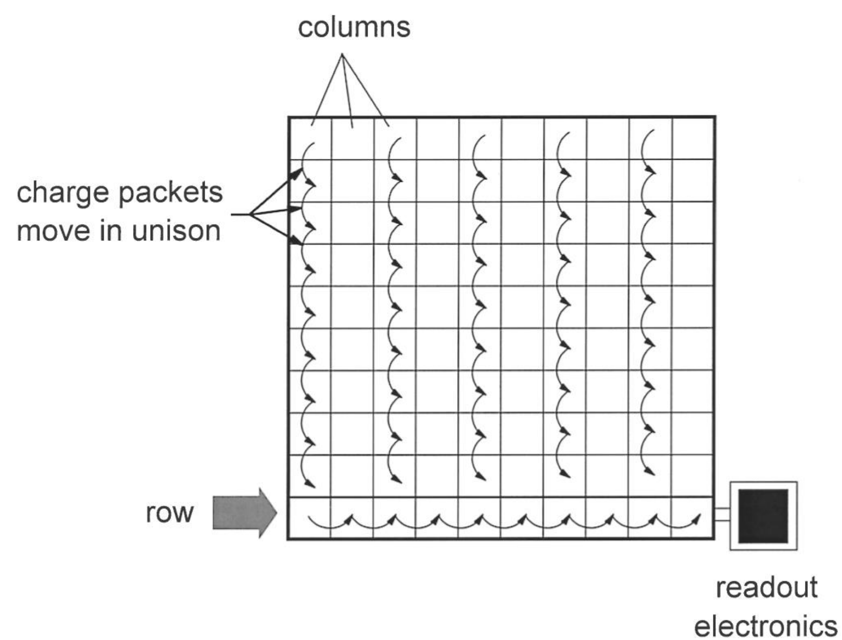
Readout lines

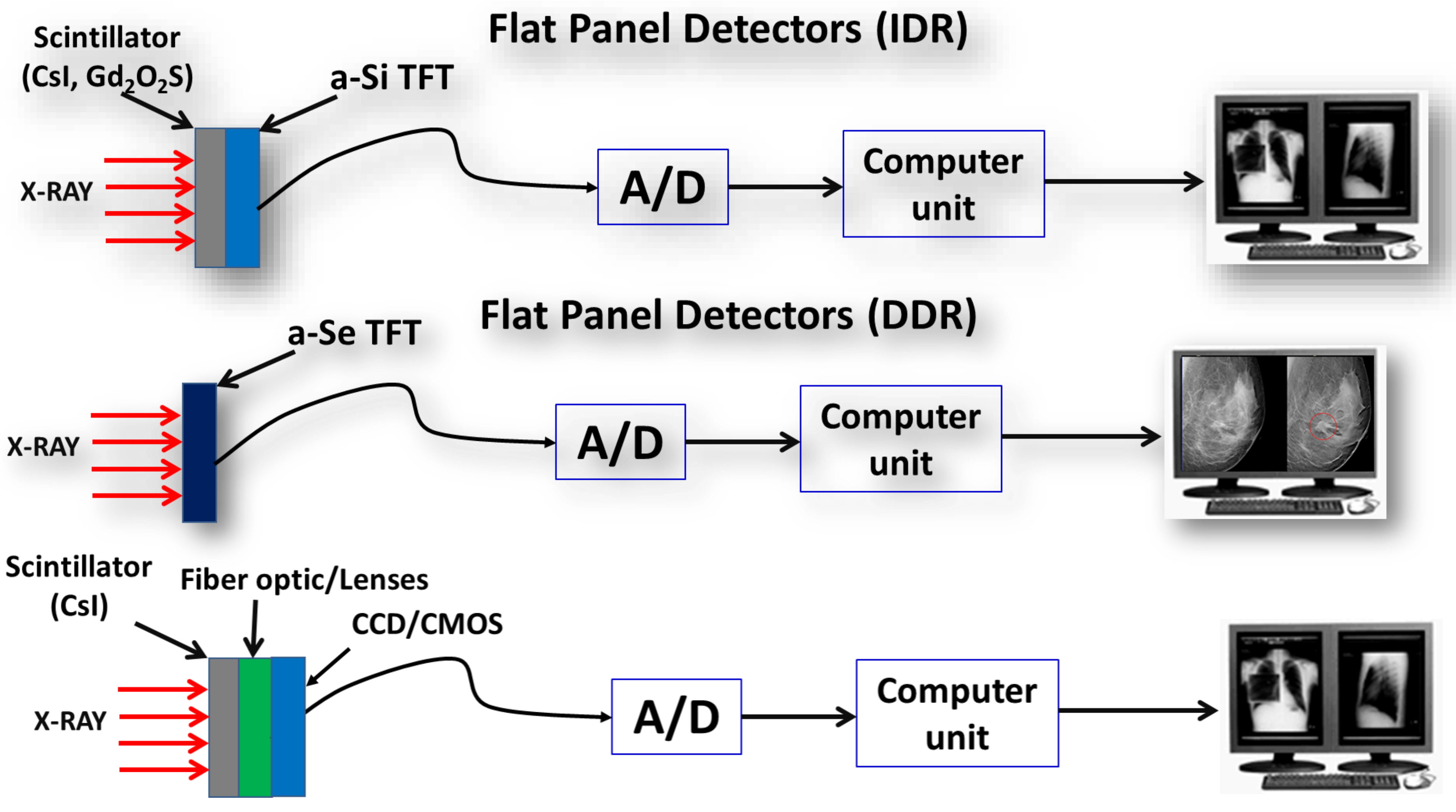
During exposure:
(-) voltage applied to gate lines (charge accumulated)

During readout:
(+) voltage applied to gate lines (so transistor turned on), one gate line at a time.

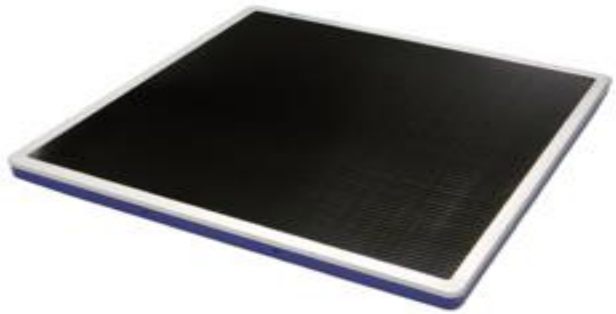
Charged-Coupled Devices

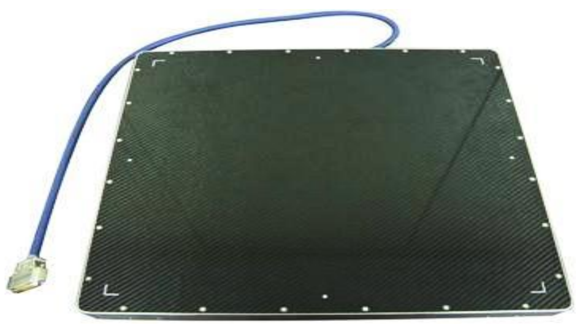
- **Convert visible light to form images**
- **CCD chip = integrated circuit made of silicon, w/ discrete pixel electronics etched into surface**
  - **Ex. 2.5 x 2.5 cm CCD chip may have 1024 x 1024 or 2048 x 2048 pixels**
  - **Electrons are liberated after visible light exposure and kept in each pixel because there are electronic barriers (voltage) on each side during exposure.**
  - **After exposure, charges move down by togging voltages between rows and read out at the last row.**



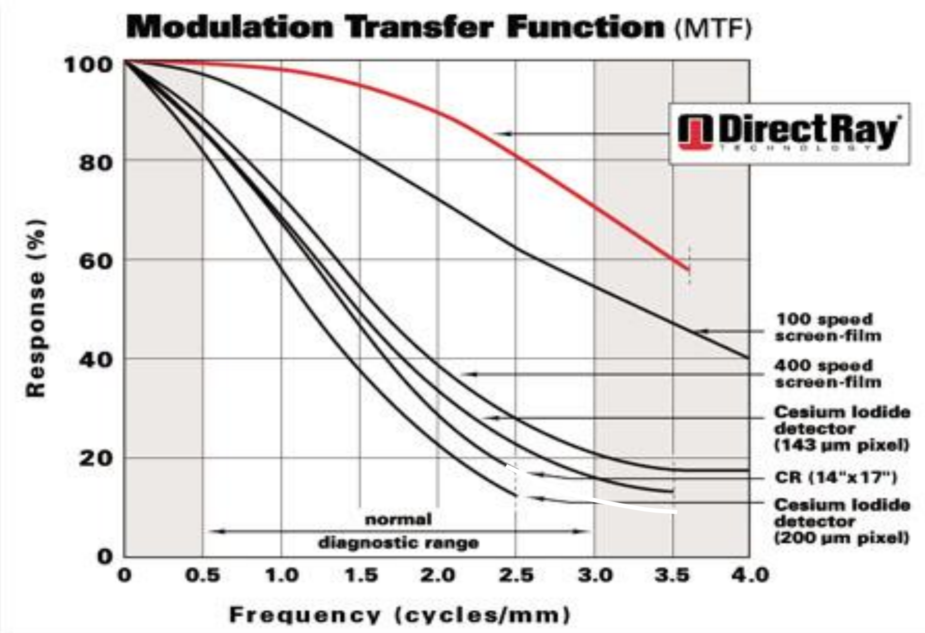

Flat Panel Detectors (IDR)
Scintillator
(CsI, Gd$_2$O$_2$S)
a-Si TFT
X-RAY
A/D
Computer unit
Flat Panel Detectors (DDR)
a-Se TFT
X-RAY
A/D
Computer unit
Scintillator
(CsI)
Fiber optic/Lenses
CCD/CMOS
X-RAY
A/D
Computer unit

Flat Panel Detectors

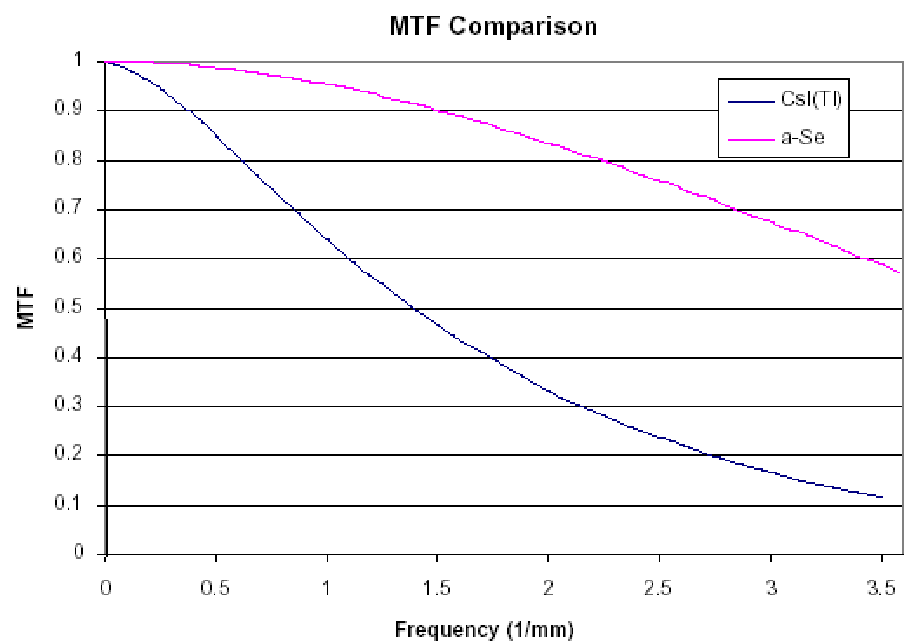




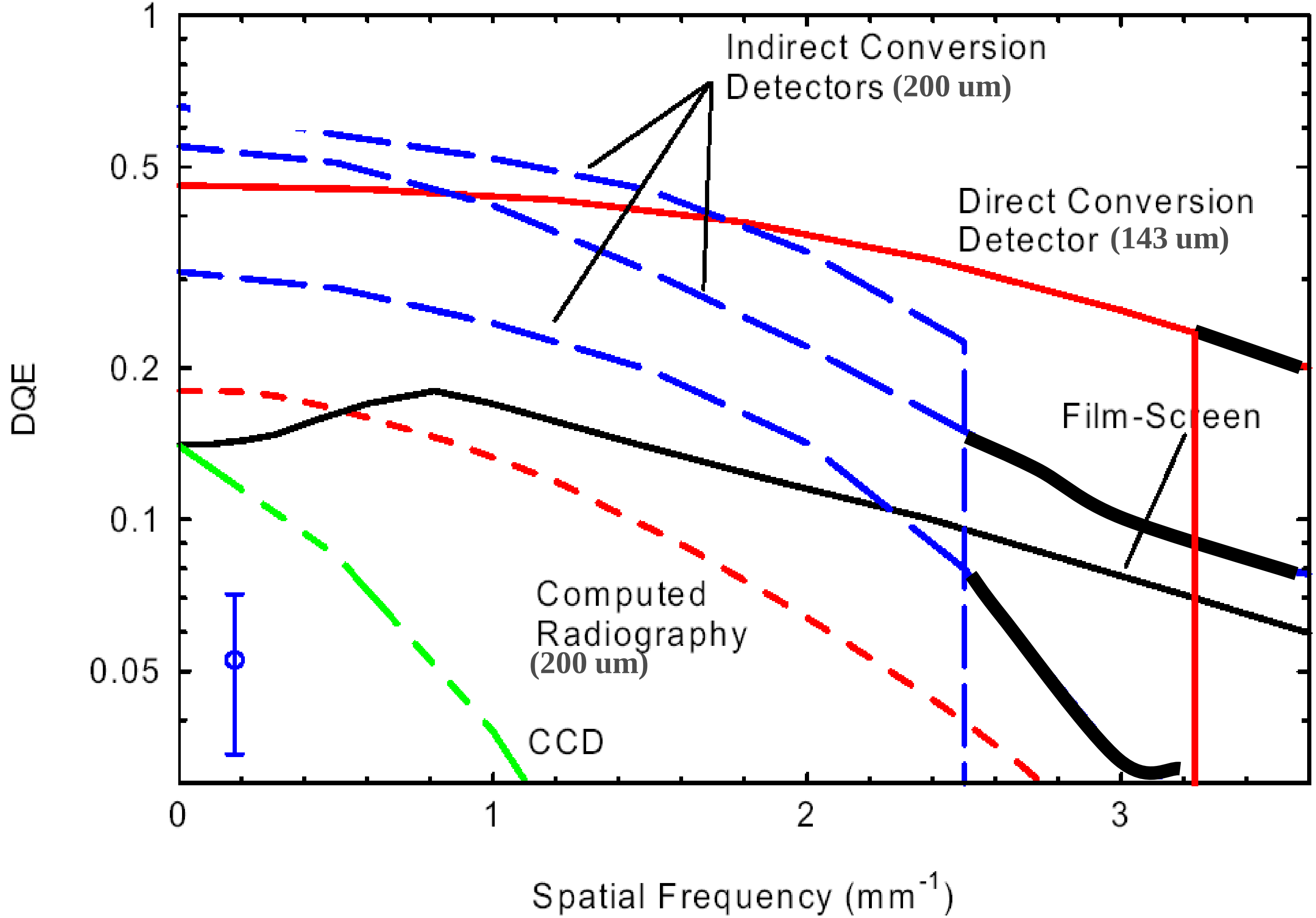
MTF of DR



The graph above shows the presampled MTF of DirectRay technology compared with the MTF of very high quality screen-film systems, Cesium Iodide detectors and a 14"x17" CR system.



DQE of DR



Indirect Conversion Detectors (200 um)
Direct Conversion Detector (143 um)
Film-Screen
Computed Radiography (200 um)
CCD
DQE
1
0.5
0.2
0.1
0.05
0
1
2
3
Spatial Frequency (mm^{-1})

Fluoroscopy

Fluoroscopy

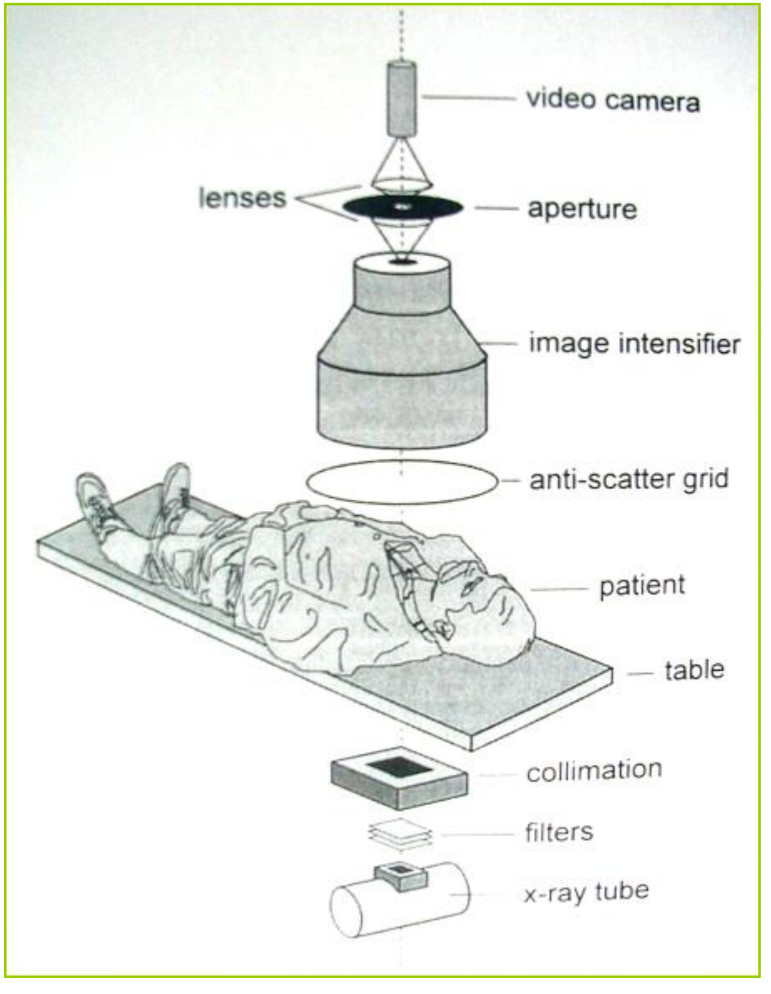
- Barium Studies
- Cardiac Catheterization
- Vascular Studies

Principles of Fluoroscopy

Major components include:

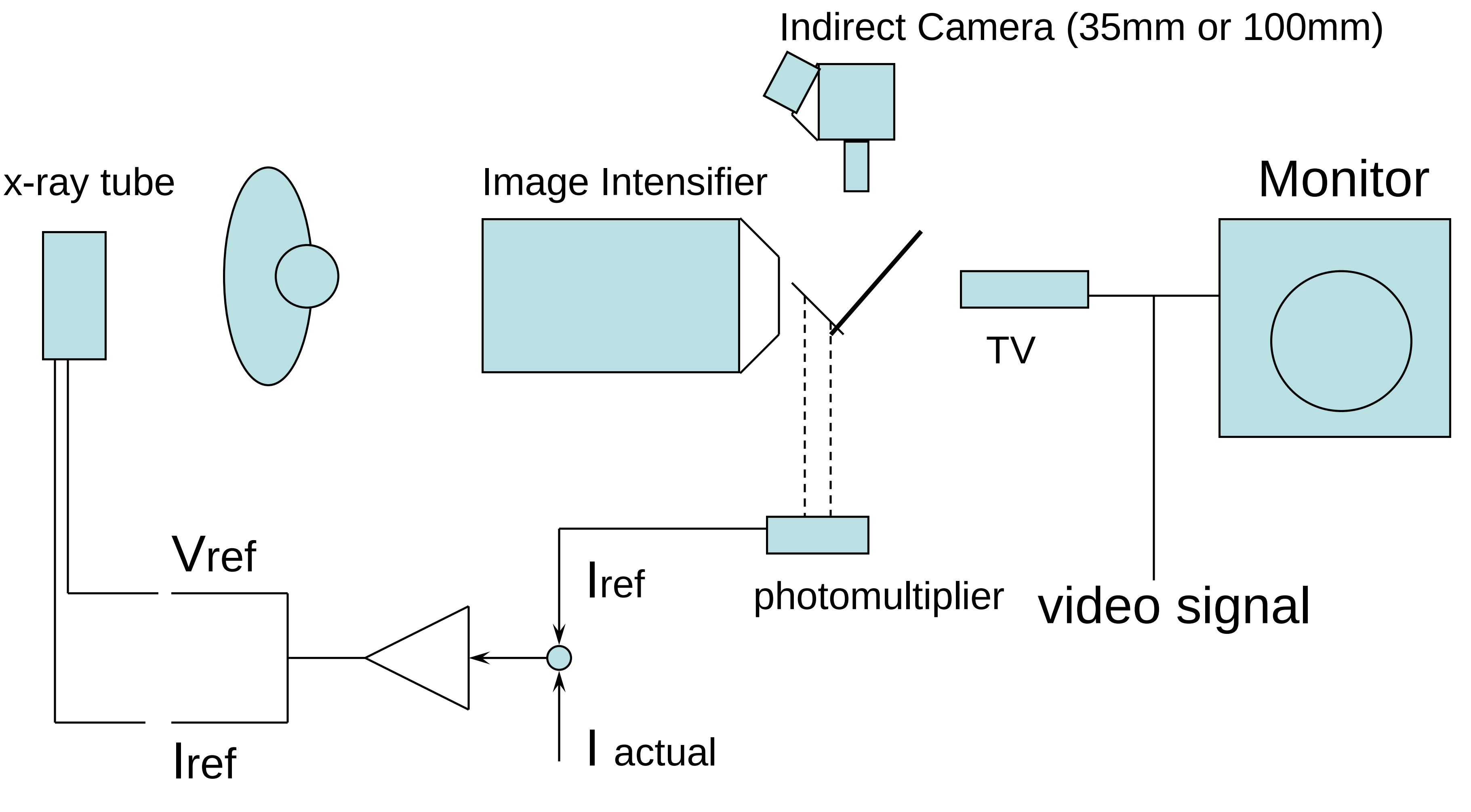
- X-ray tube: usually btw 50~110 kV
- Image intensifier (9~23 inches input size)
- Optical system
- TV or/and indirect camera
- TV monitor
- AEC (automatic exposure control) system

Fluoroscopy System



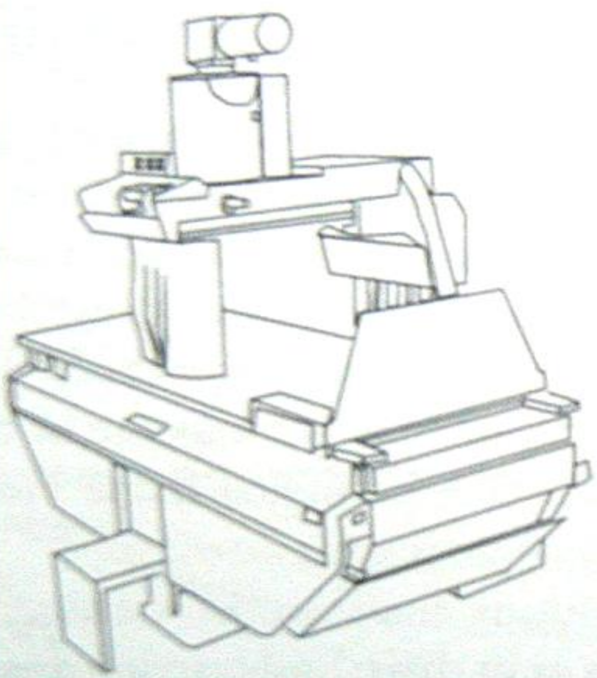
- X-ray tube positioned under table
- Collimation under table
- Grid at II entrance surface
- II positioned close to patient
- Output of II may be directed to several different recording devices

The Imaging Chain in Fluoroscopy

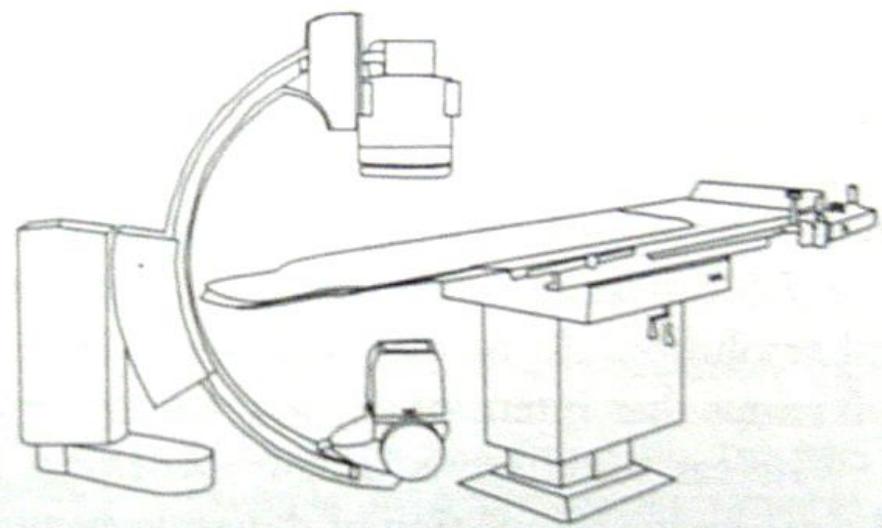


Fluoroscopy Suites

R & F System



Angiography System



- II moves up and down, can be moved to image different body parts
- Whole table can move up to 90^{o} to put patient in erect position

- C-arm type system
- Table moves up/down, swivels
- X-ray tube & II in fixed orientation

X-ray Tube and Collimation

- Setting of tube voltage and current must take into account the required contrast and the lowest possible dose for patient and operator
- Different operational range of the tube voltage and current is applied for particular applications
- Object contrast is a function of tube voltage

Imaging Intensifier (I. I.)

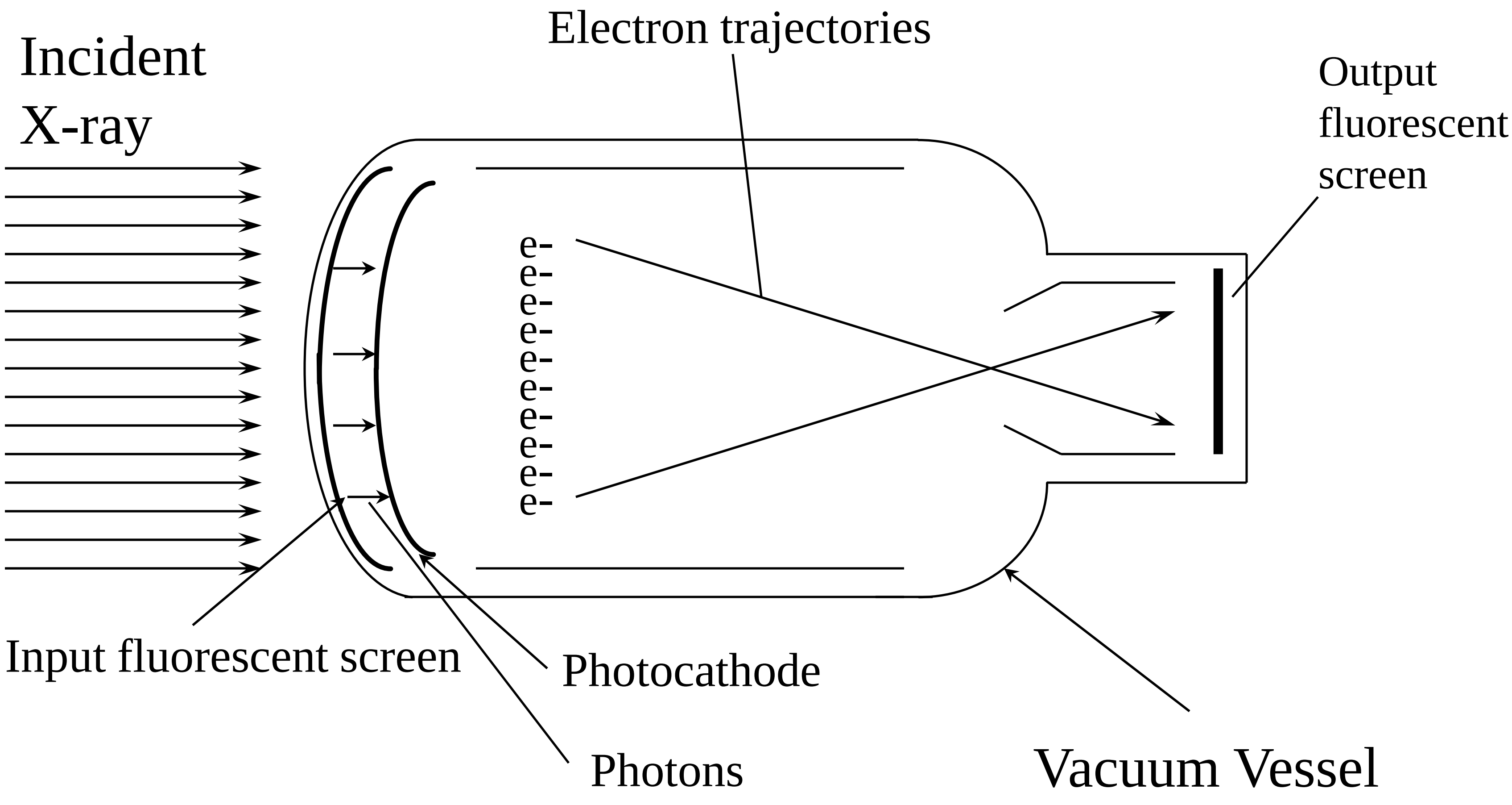


Image Intensifier (II): Principles of Operation

- The X-ray side of the I.I. is coated with metal. X-ray photon is converted to light photons at the input fluorescent screen (CsI crystals) which is thicker than for radiographic screens. The light spectrum is well-matched to the photocathode.

I.I. Principles of Operation

- The light photons are absorbed by the photocathode and converted to electrons. Photocathode is a metal compound which requires little energy to remove an electron form its surface (photoelectric effect, electrons have energy of approx.1-3 eV)

Structure of Cesium Iodide Needles

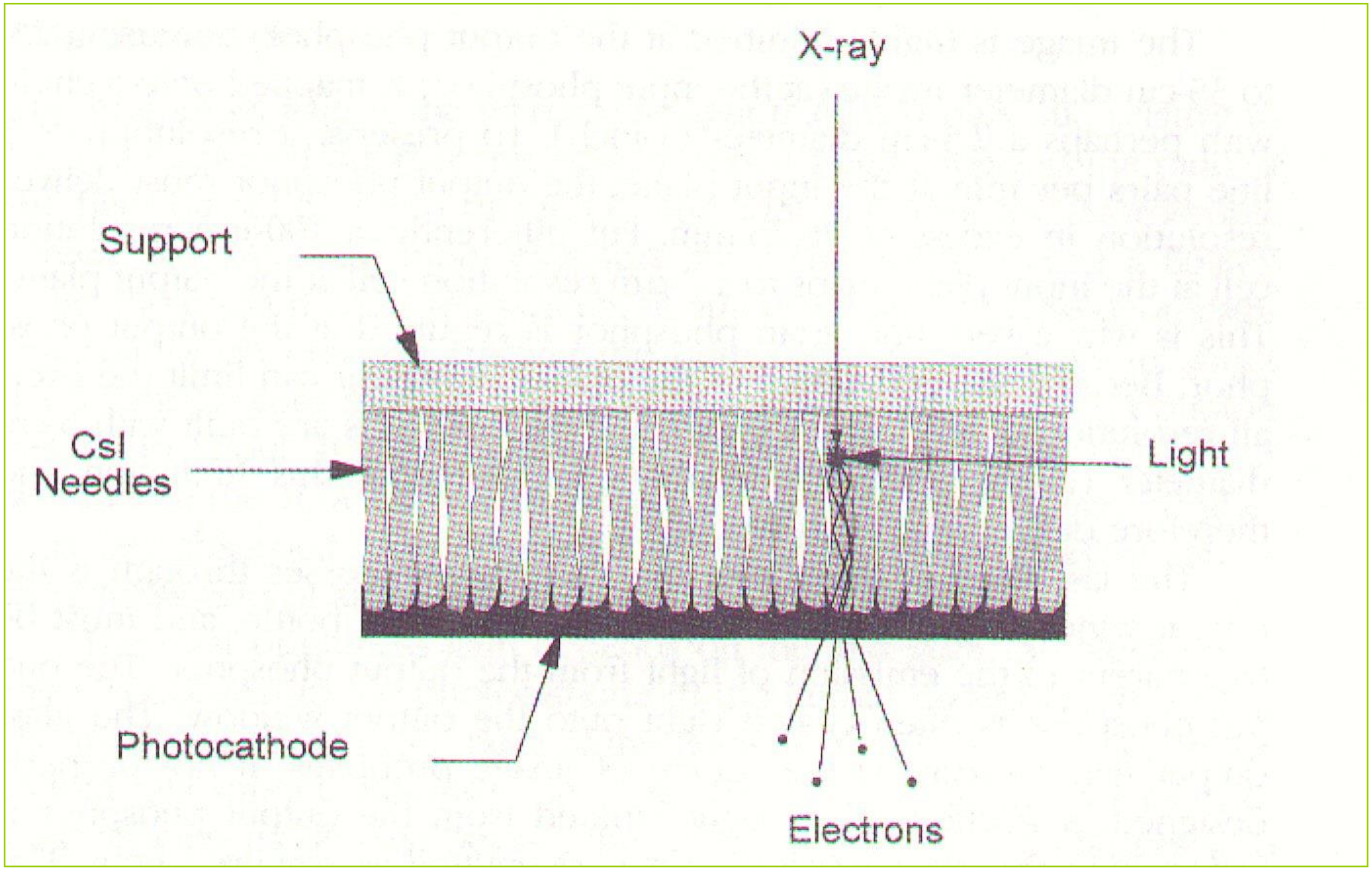
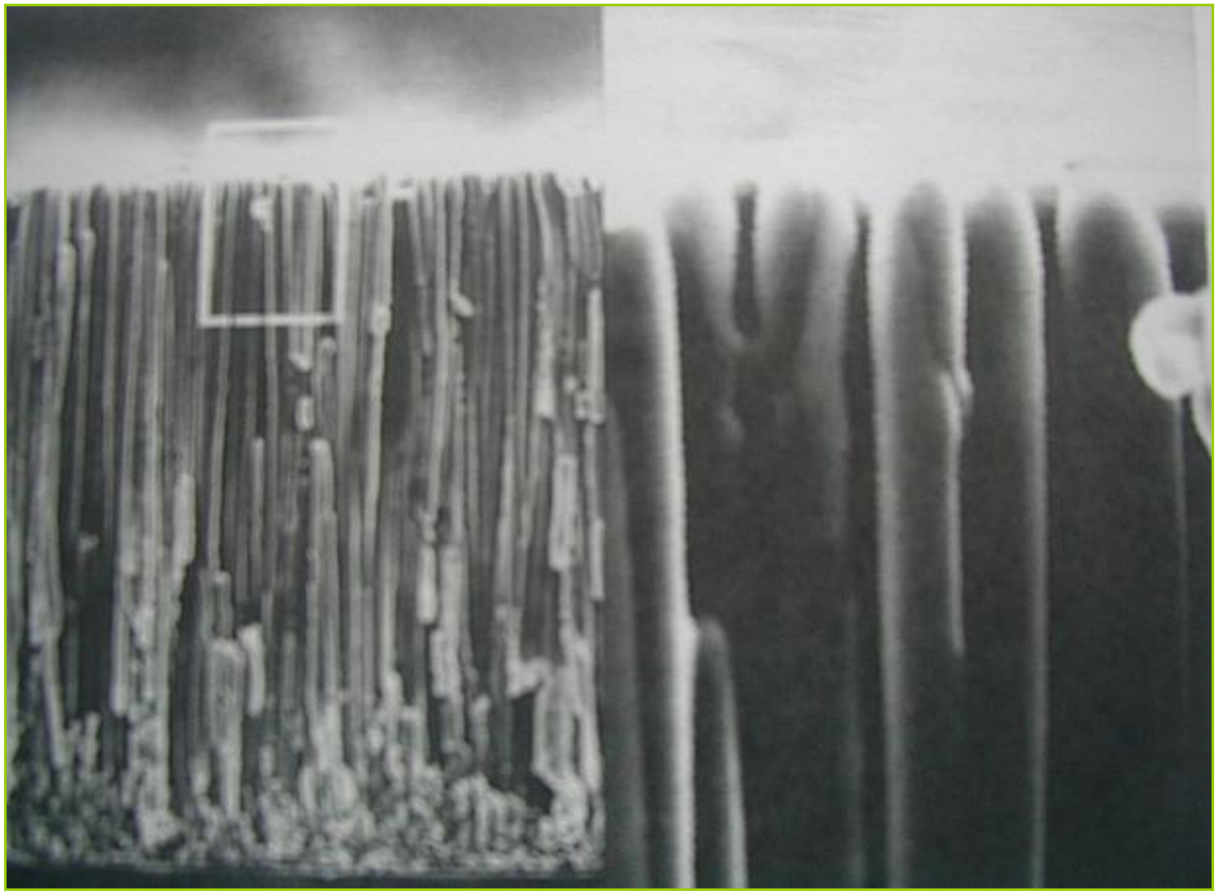


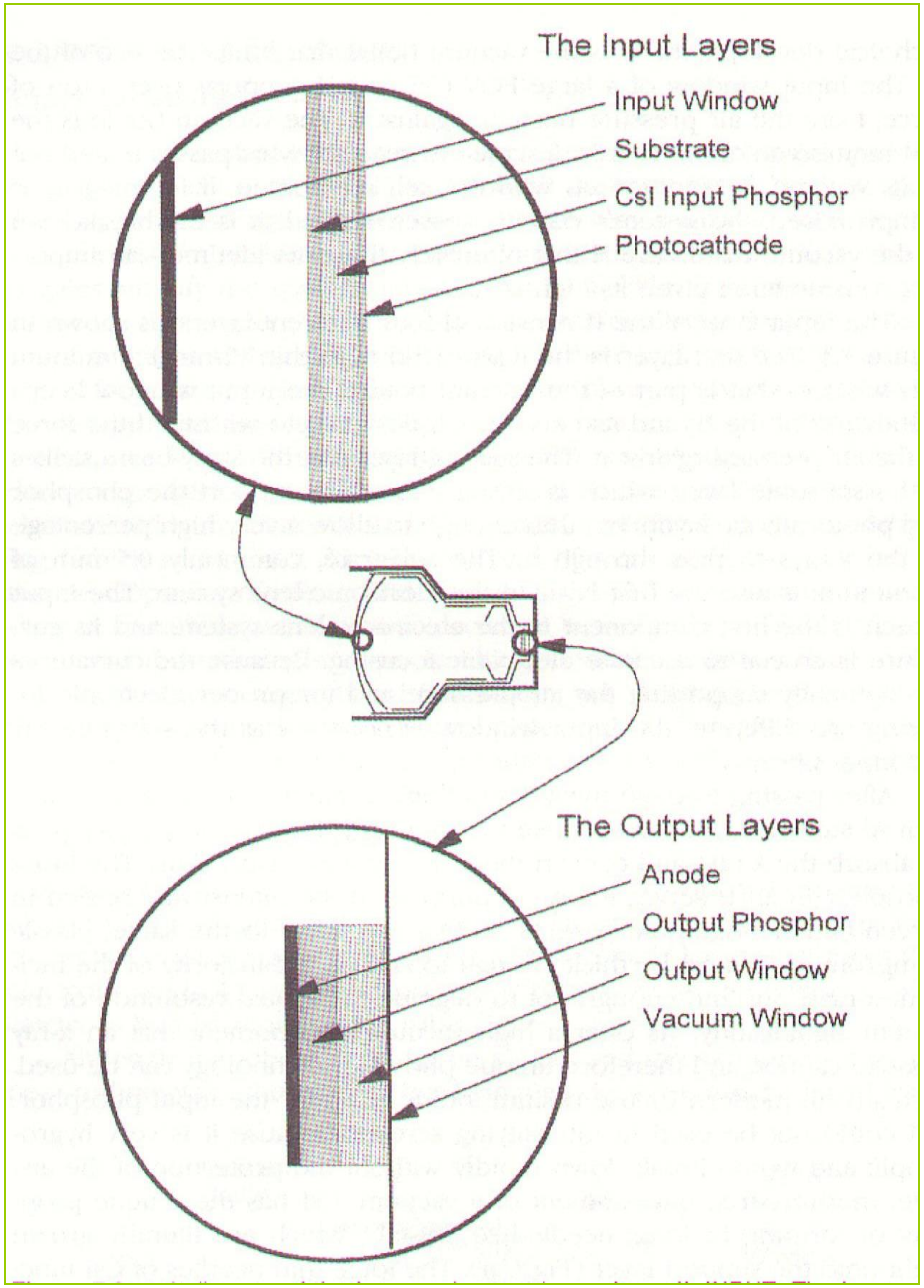
Photo of CsI Crystals



Scanning electron micrograph illustrates needle-like structure of CsI input phosphor

Image Intensifier Tube

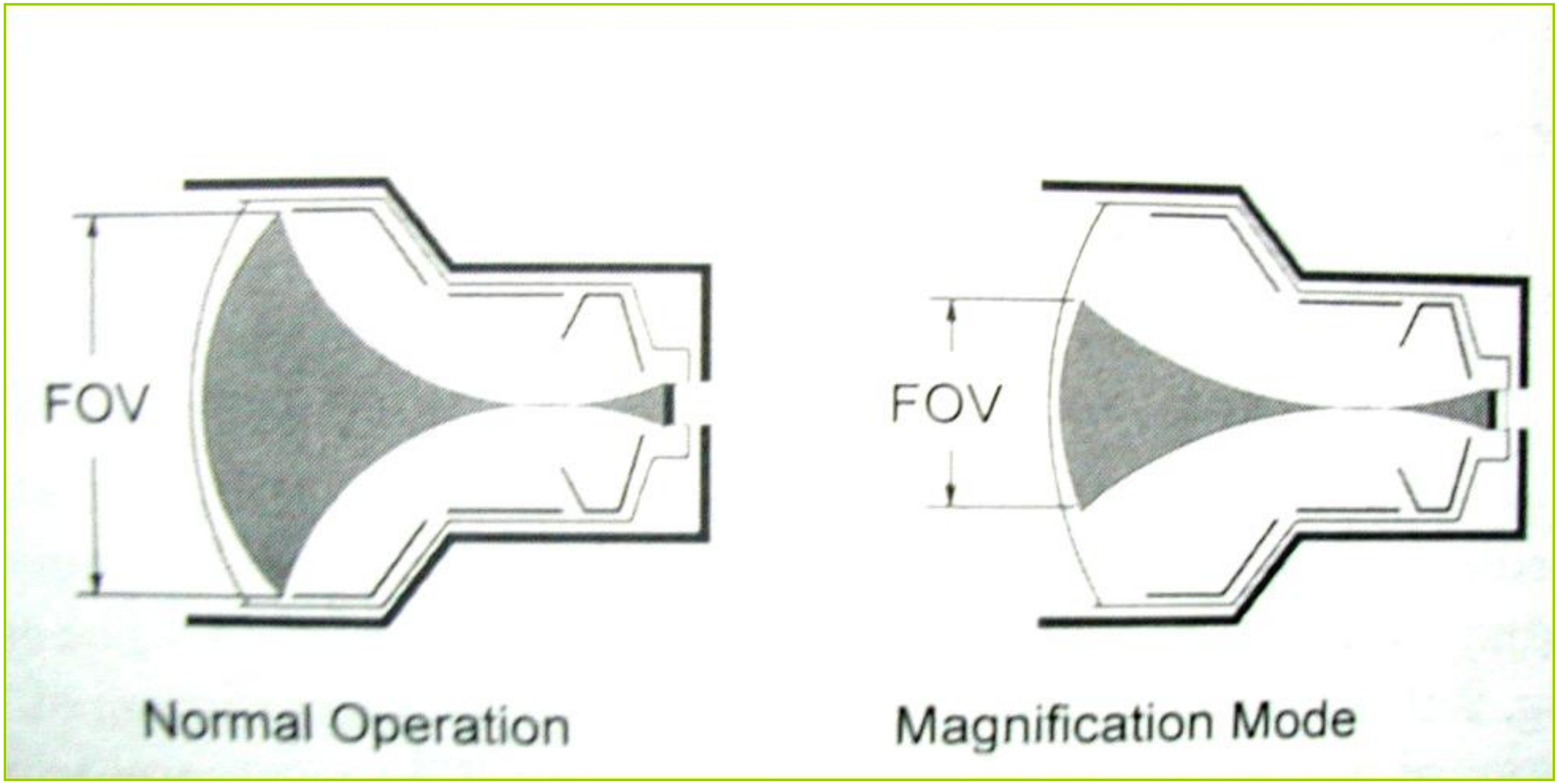
Input & Output Windows



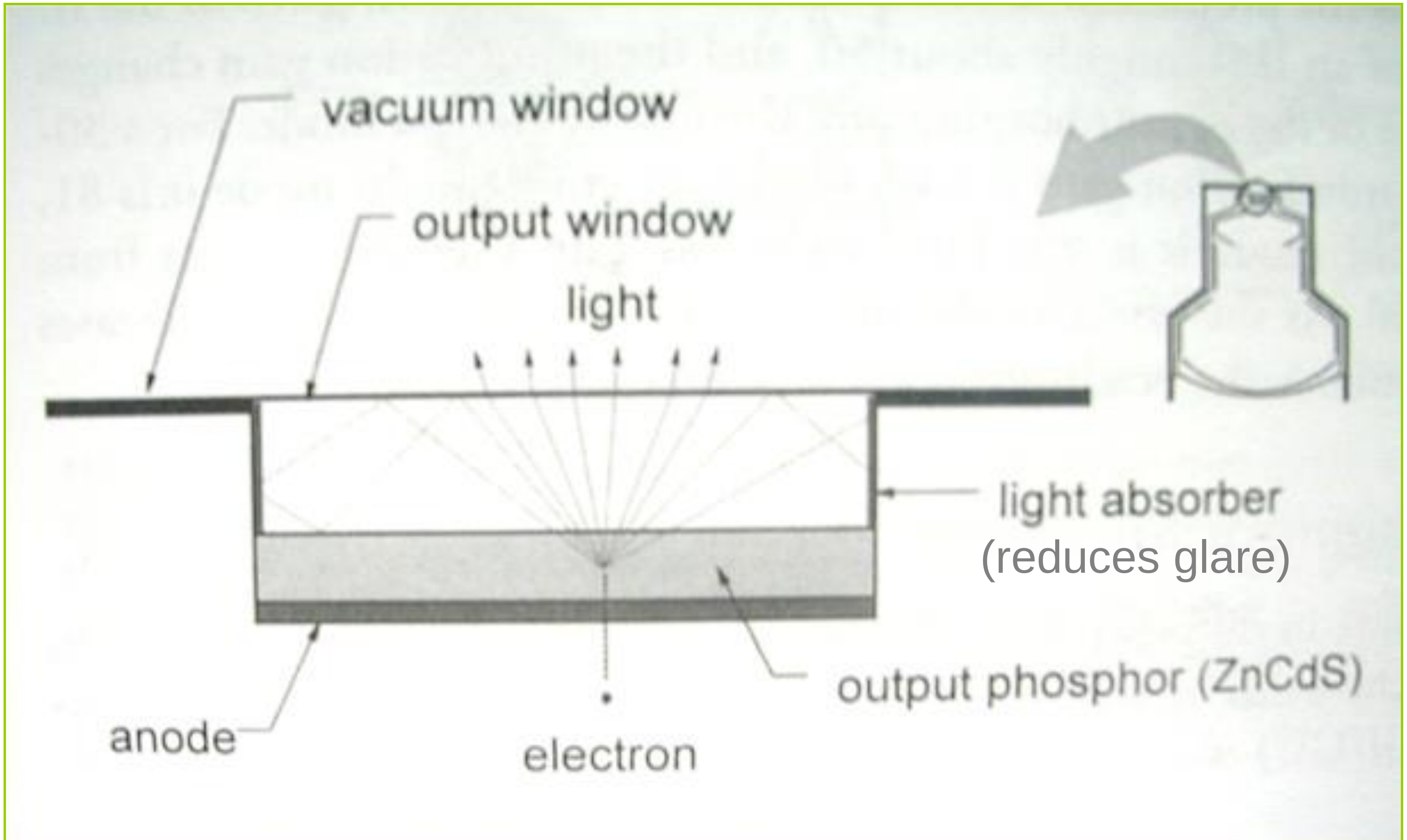
I.I. Principles of Operation

- The electrons accelerated by hi-voltage (~25kV) (Produces x-rays so I.I. is shielded to prevent escape)
- A series of electrodes acts like a optical lens system to focus e- to an output phosphor (diameter of ~1 inch)
- The electrons hit the output fluorescent screen and generate light photons

Normal and Magnification Modes



II Output Window



Automatic Exposure Control

- AEC accommodates changes in patient attenuation and keeps brightness constant with AGC in the video system.
- Incident exposure rate is adjusted in accordance to the attenuation
- The AEC control may operate under different modes (such as isowatt, anti-watt, minimal radiation, high image contrast). These modes ramp the kV and mA according to a predetermined curve and attenuation of patient.

Fluoroscopic Imaging Chain

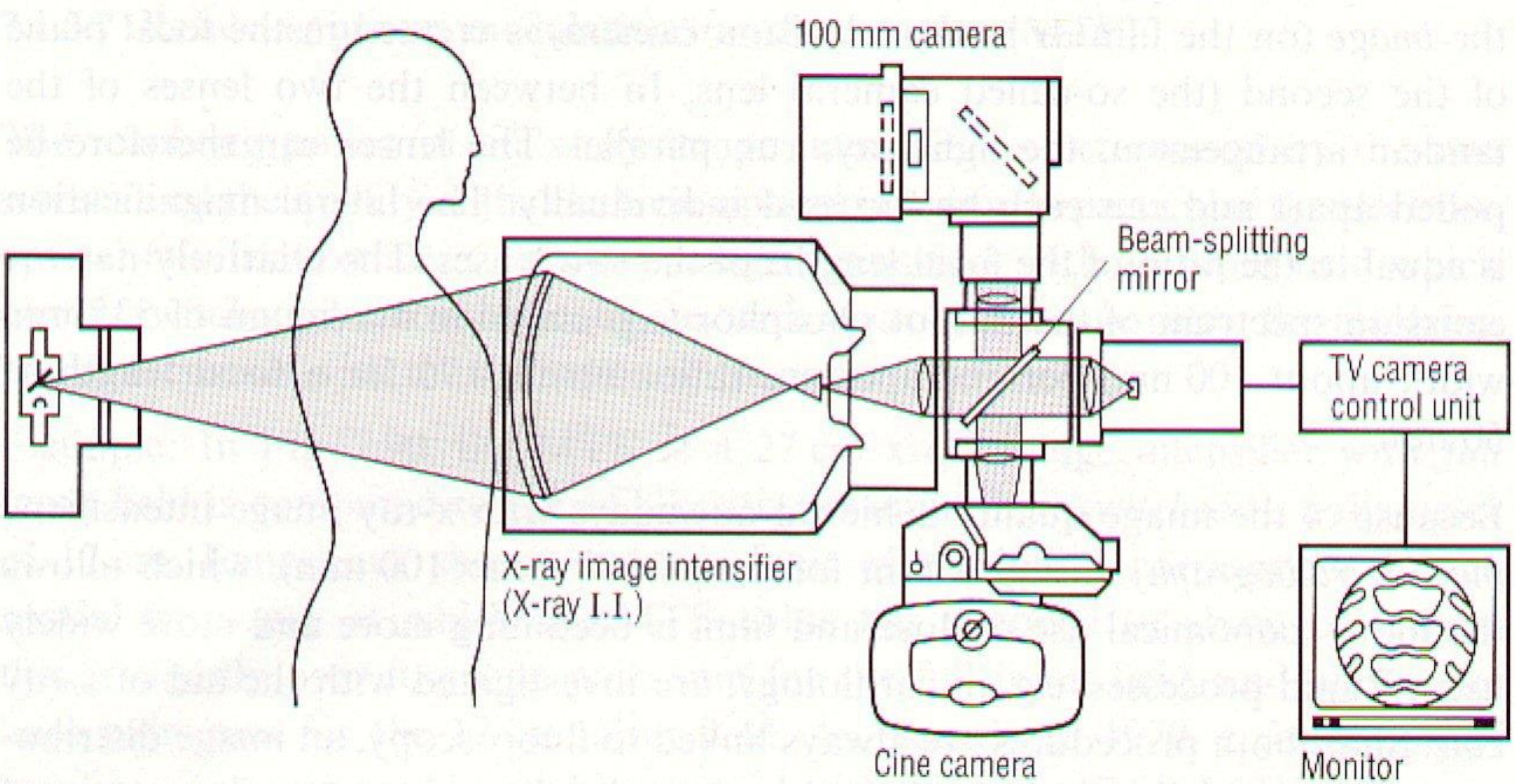


Fig. 7.92

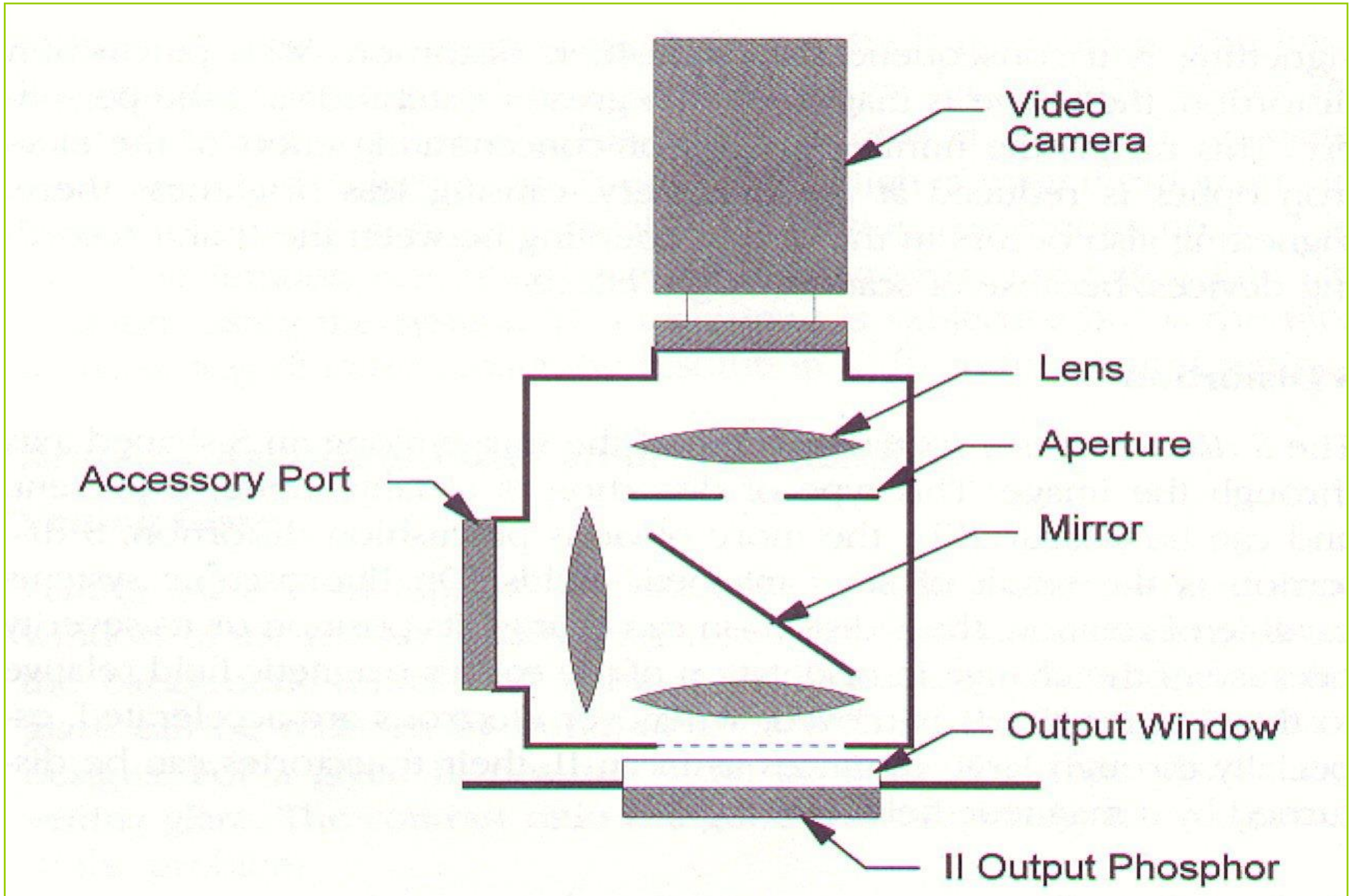
Optical Systems in Fluoroscopy

- The “lens” in the I.I. is made up of a series of positively charged electrodes which focus the electron beam as it flows from the photo-cathode toward the output phosphor. Electron focusing inverts and reverses the image.
- For undistorted focusing, all electrons must travel the same distance. That is why the input phosphor is curved to ensure the equal distance for e- from peripheral regions of II

Optical Systems

- Both the film camera and the TV camera focus on the I.I. output phosphor
- The I.I. light output is split into two paths by a semitransparent mirror; 90% for film, 10% for TV.
- Well adjusted film and TV camera usually do not cause the distortion, the I.I. does.
- Almost no loss of spatial resolution by the optical system itself.

Optics System

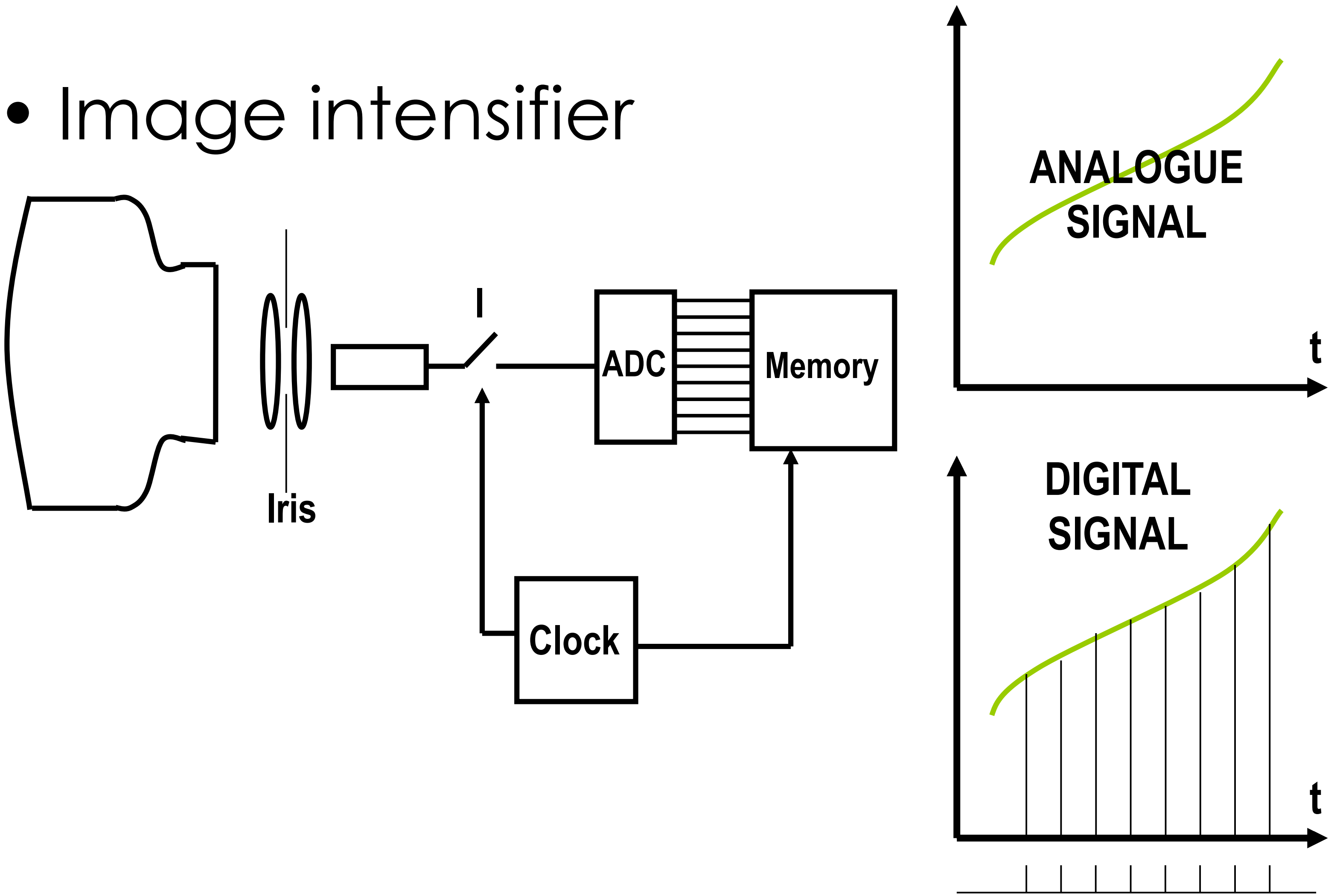


Indirect Camera

- Coupled directly to the I.I. output phosphor
- No loss of spatial resolution as the optics does not introduce losses and spatial resolution of the phosphor is the limiting factor
- Cine mode requires much higher patient exposure rate than fluoroscopic

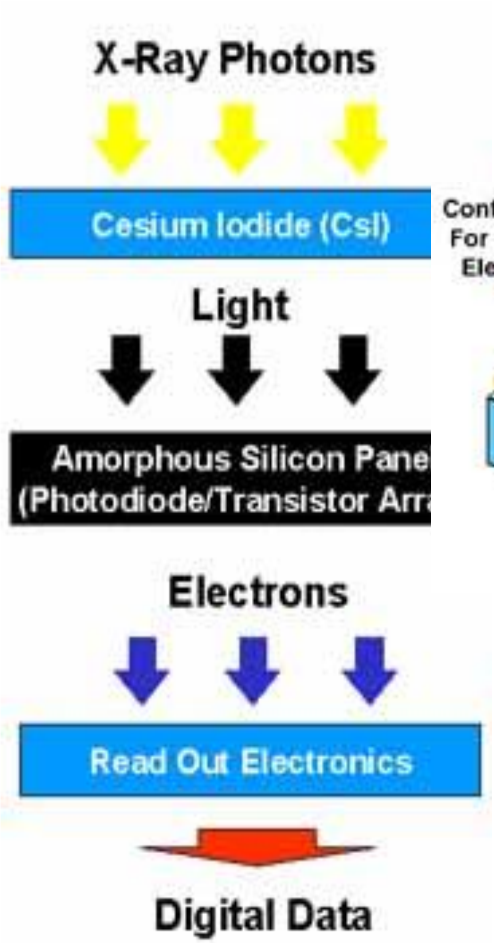
Digital radiography principle

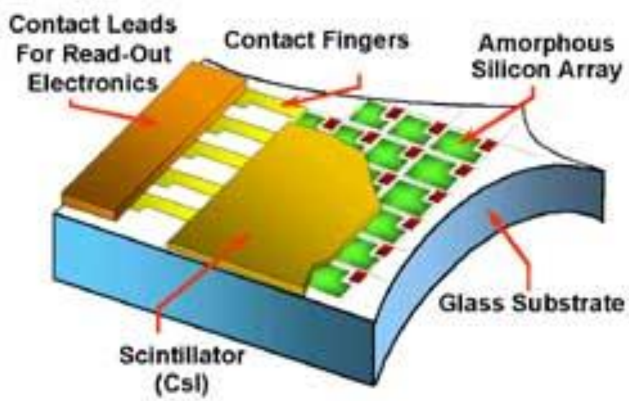
- Image intensifier

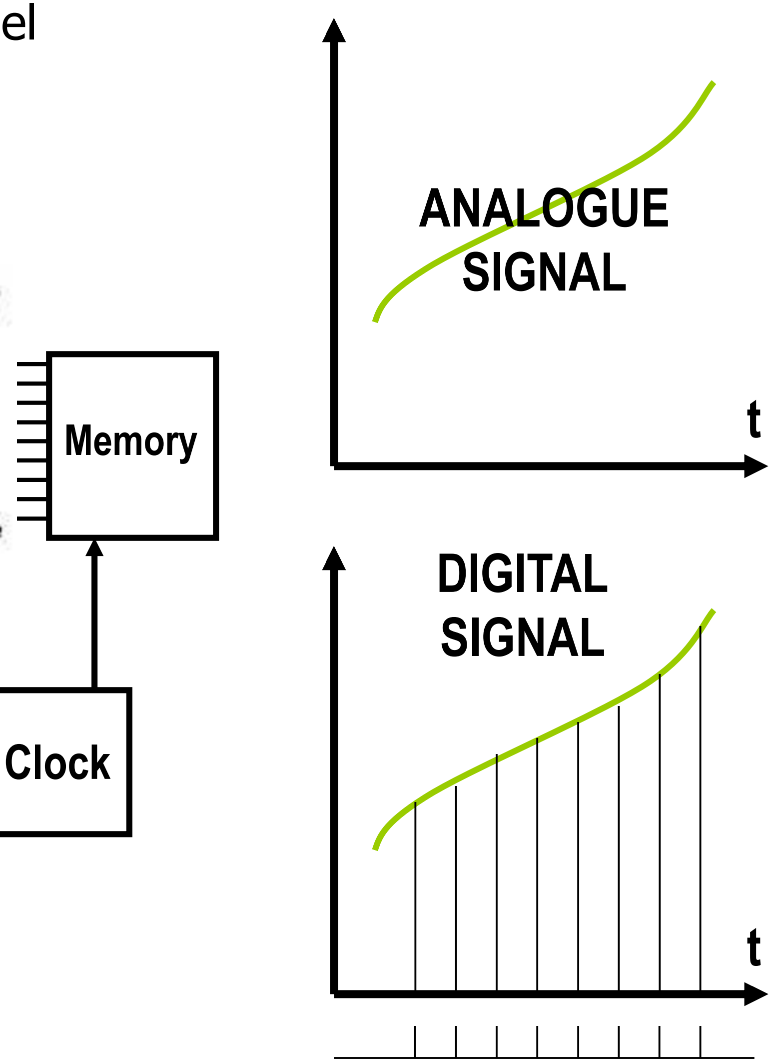


Digital radiography principle

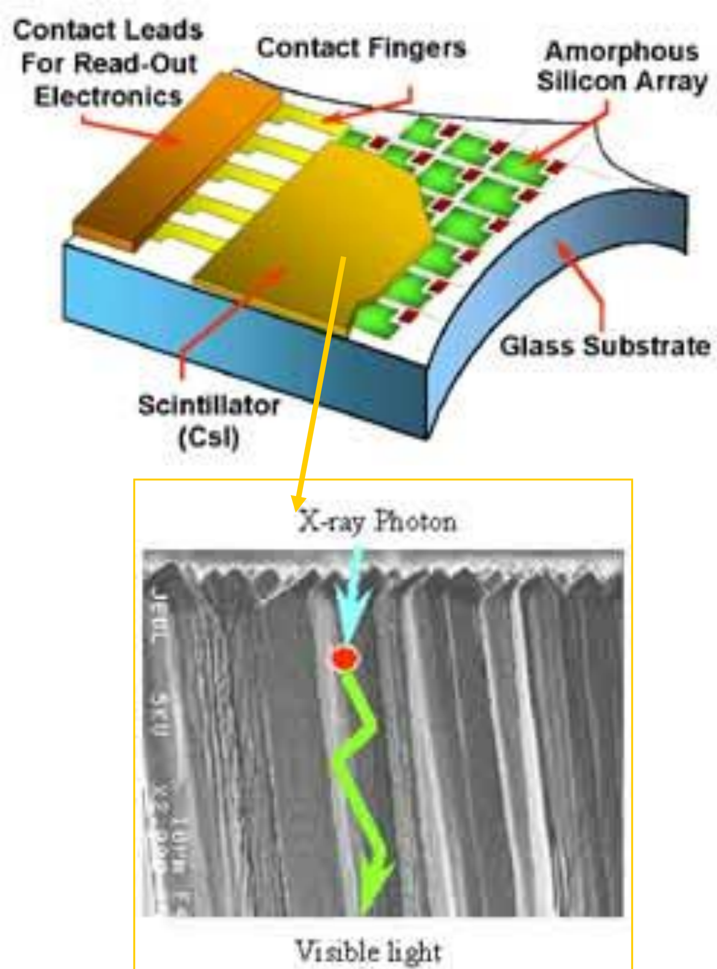
Dynamic Digital Flat Panel

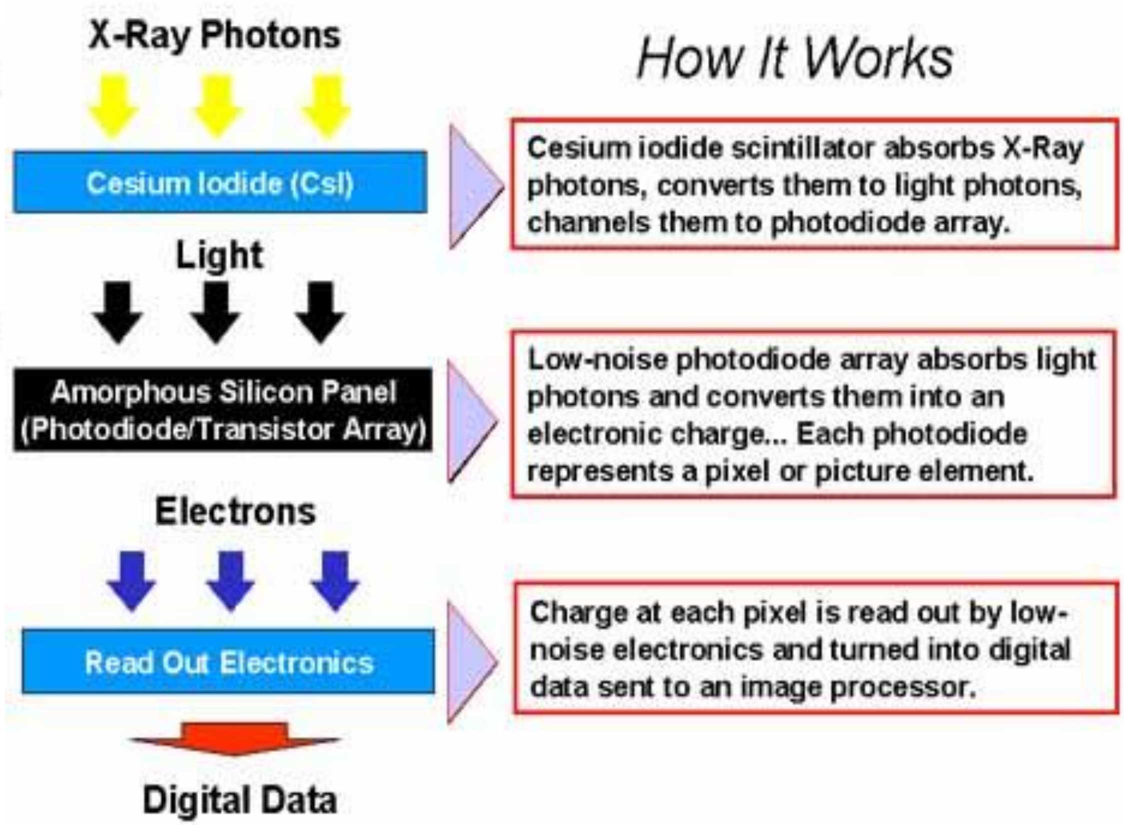






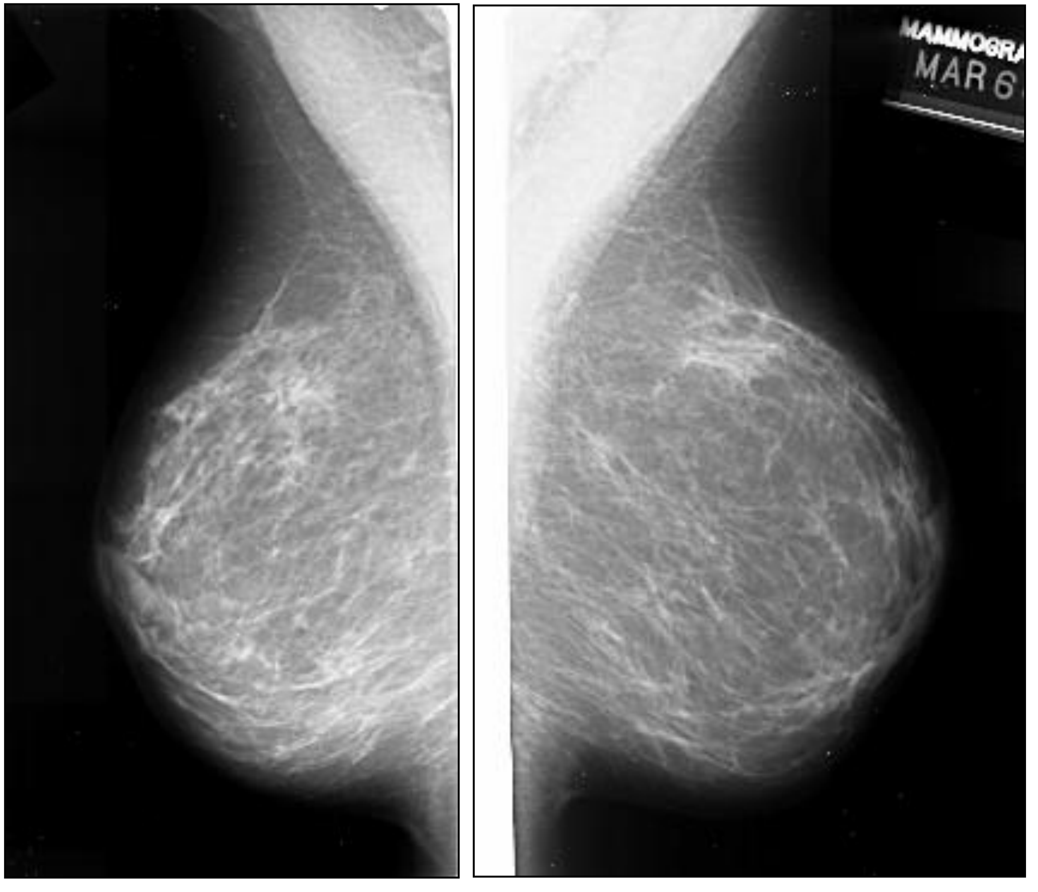
Flat panel technology: indirect conversion





Mammography

Screening Mammography

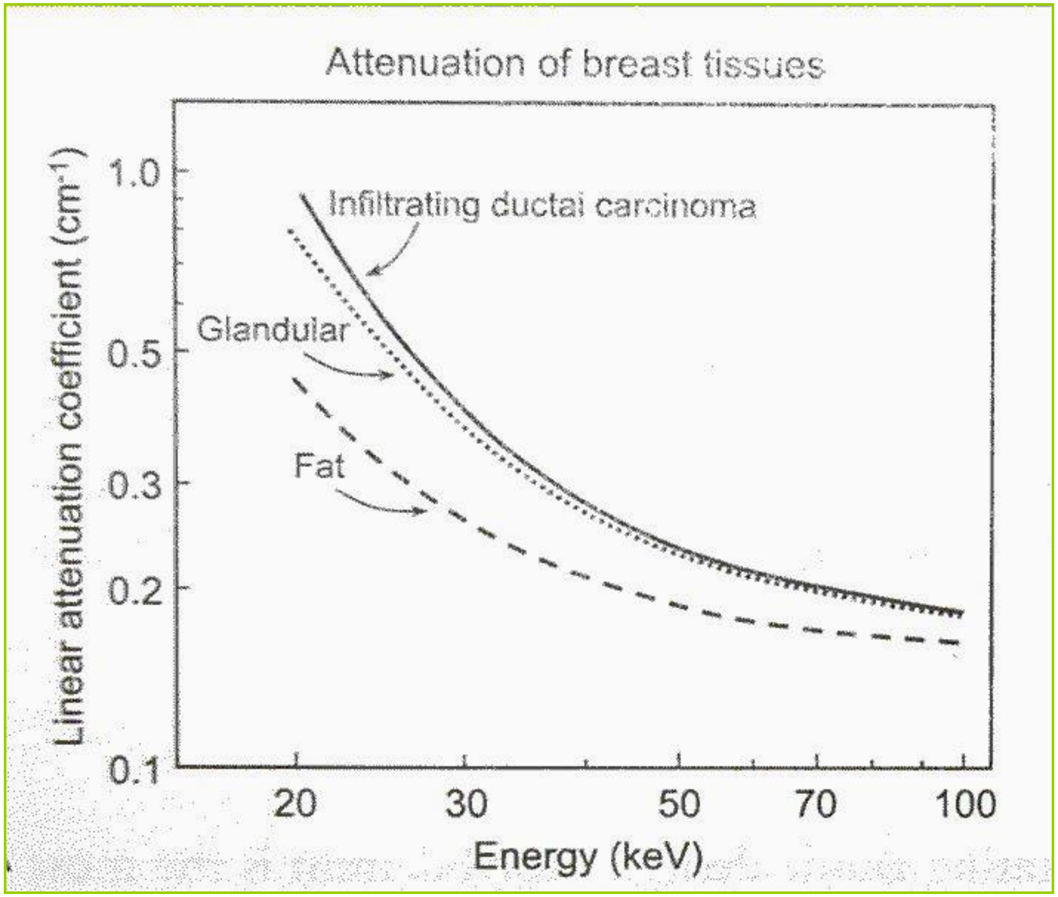


Goals of Screening Mammography

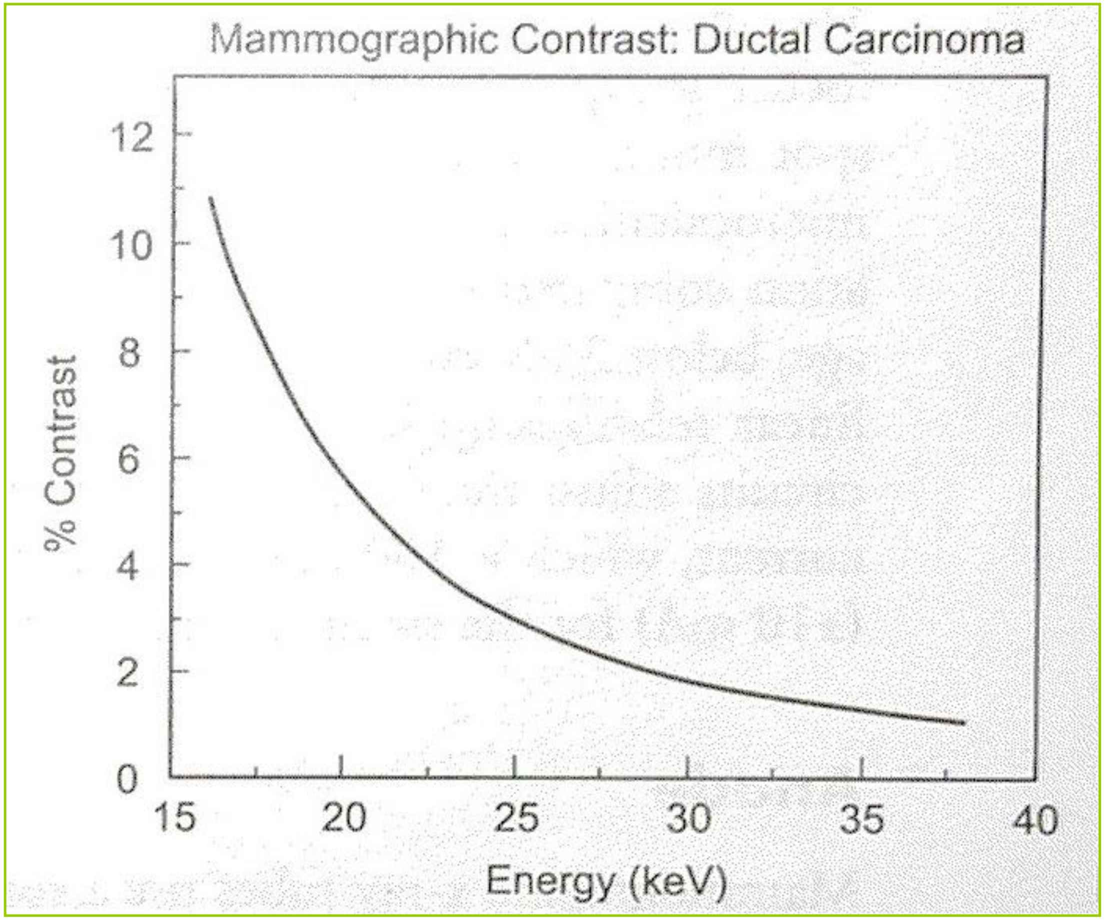
- To get the best quality of images with the lowest dose possible to the patient's breasts
- Image contrast and dose are the two factors that work against each other in screening mammography

How Does Mammography Technique Differ from Other Kinds of X-ray Imaging?

Mammography Requires Soft Tissue Contrast



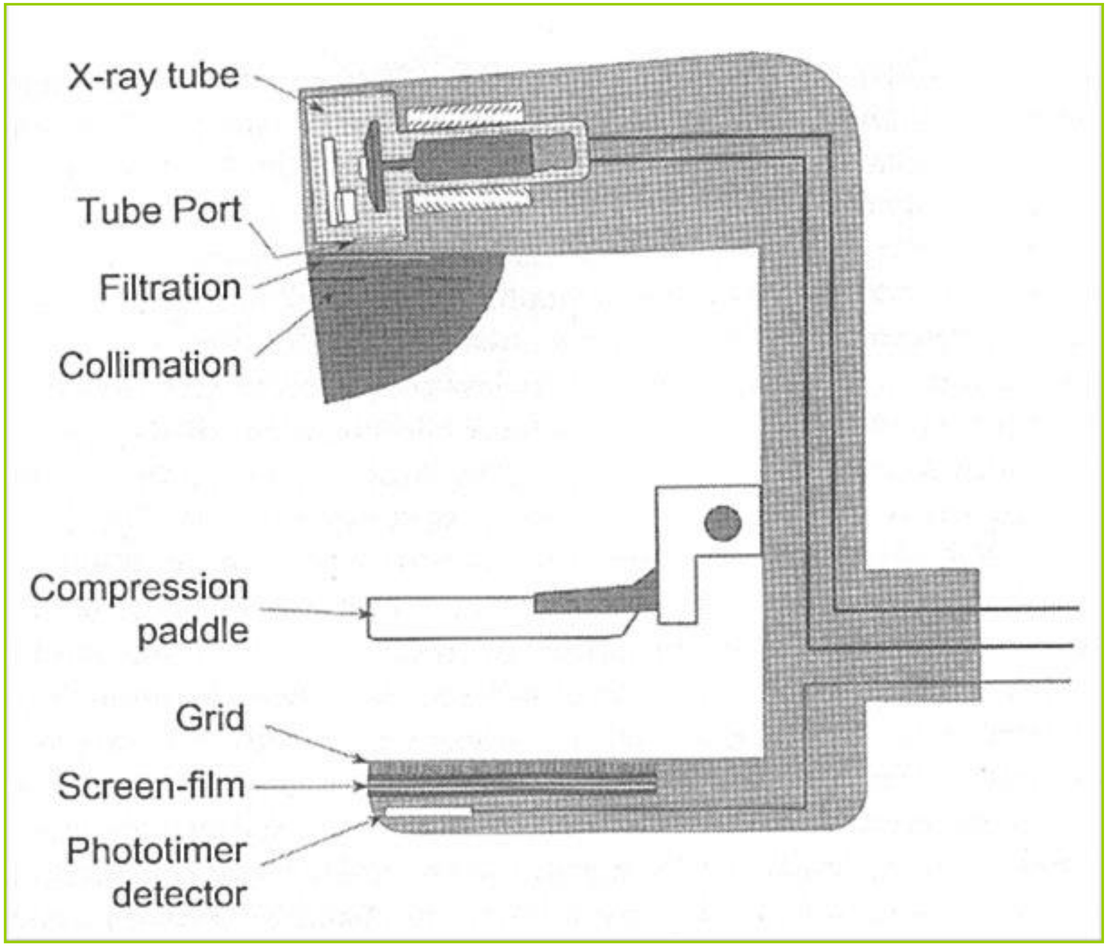
Percent Contrast of Ductal Carcinoma



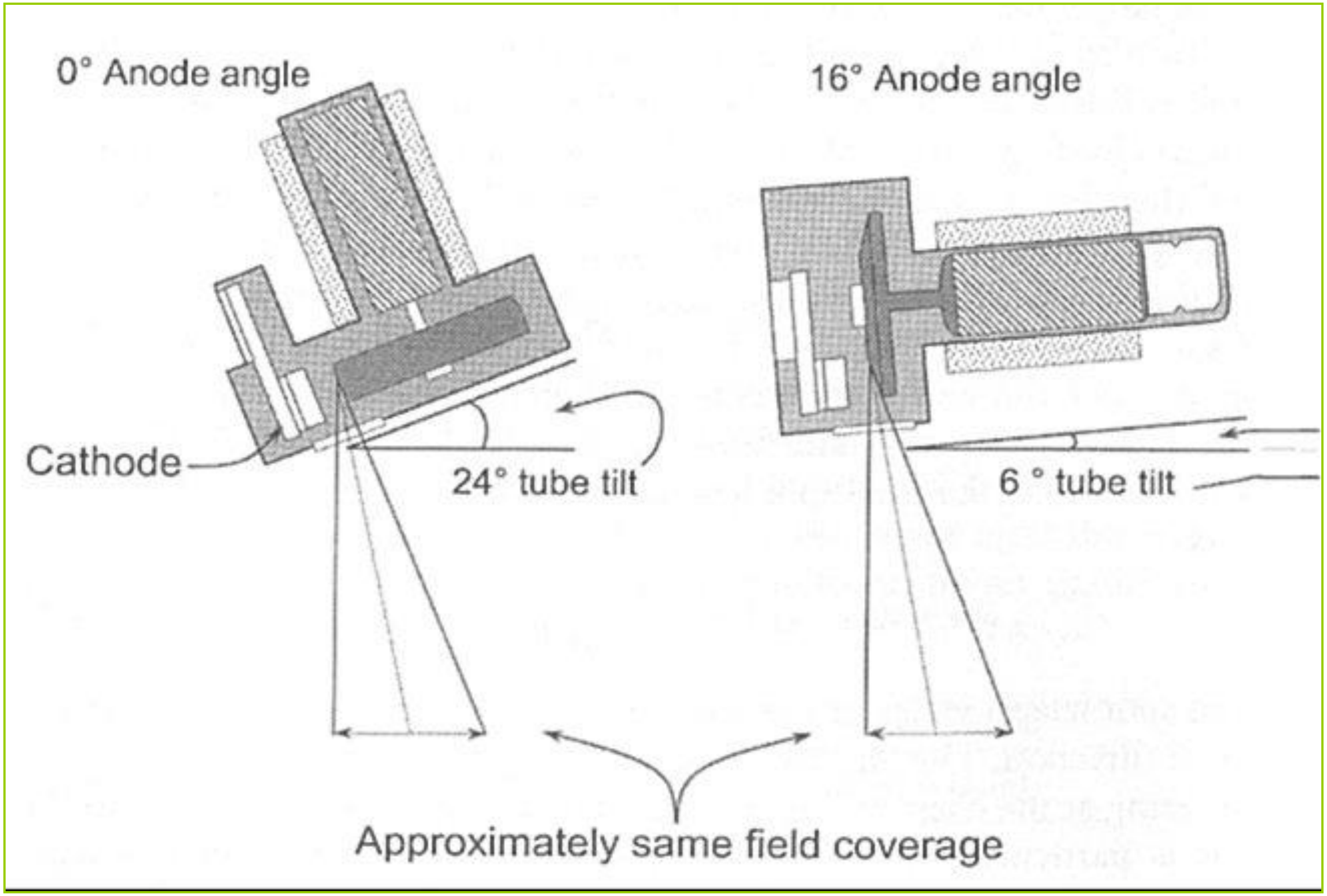
kVp

- Image quality of a mammogram and the glandular dose depends on x-ray spectrum.
- The shape of spectrum is determined by the anode material, filter and the kV.
- Contrast decreases as kVp increases
  - (because $\sigma \sim 1/E^3$)
- Absorbed dose decreases as kVp increases
- Thick, dense breasts require higher energies

Mammo System Design



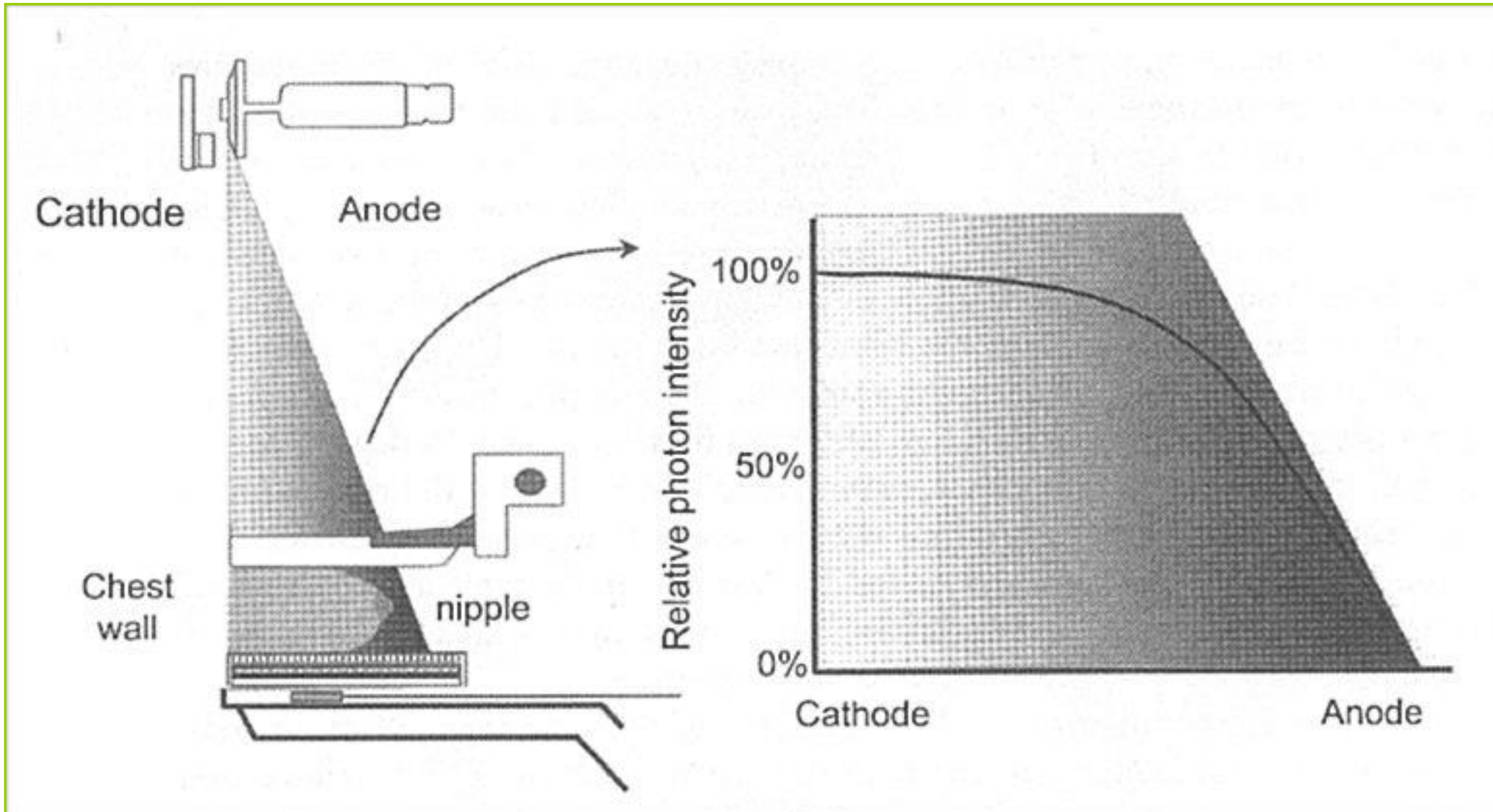
X-Ray Tube Design



Cathode & Filament

- Dual filaments
- 0.3 mm and 0.1 mm focal spot sizes
  - Minimizes geometric blurring
  - Maintains spatial resolution
- Space charge effect @ <35 kVp
  - Requires feedback circuit to measure nonlinear relationship between filament current and tube current
  - Modern mammo generators are multipulse type

Orientation of Cathode-Anode Axis



Left: Positioning of chest wall away from anode side of tube.
Right: Change in beam intensity due to Heel effect.

X-ray Source

- Small focal spot size (less blur)
  - Large FS = 0.3 mm to 0.4 mm
  - Small FS = 0.1 mm to 0.15 mm
- Good Geometry
  - SID $\geq$ 66 cm improves resolution
- Large output –
  - >short exposure time
  - >Less motion blur
- Stable, consistent and linear

Filtration and Beam Quality

- Optimal mammography beam would be monochromatic with energy varied from 15 keV – 25 keV with breast thickness
- Specific x-ray tube target materials are used to try to come as close as possible to this goal

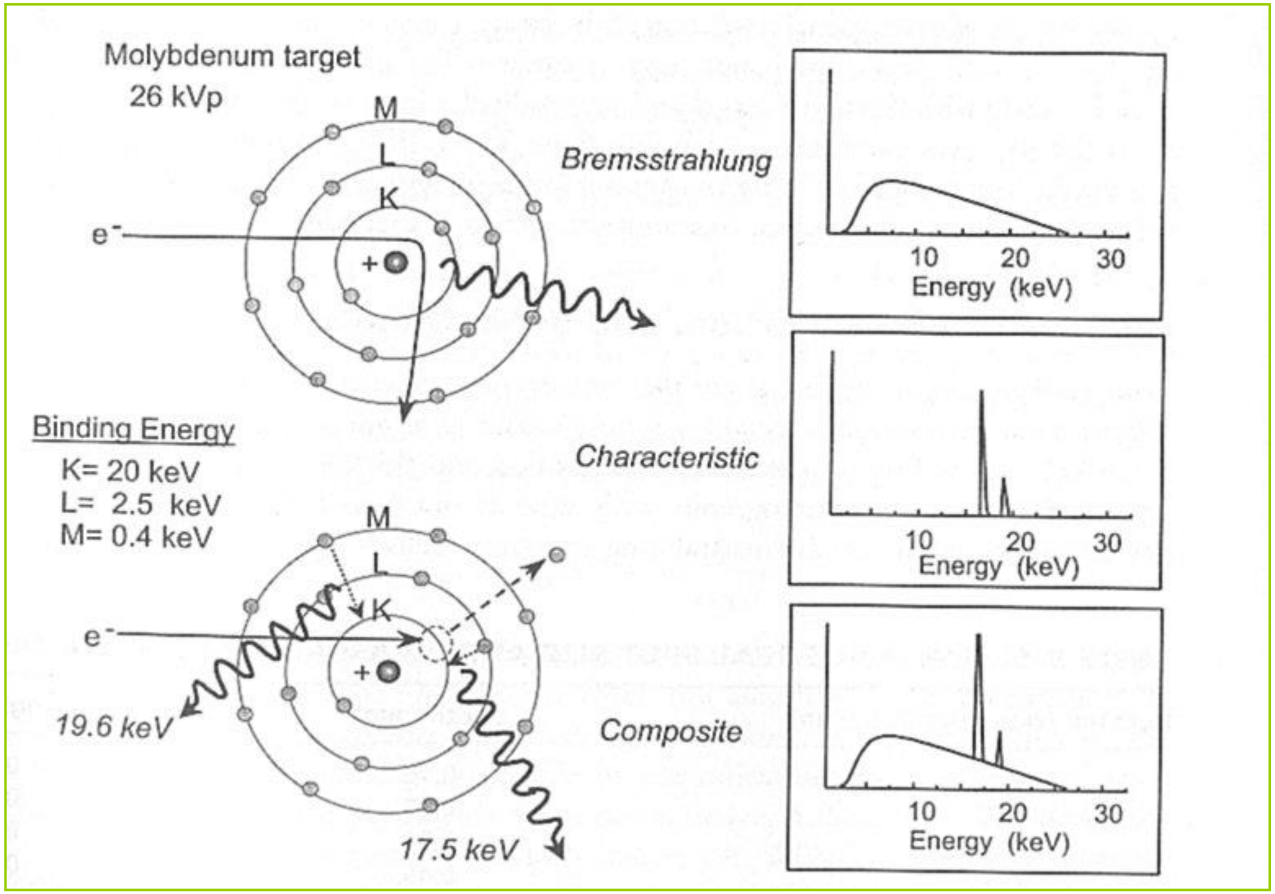
Imaging Systems
Anode/Filter Materials

- Molybdenum (Mo/Mo, Mo/Rh)
  - K_α =17.9, K_β = 19.5 keV
- Rhodium (Rh/Rh)
  - K_α =20.2, K_β = 23.2 keV
- Tungsten (W/Al)
  - K_α =59, K_β = 68 keV

Anode Selection

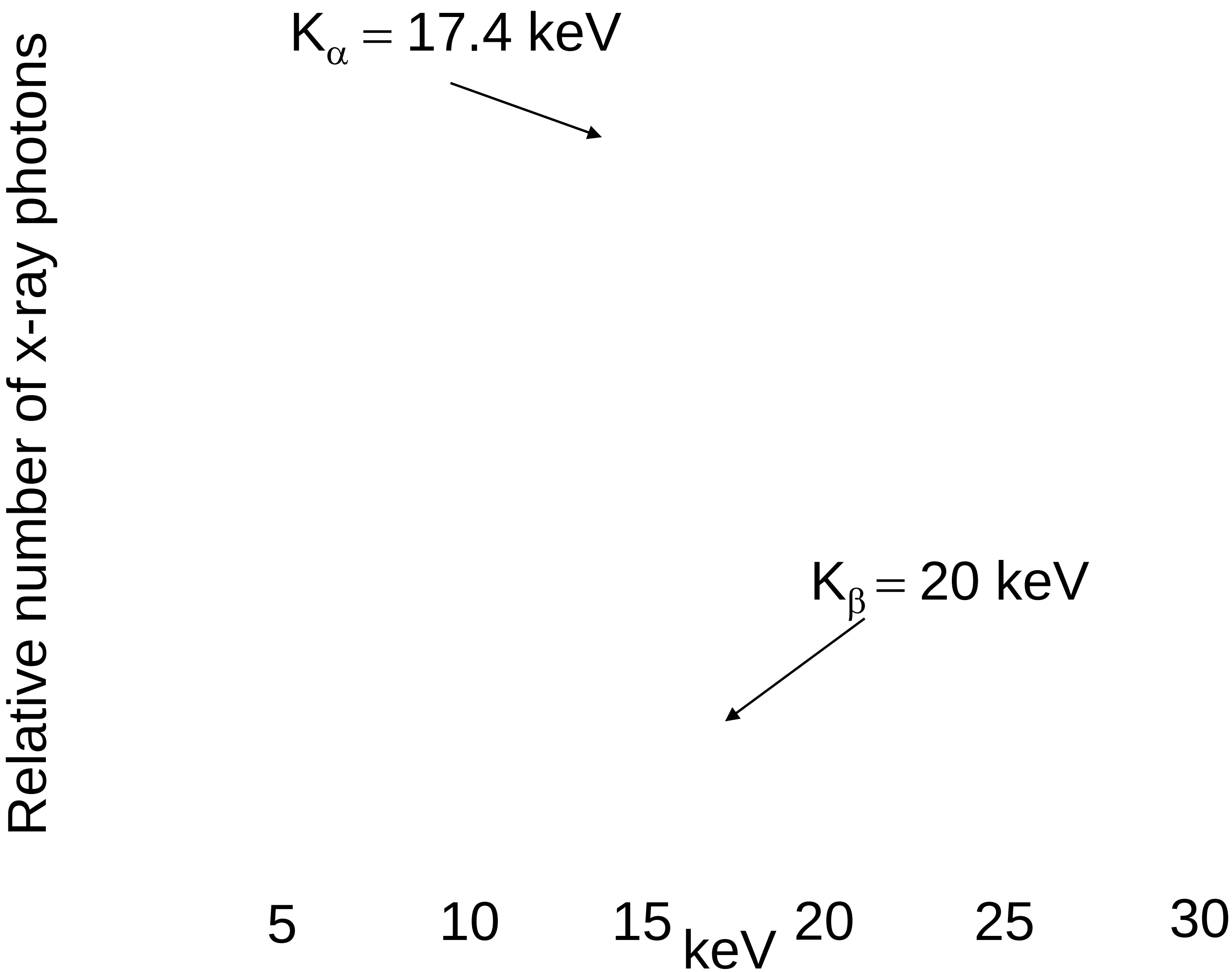
- Mo/Mo tube for small or medium sized for dose optimization
- Change filter from Mo to Rh increase average energy
- W/Rh tube for thick/dense breast and more used in digital mammography
- <4 cm: 26kV Mo/Mo; 4~4.6cm: 27kV Mo/Mo; 5.5~7cm: 27kV Mo/Rh; >7cm: 26kV W/Rh

Bremsstrahlung & Characteristic

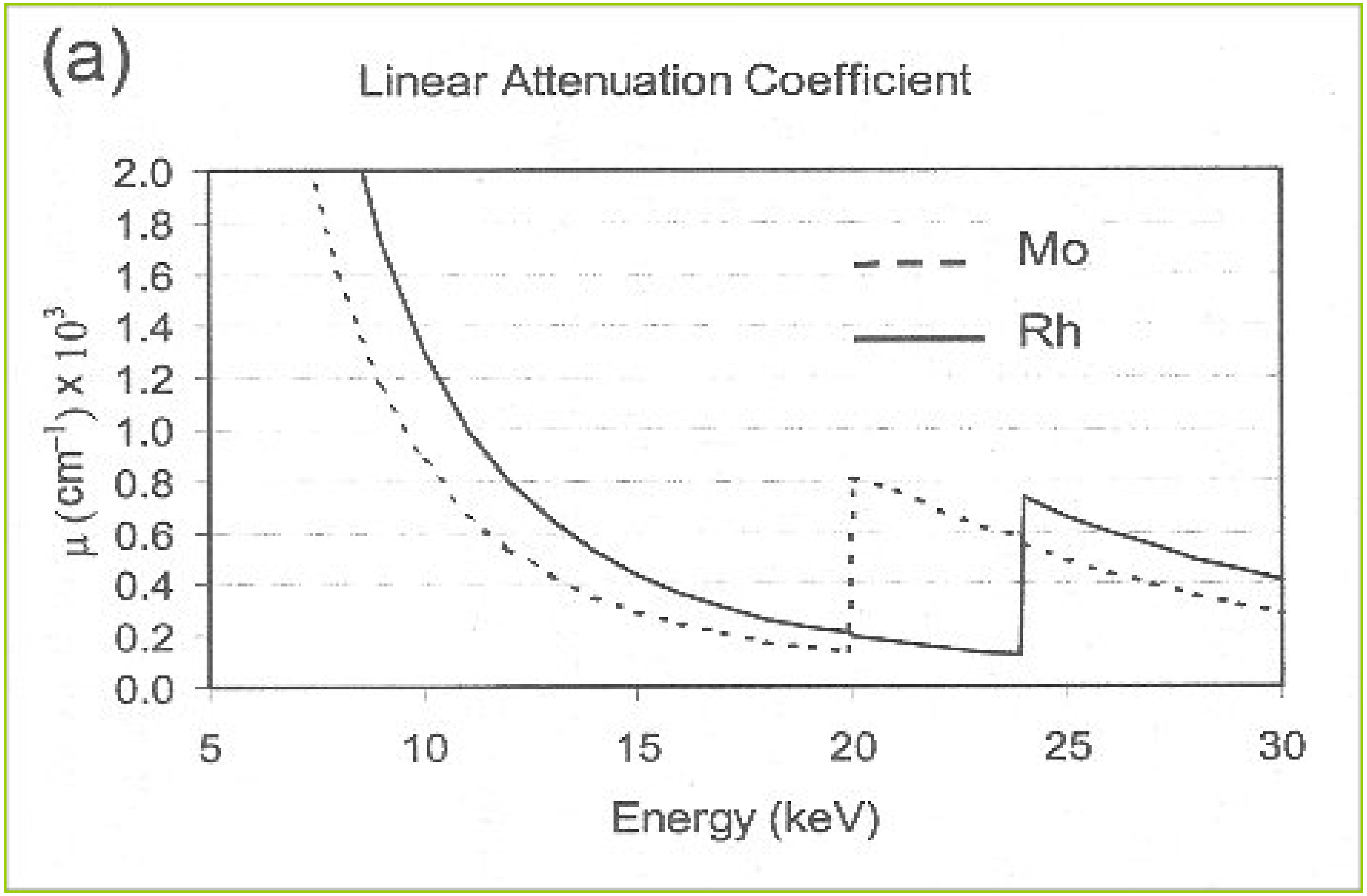


Characteristic energies of molybdenum (17.5 & 19.6 keV) are nearly optimal for detection of low contrast lesions in breasts from 3-6 cm thick.

Molybdenum Anode Spectrum

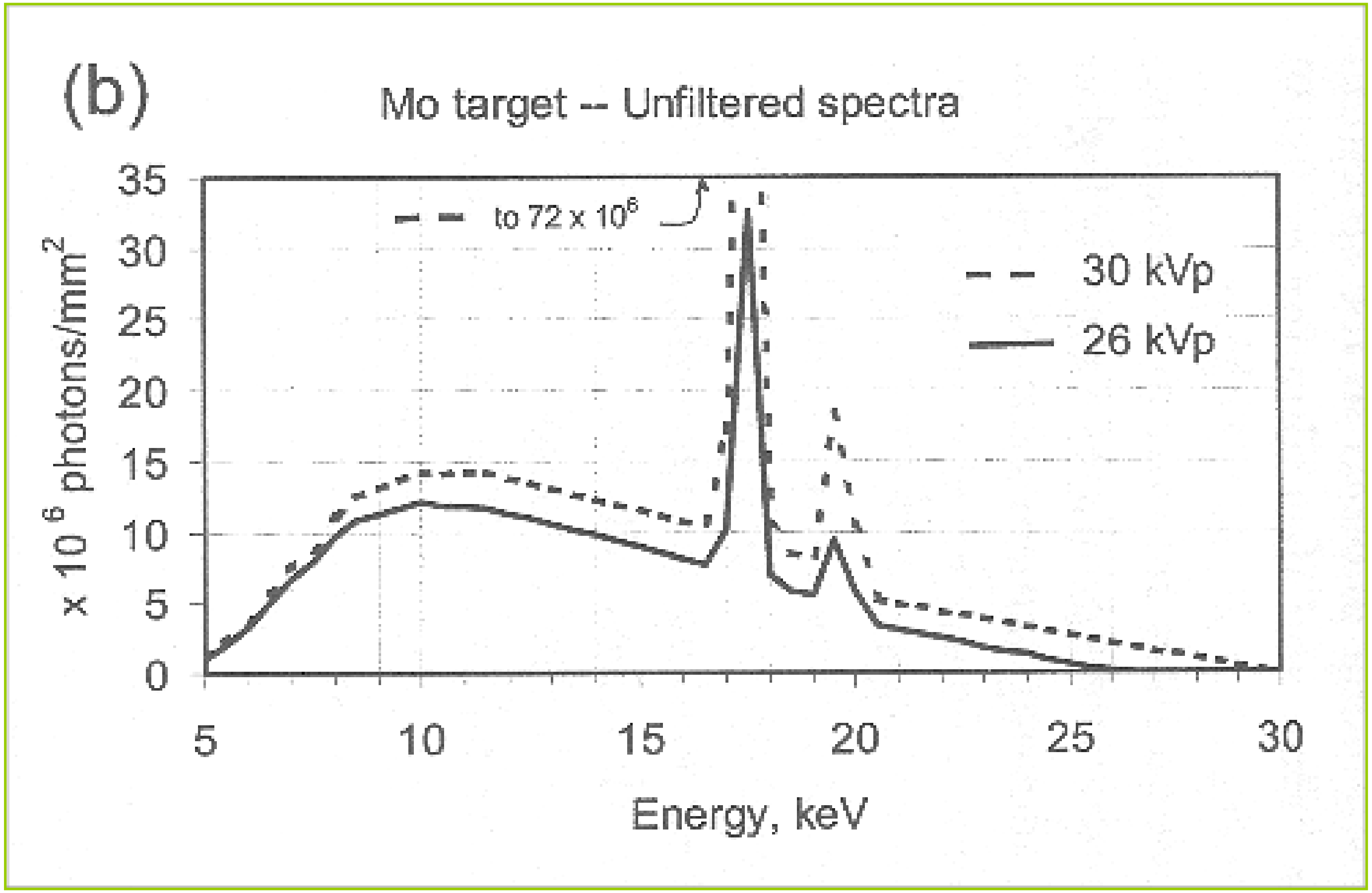


Linear Attenuation Coefficients



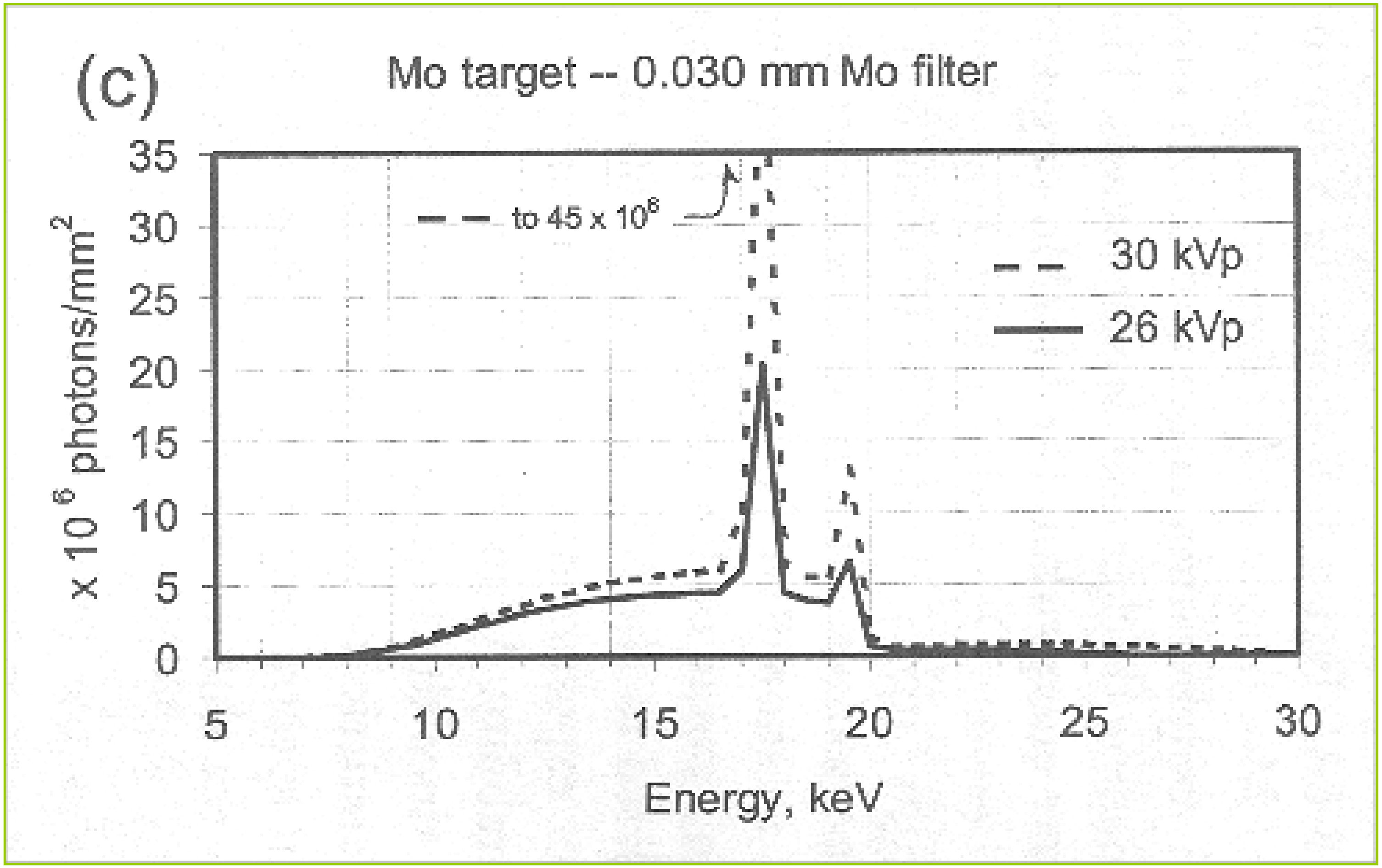
Note low attenuation window that exists below the K-edge energy

Mo Target – Unfiltered Spectra



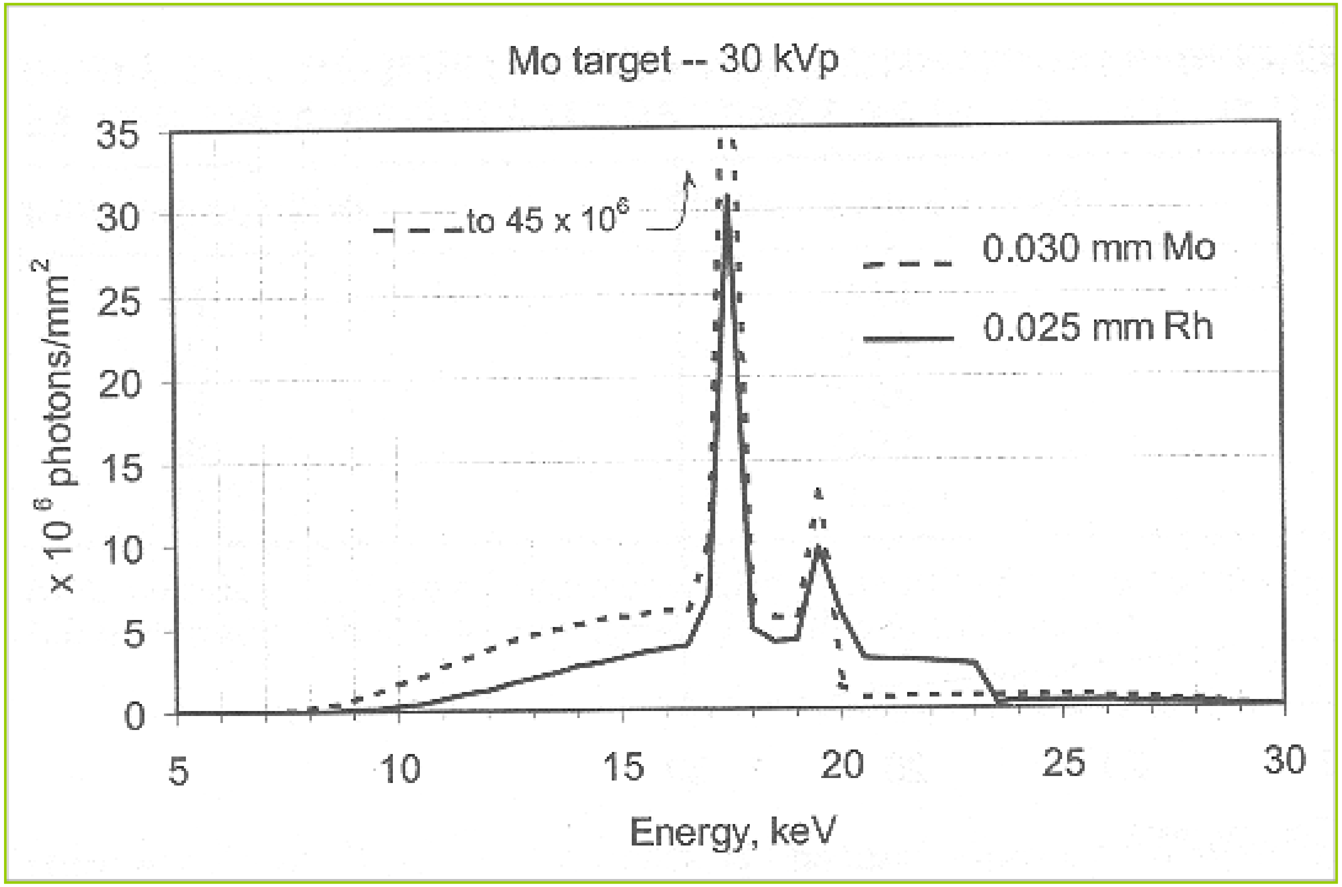
Contain relatively large fraction of very low and very high energy photons

Mo Target – 0.030 mm Mo Filter



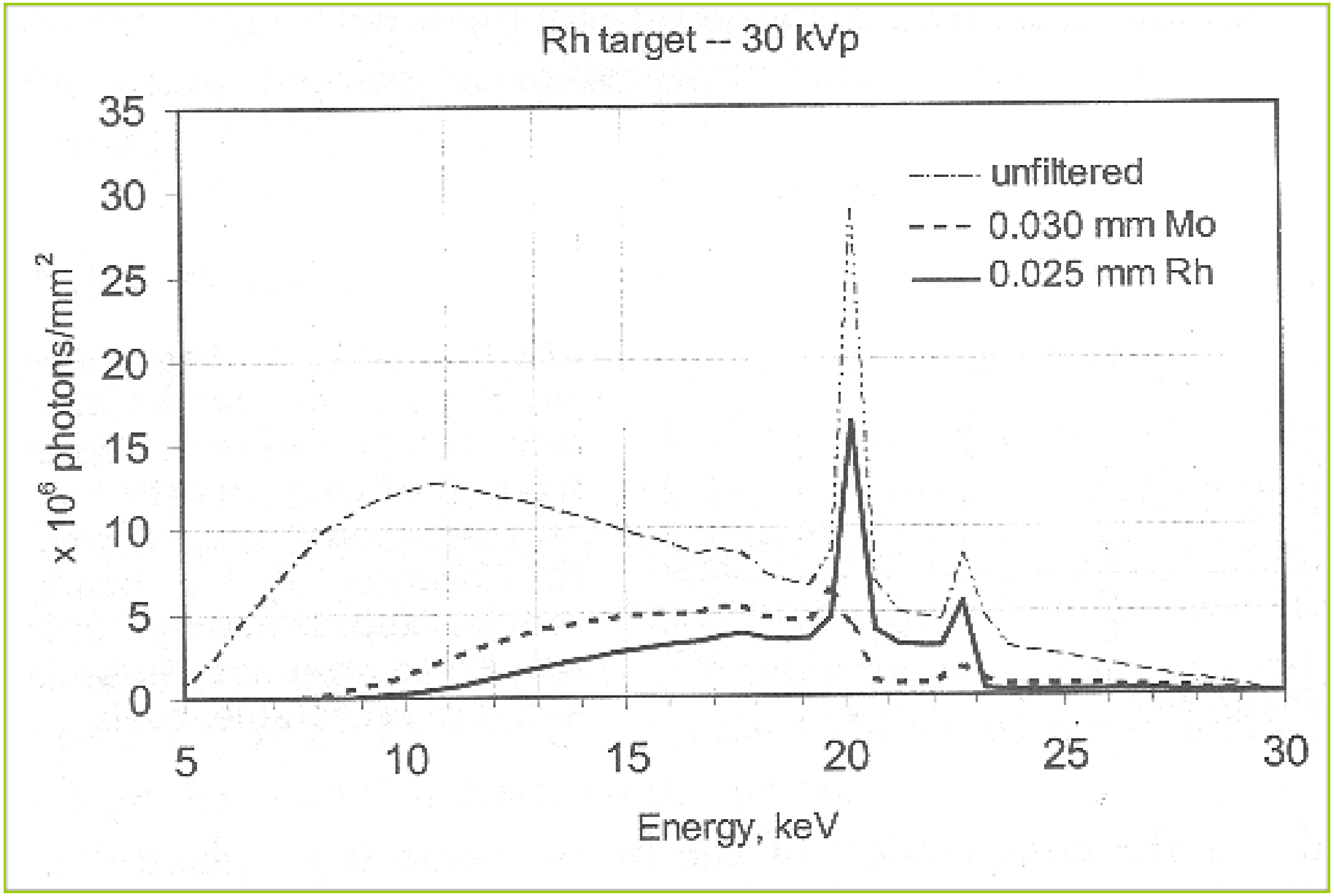
30 μm filter eliminates the majority of low- and high-energy x-rays

Mo Target – Mo & Rh Filters



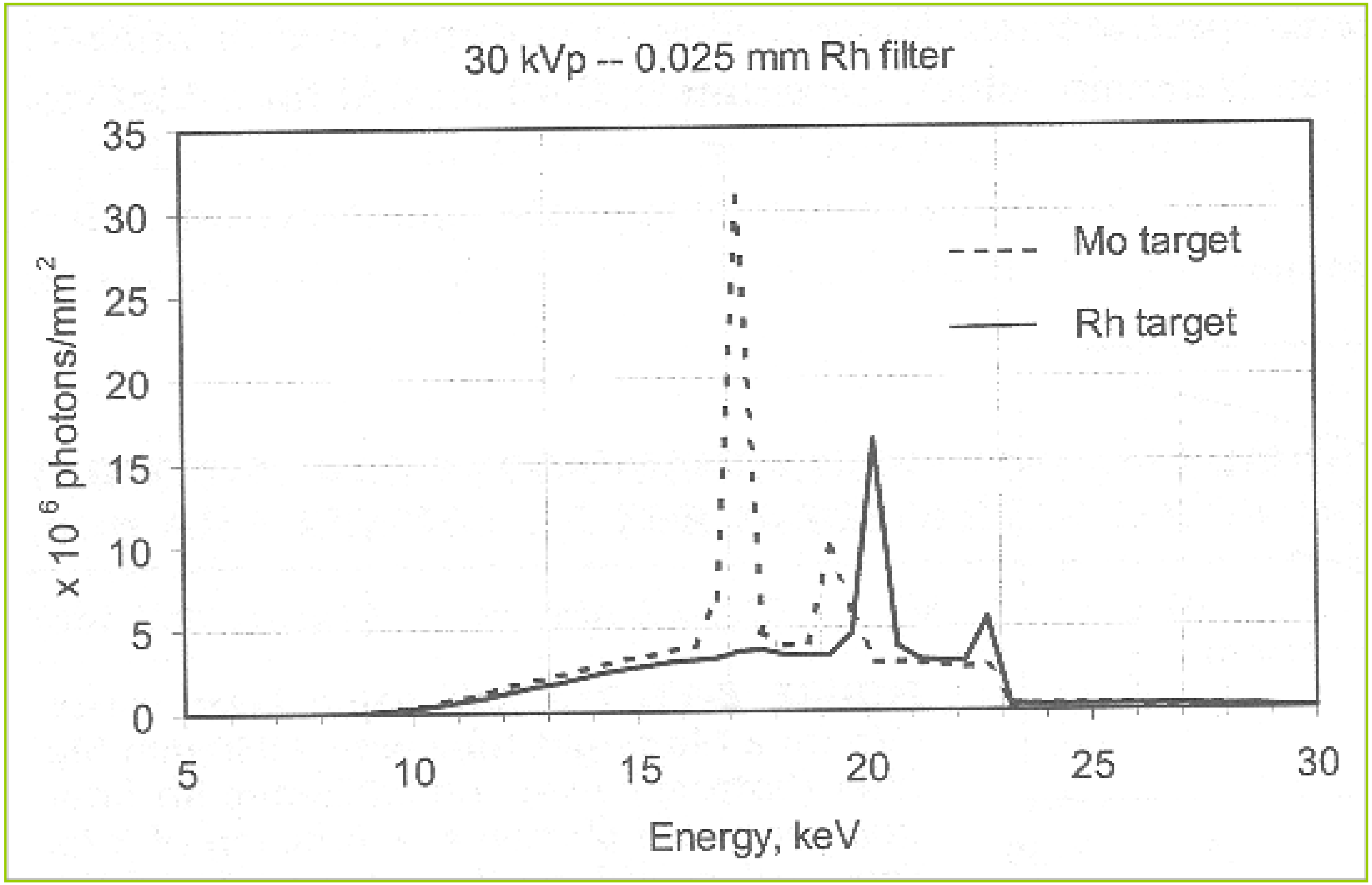
Relative bremsstrahlung photon transmission windows below K-edges

Rh Target – Mo and Rh Filters



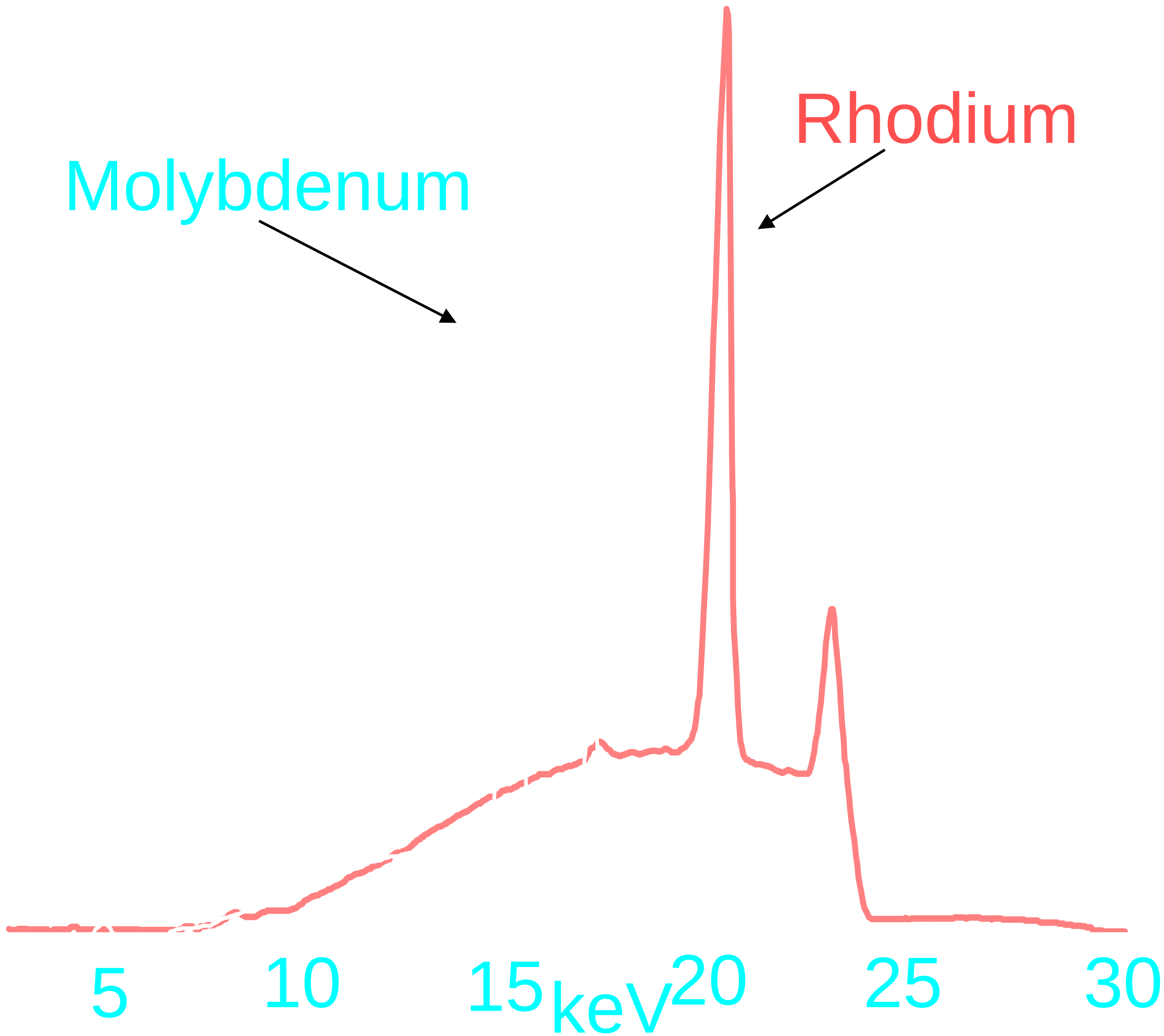
Mo filter with Rh target inappropriately attenuates Rh characteristic radiation

Mo & Rh Target – Rh Filter

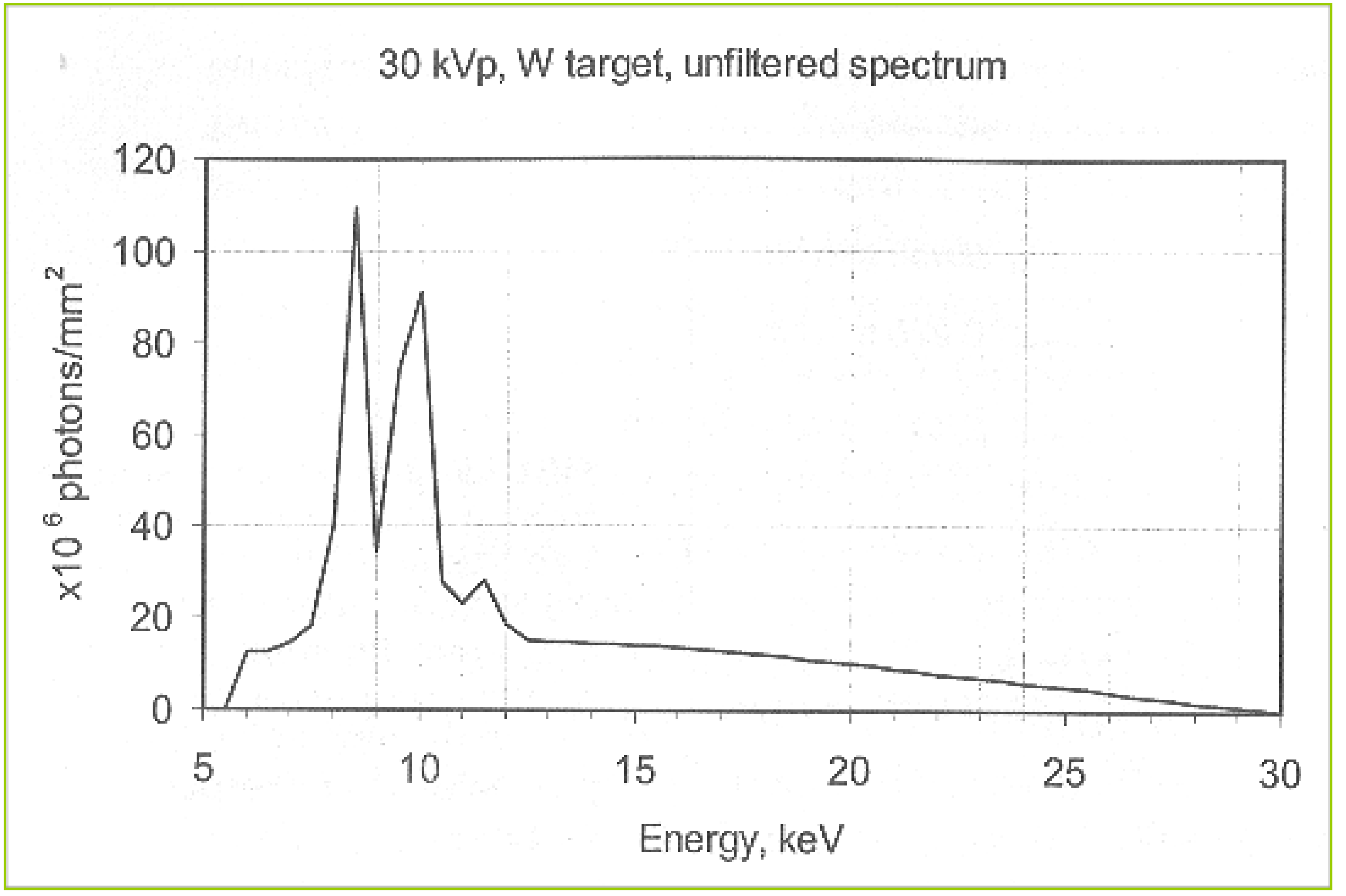


Higher energy x-ray generated by Rh target are better for thick breasts

Dual Target X-ray Tubes

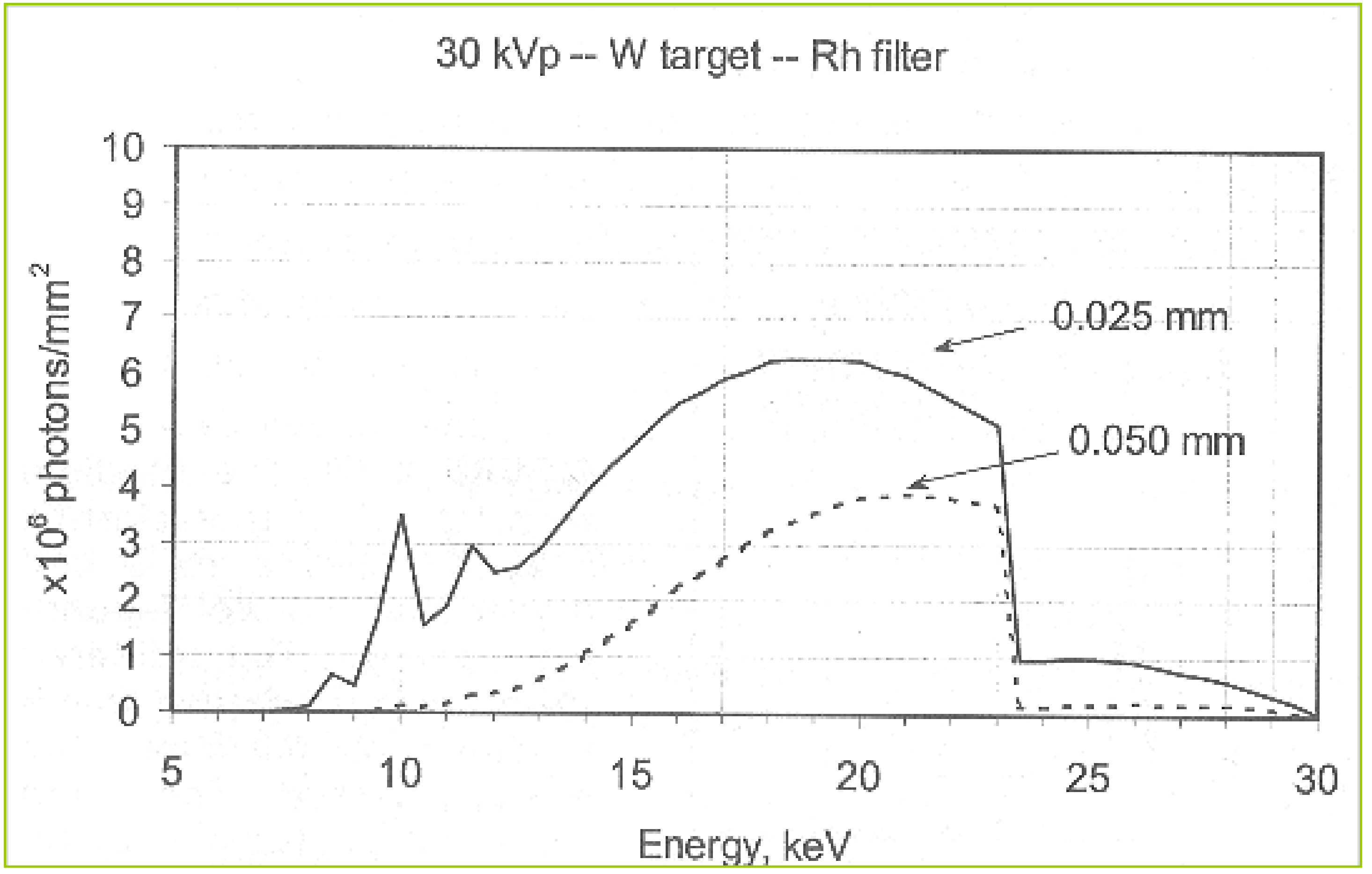


W – Unfiltered Spectrum



Characteristic x-rays between 8 and 10 keV

W Target – Rh Filter



Attenuates L-characterstic radiation to acceptable levels

Mammography X-ray spectra

- Digital mammography tends to use W/Rh combination as digital gives much better contrast than the fixed dynamic range of film
  - Hence, the poorer contrast due to the harder X-ray beam is less significant than the gain due to the use of digital (window and level, image processing, etc)
  - Allows lower doses to be used for the same image quality
  - In practice, contrast resolution in digital is **much** better than film

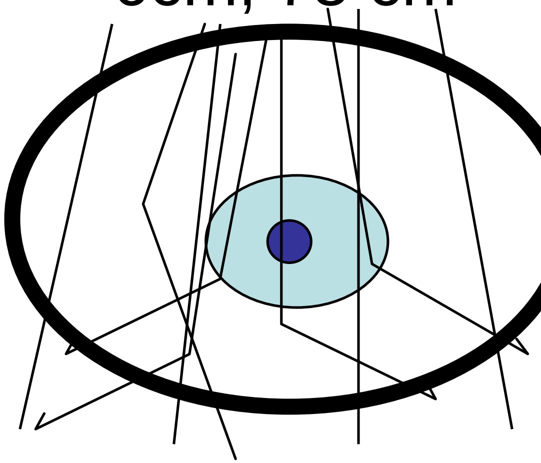
Breast Compression

- Scatter radiation degrades image contrast
- Primary radiation is the useful radiation which creates the image
- The thicker the breast, the more the scattering centers->more scatter

Breast Compression

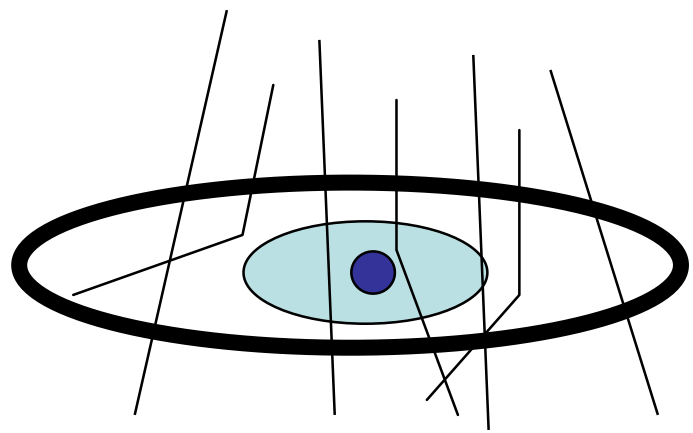
Uncompressed

6cm, 75 cm^2



Compressed

3cm, 150 cm^2



Scatter/Primary=1.0

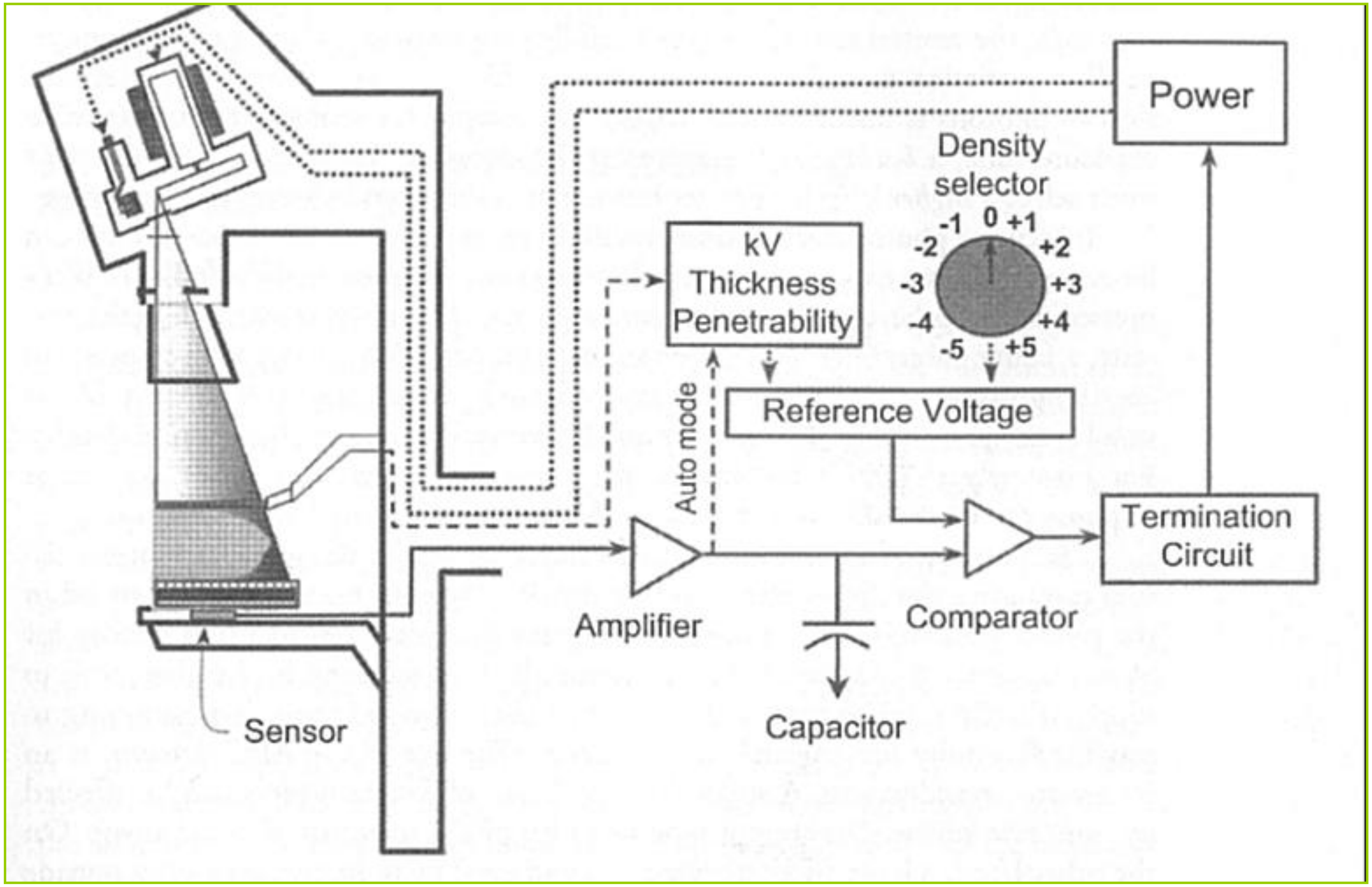
Scatter/Primary=0.40

Improves image contrast by 1.43

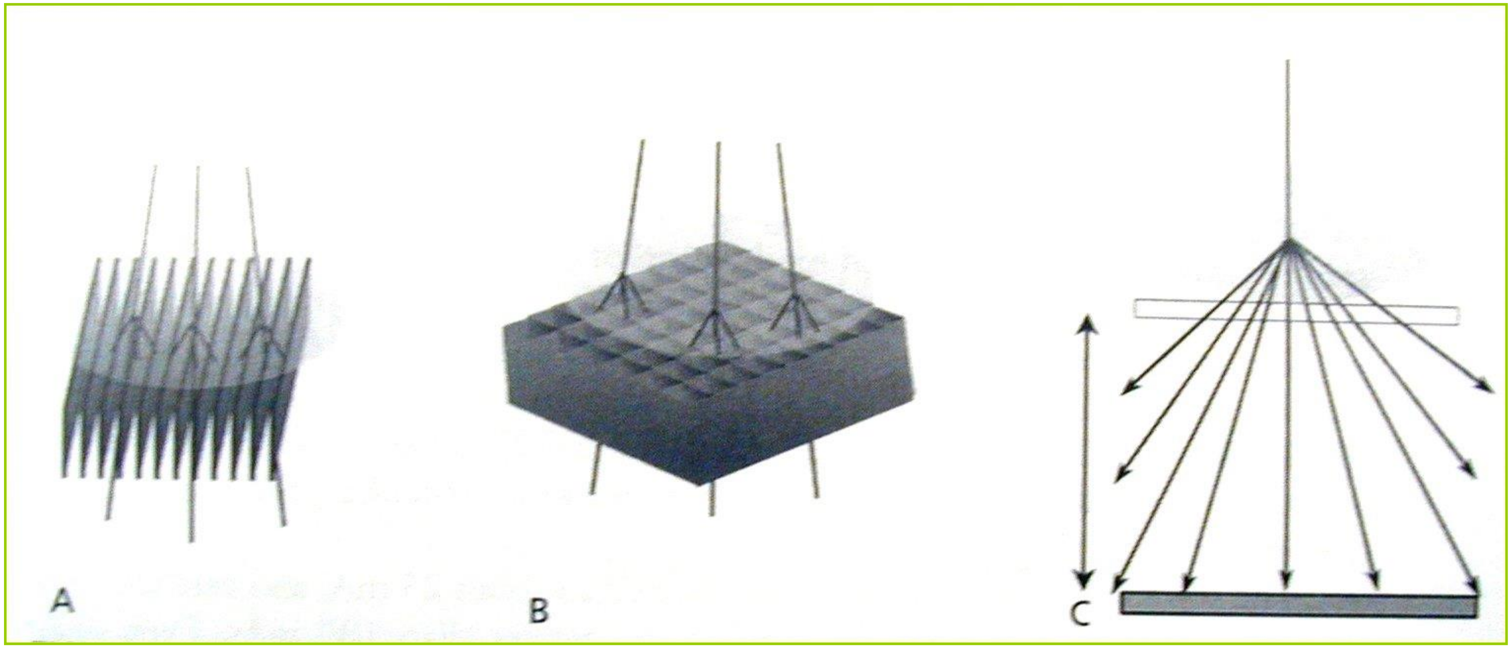
Auto Exposure Control(AEC)

- To have consistent film densities even with various breast thickness and densities
- More importantly, to have film properly exposed to achieve maximum contrast on the screen/film’s HD curve

Automatic Exposure Control



Anti-Scatter Devices

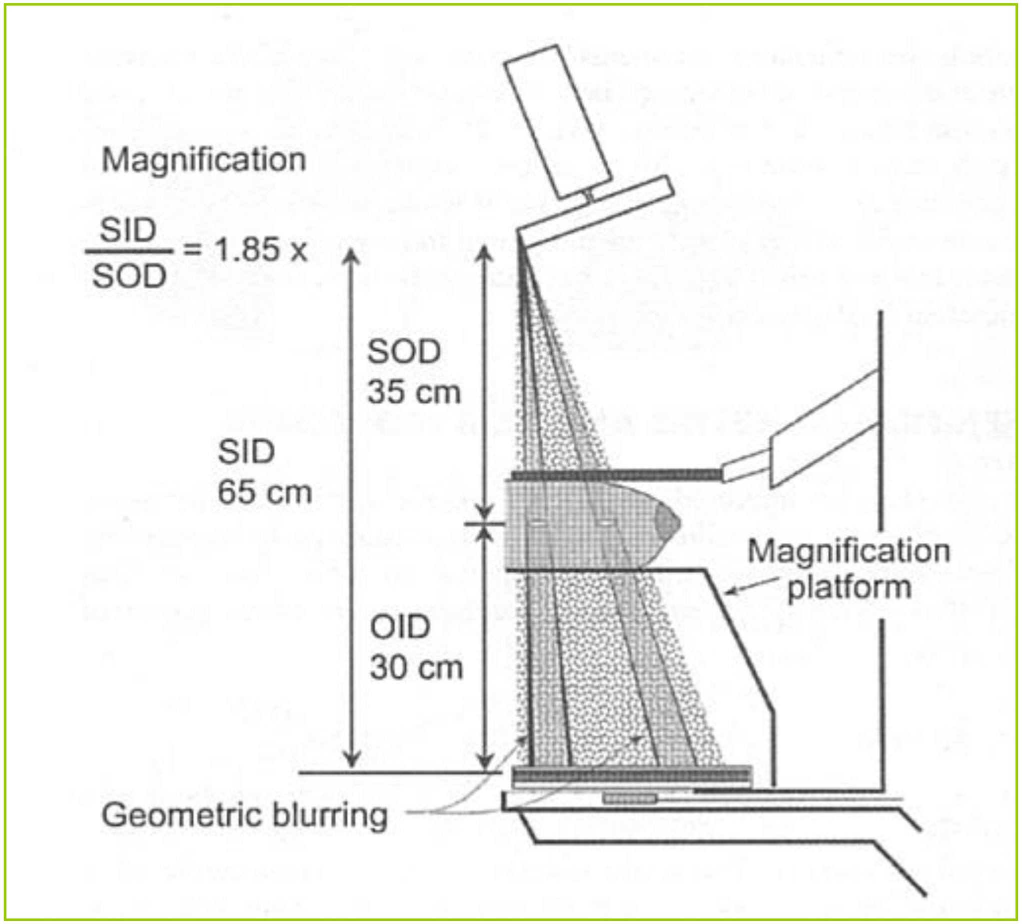


Left: linear grid (5:1 ratio)
Middle: Cellular grid rejects scatter in two dimensions
Right: Magnification procedure

Reciprocating Grids

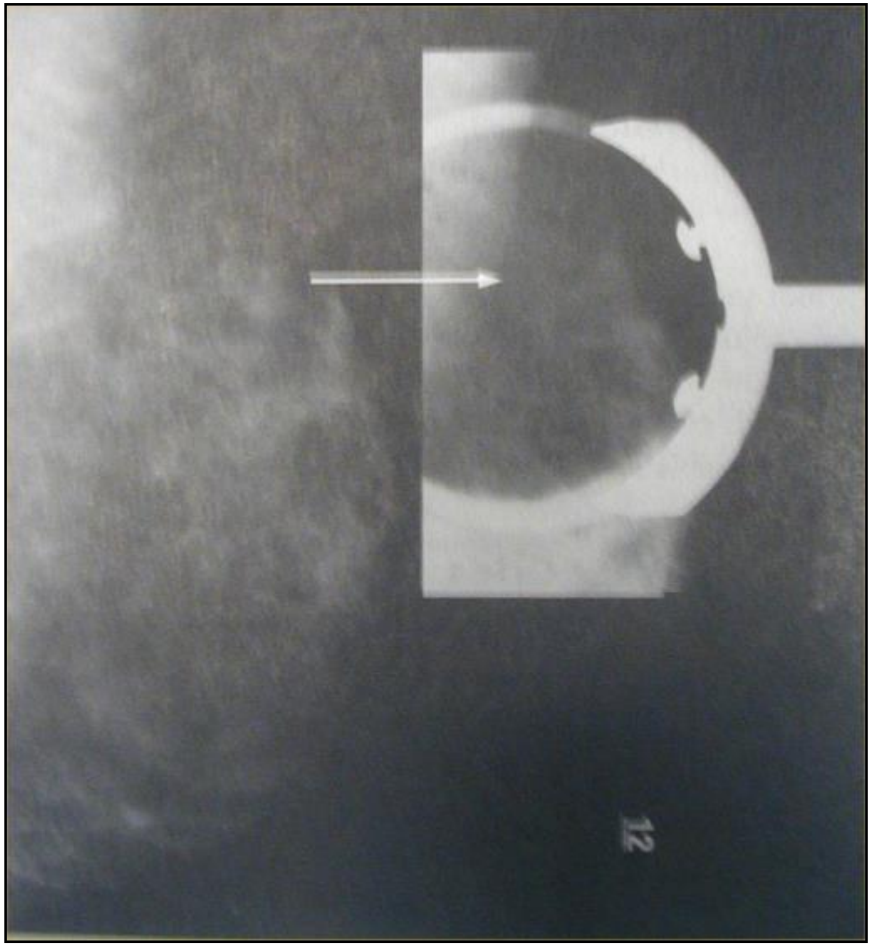
- Only 40~75% of the possible contrast is imaged unless scatter is controlled
- Typical grid ratio of 4~5:1 and 30 to 50 lines/cm
- Most mammo units have moving grid during exposure so that grid line is blurred
- Grids transmit 60~70% of the primary x-ray and absorb 75~85% of the scatter
- We pay "dose penalty" to achieve contrast improvement

Magnification Technique



- Geometric magnification uses a support platform giving 1.5 x to 2x magnification.
- Small focal spot used
- Best resolution on anode side

“Spot” Compression for Diagnostic Workup



- Spreads tissues over a localized area
- Used extensively with magnification radiography

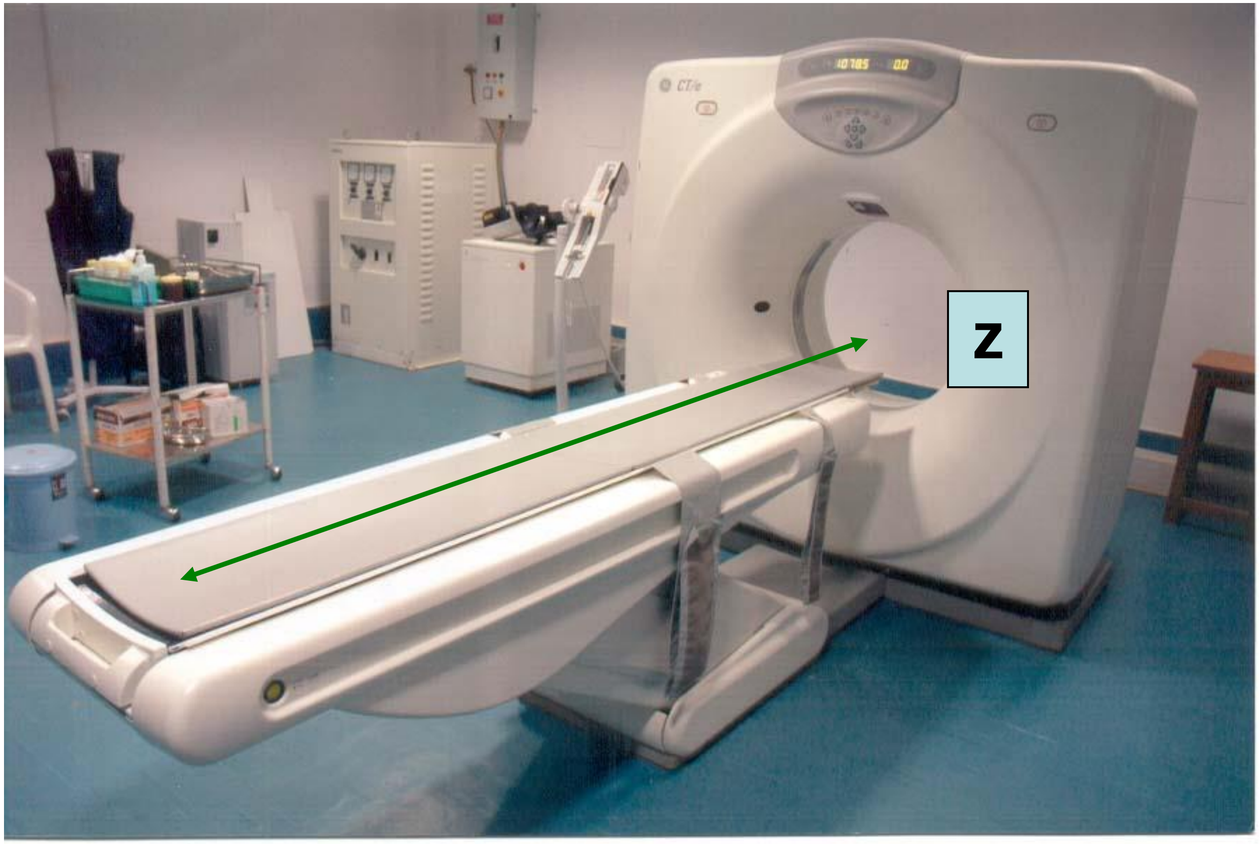
Computed Tomography

Why CT?

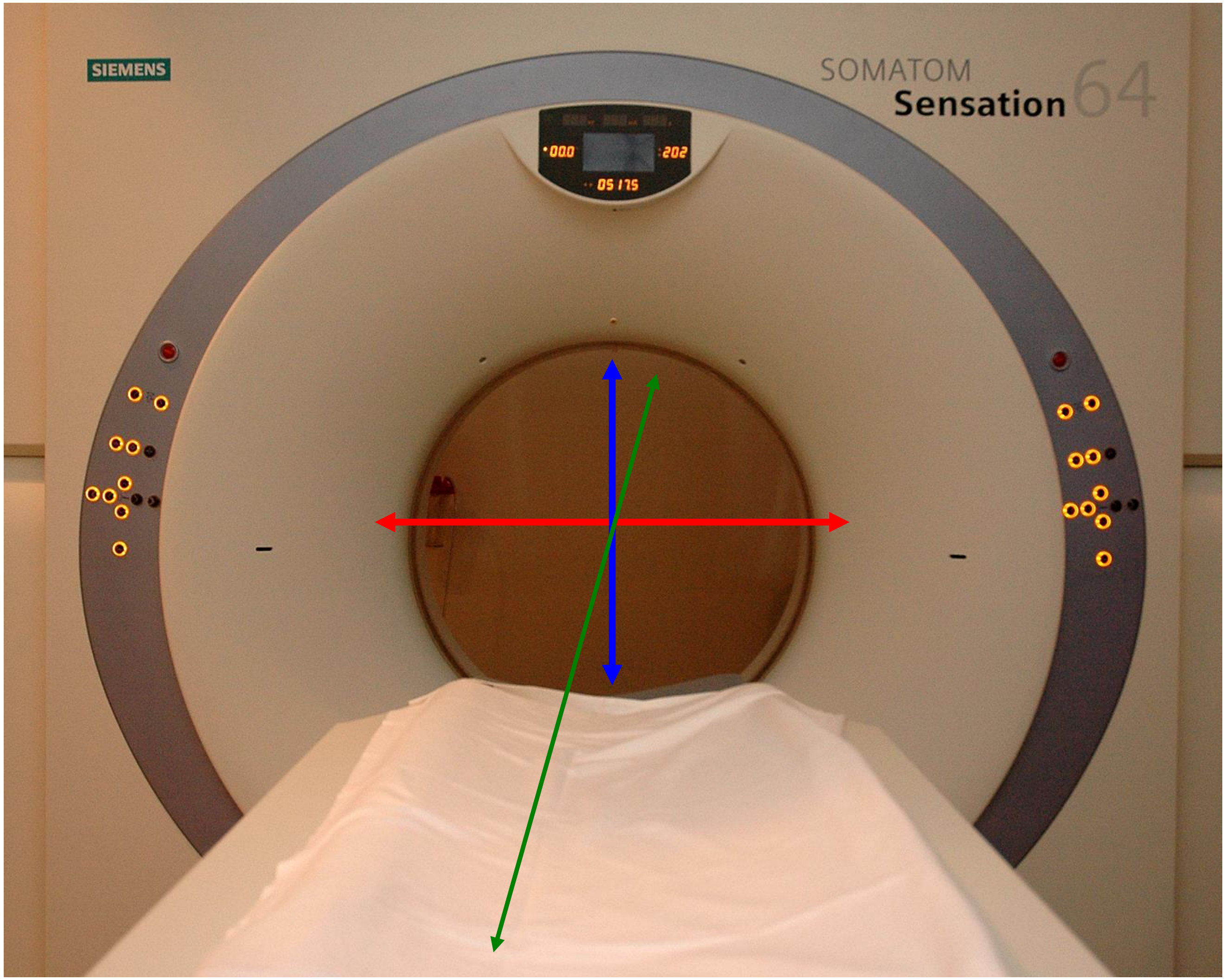
- Fast
- Multi plane imaging
- Good spatial resolution
- Good temporal resolution
- Quantitative possible



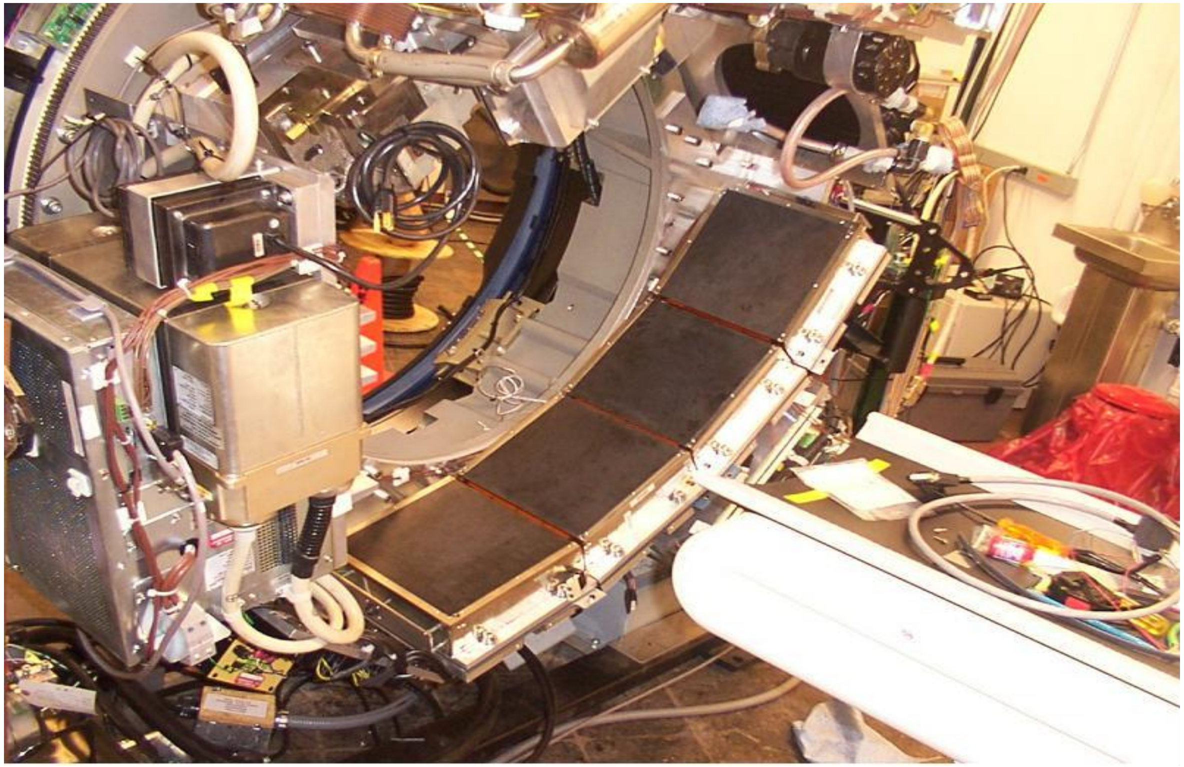

COORDINATE SYSTEM



ISOCENTER



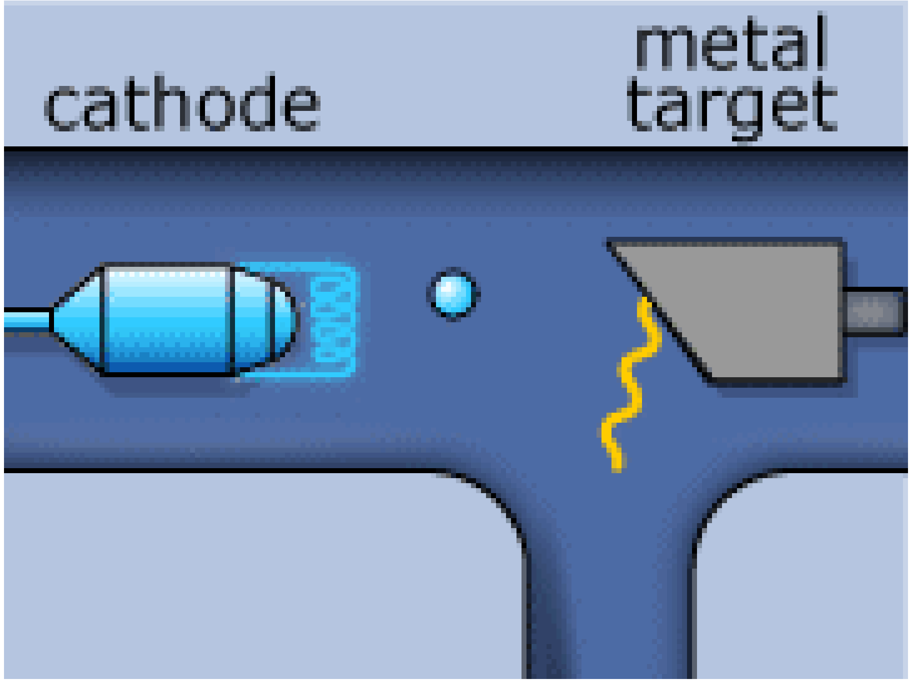
Inside CT scanner



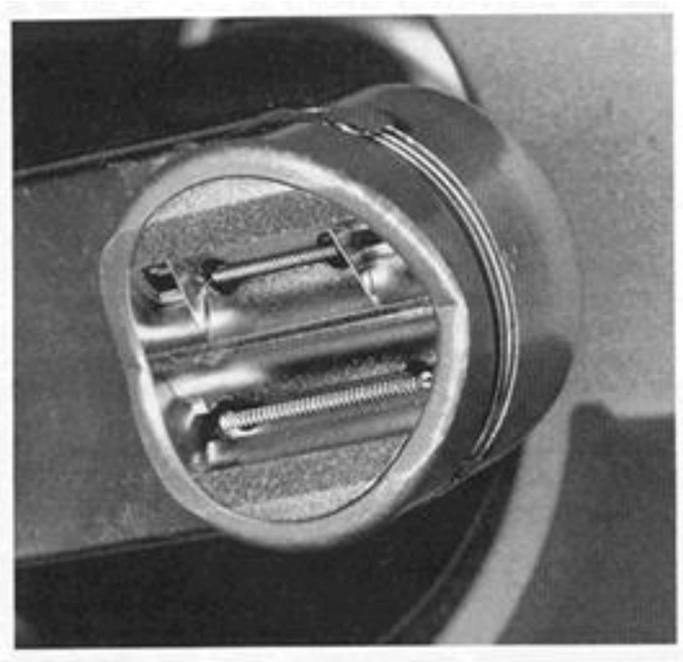
IMAGING SYSTEM COMPONENTS

- X-RAY TUBE
- GENERATOR –HIGH VOLTAG
- COLLIMATORS
- FILTER
- DETECTORS
- DETECTOR ELECTRONICS

X-RAY TUBE AND X-RAY PRODUCTION

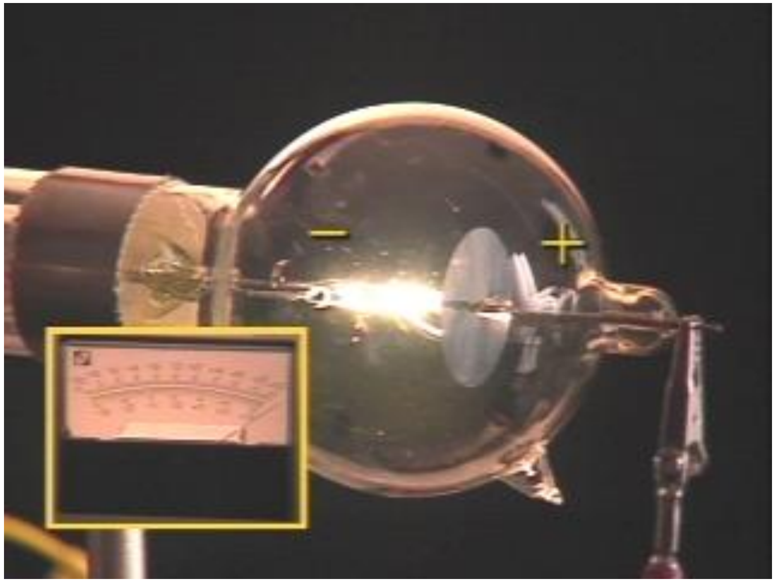


CATHODE --------- MADE OF TUNGSTEN



IN CT – STILL SMALL AND LARGE

THERMIONIC EMISSION



CATHODE HEATED UP TO AT LEAST 2,200 DEG. CELSIUS TO LIBERATE ELECTRONS FOR TRANSIT TO ANODE

FOCAL SPOT- CT UTILIZES DIFFERENT FOCAL SPOTS

- THE FILAMENT SIZE – LENGTH – **FOCAL SPOT**

SMALLER FOCAL SPOT - **Low mA**

SMALLER FOCAL SPOT – **sharper image**

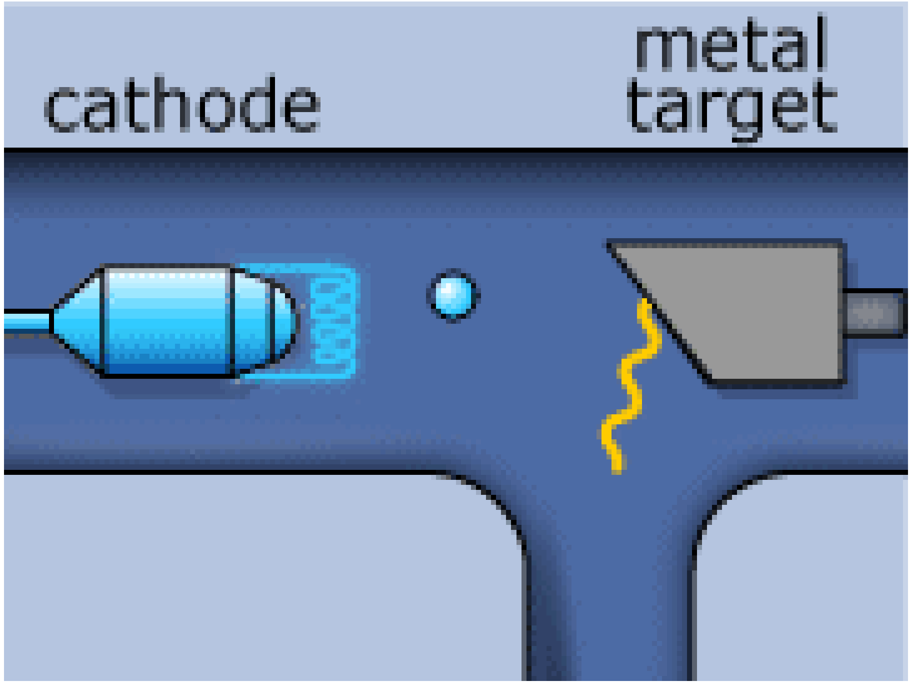
ANODE +++++ MADE OF TUNGSTEN AND MOLYBDENUM

TUNGSTEN TARGET

TARGET MADE OF TUNGSTEN
AND
RHENIUM

mA – tube current

- The number of electrons flowing from cathode to anode



kVp

- Potential difference between cathode and anode (Volts) kilo means 1,000 x.

S –time of exposure

mAs tube current for certain length of time

X-RAY PRODUCTION RESULTS IN A LOT OF HEAT AND VERY LITTLE X-RAYS BEING GENERATED

HEAT UNITS CALCULATION

HU= kVp X mA x time

MOST CT TUBES HEAT CAPACITY
3-5 MILLION HU

TUBE CURRENT CHANGE

CURRENT ⬆

➡

INTENSITY ⬆

ENERGY – NO CHANGE

2 * mA = 2 * number of photons

4 * mA = 4 * number of photons

Why changing mA or time

- Avoiding motion – mA ↑ time ↓
- Pediatric technique modification
- Reducing noise - mAs ↑

NOISE

MOTION

Tube voltage (kVp) CHANGE

kVp ↑

→

INTENSITY - ↑

ENERGY – ↑

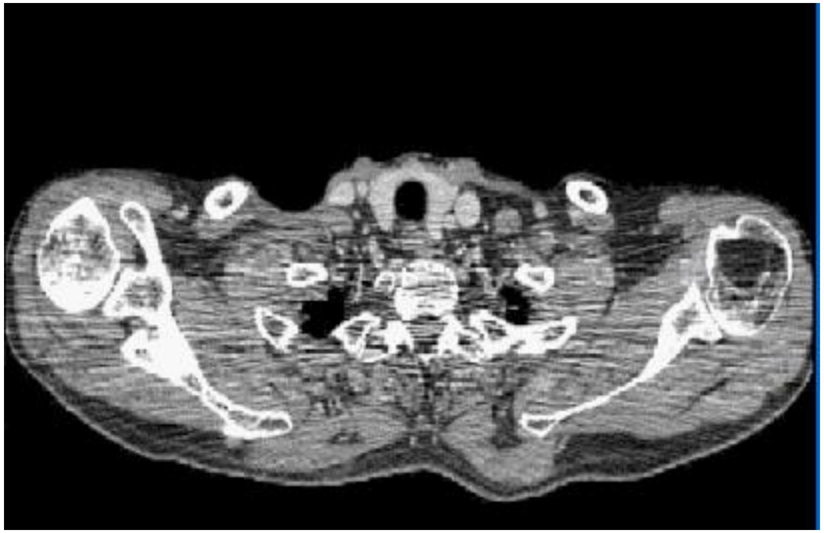
15% INCREASE OF KVP = 2 * mAs

kVp IN CT

- 80-140
- TOO LOW – NOISE

(NOT ENOUGH PENETRATION OF THE PATIENT)

PHOTON STARVATION - NOISE!!!!!



HIGH VOLTAGE GENERATOR – (HVG)

- GENERATES HIGH VOLTAGE POTENTIAL BETWEEN CATHODE AND ANODE OF AN X-RAY TUBE

CT GENERATOR

- 5-50 kHz
- 30-60 kW

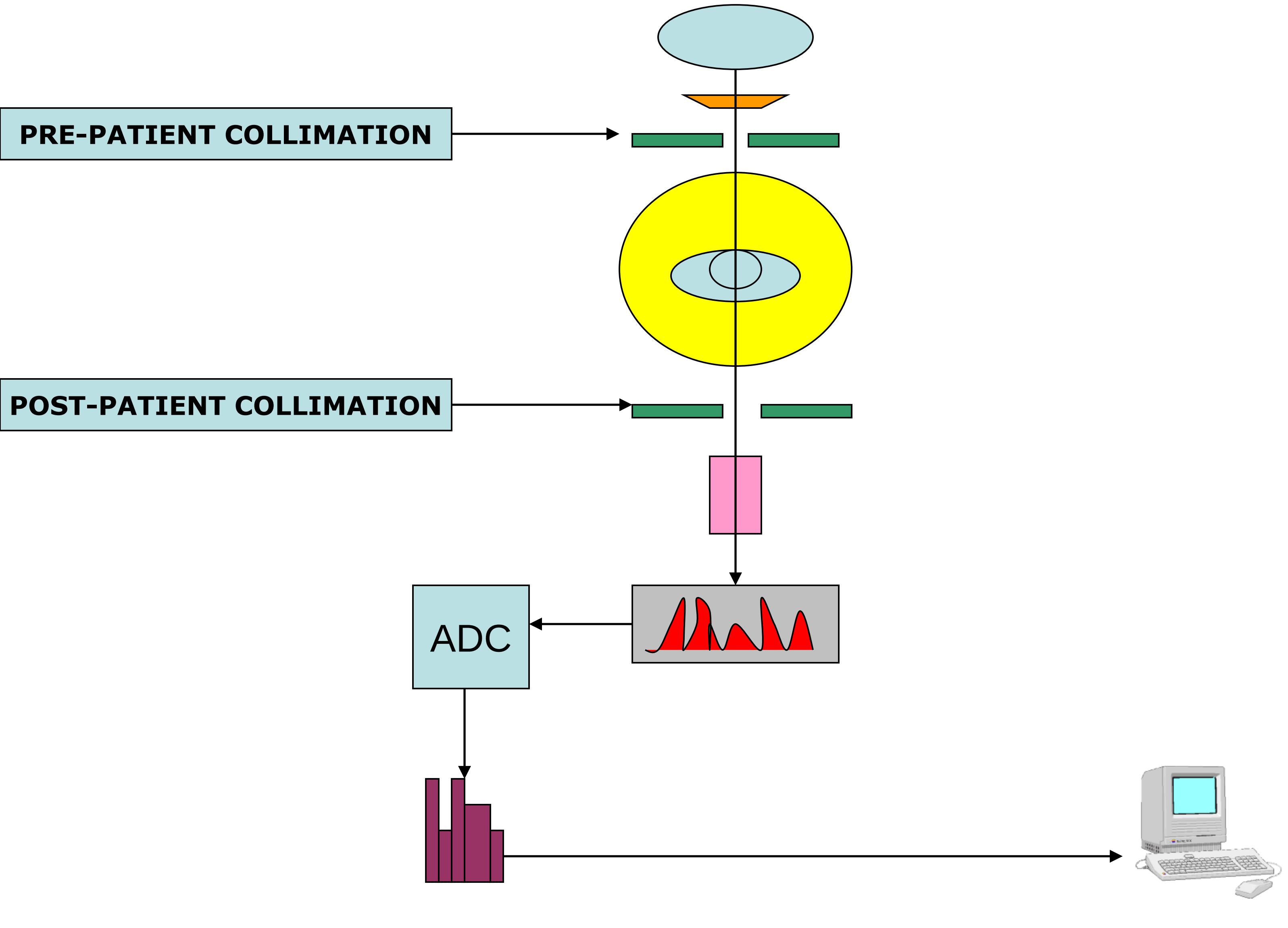
KVP SELECTION:

80, 100, 120, 130,140

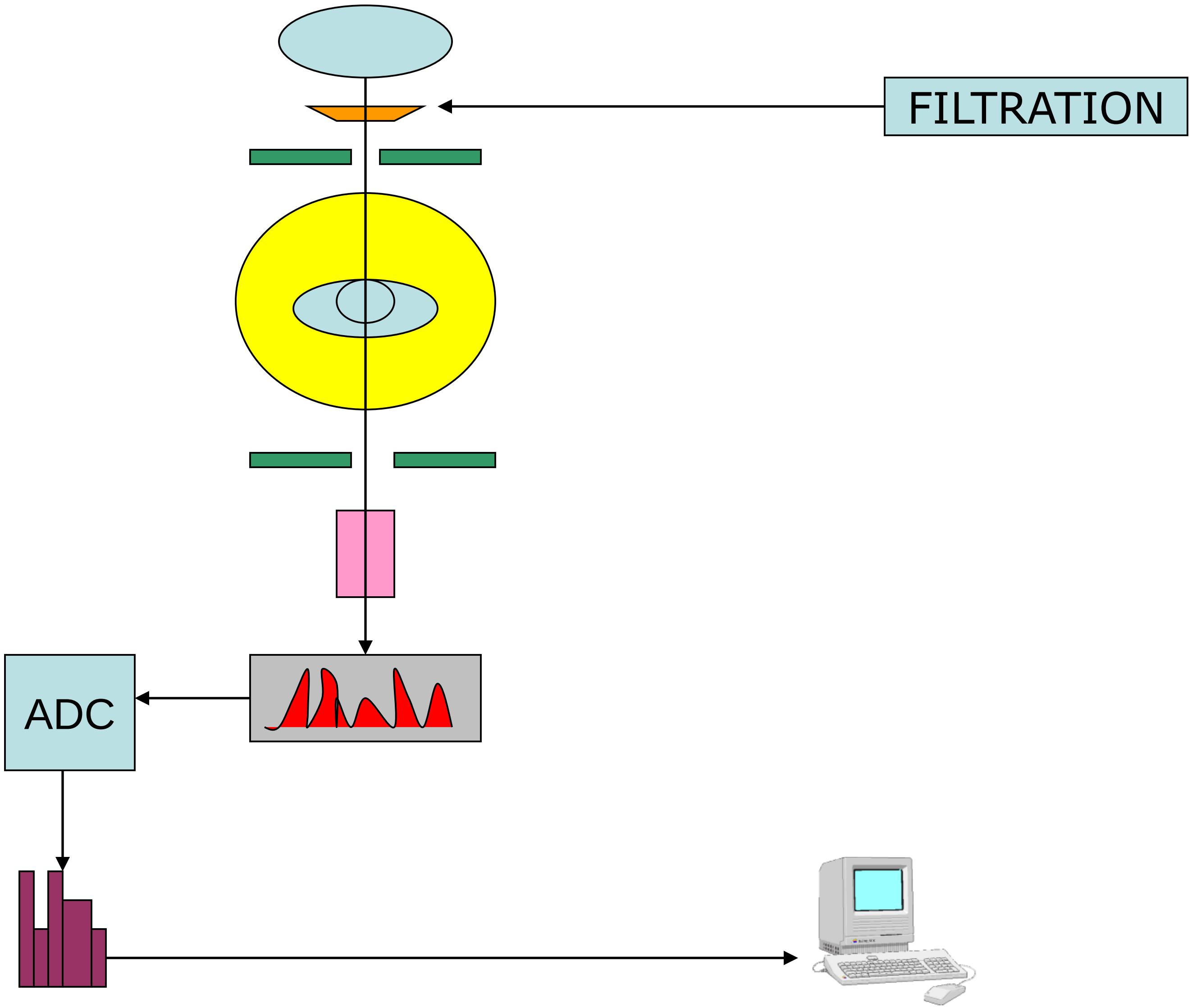
mA selection:

30, 50, 65, 100, 125, 150, 175, 200, 400

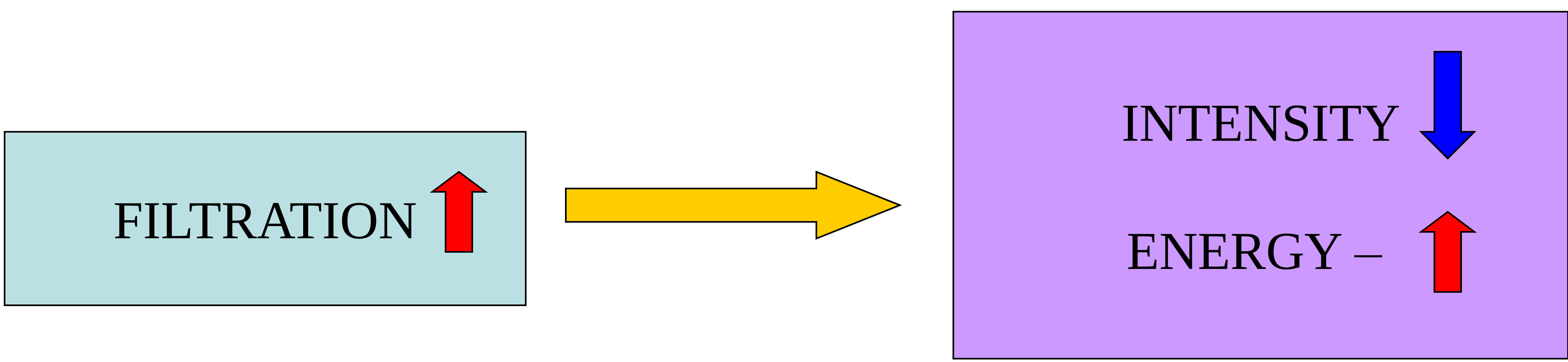
COLLIMATION IN CT



BASIC DATA AQUSITION SCHEME IN CT



FILTRATION CHANGE

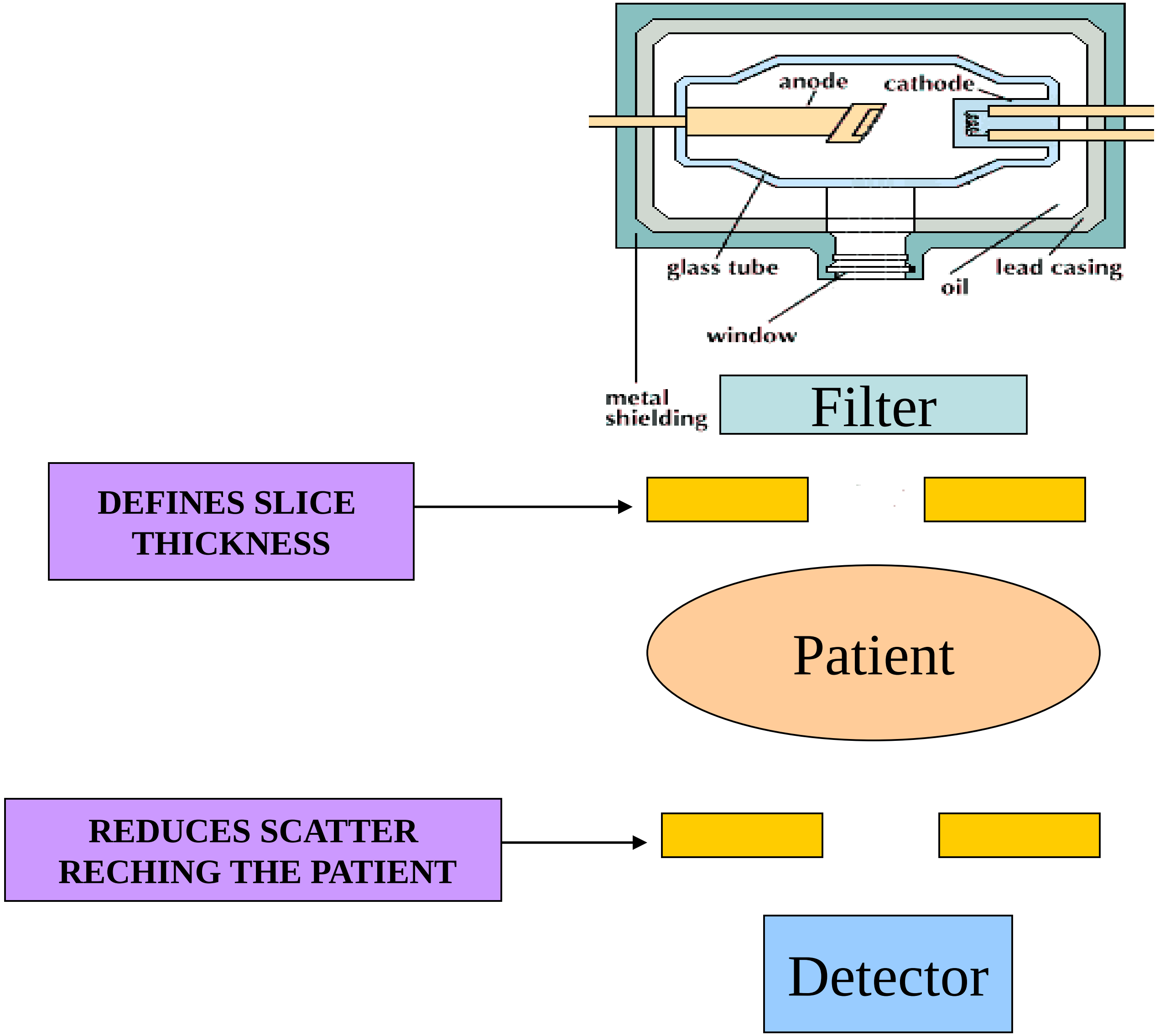


FILTRATION MATERIAL

- ALUMINIUM (SPECIAL FILTER IN CT)

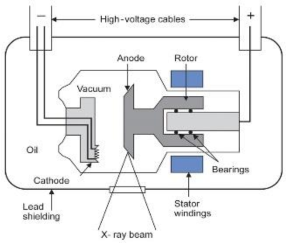
BOWTIE

TO MAKE THE BEAM ***HARDER*** AND MORE ***MONOENERGETIC***

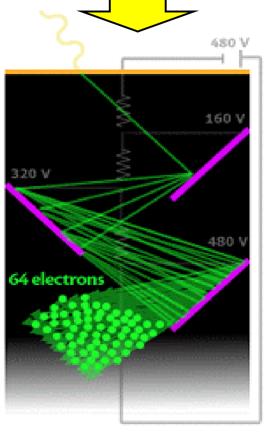

anode
cathode
glass tube
oil
lead casing
window
metal shielding
Filter
DEFINES SLICE THICKNESS
Patient
REDUCES SCATTER RECHING THE PATIENT
Detector

CT DETECTORS

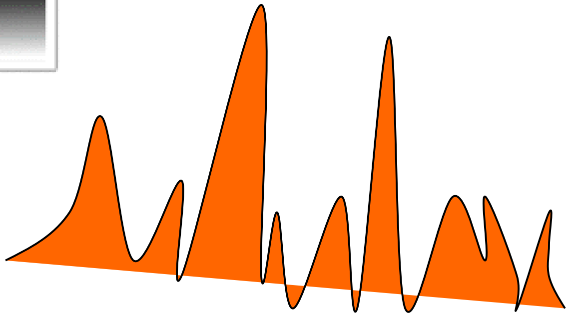
DETECTOR TYPES: SCINTILLATION

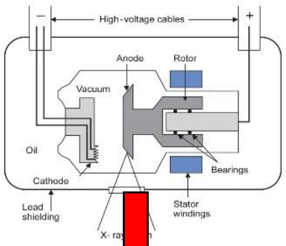


S. CRYSTAL



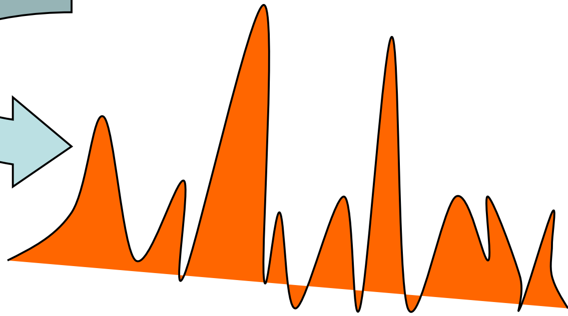
PM TUBE





S. CRYSTAL

PHOTODIODE



SCINTILLATION CRYSTALS USED WITH PM TUBES:

- SODIUM IODIDE –AFTERGLOW + LOW DYNAMIC RANGE **(USED IN THE PAST)**

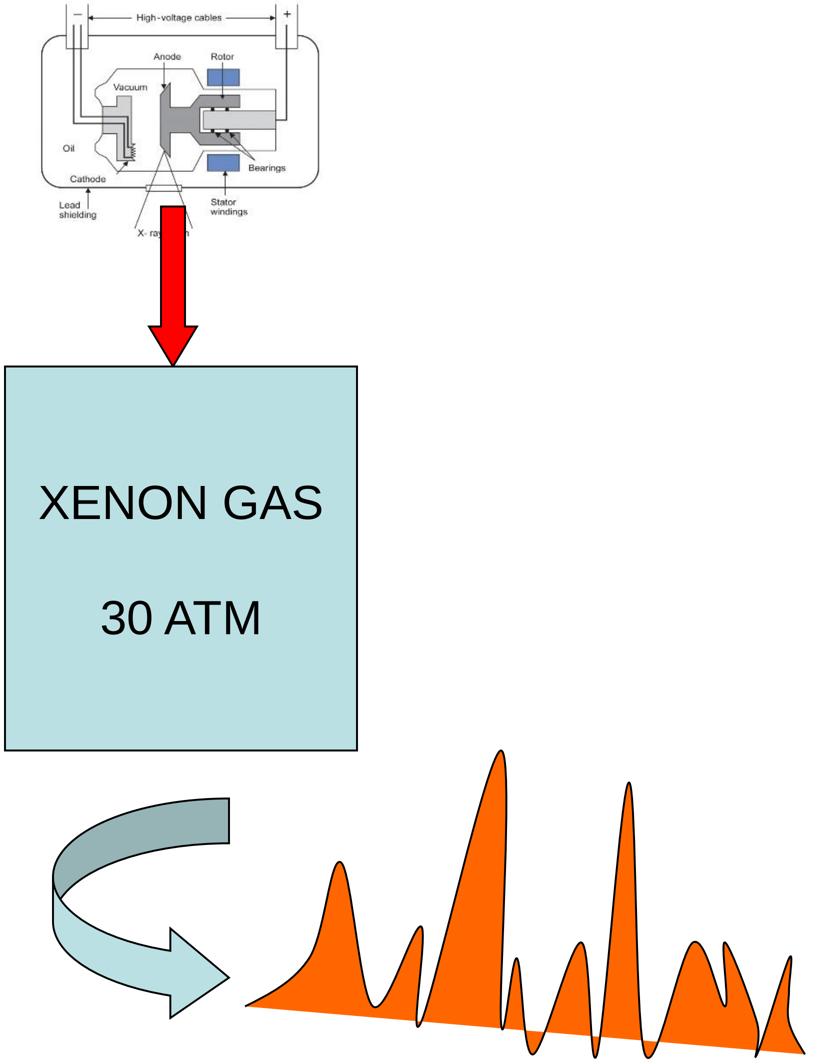
- CALCIUM FLUORIDE

- BISMUTH GERMANATE

S. CRYSTAL USED WITH PHOTODIODE

- CALCIUM TUNGSTATE
- RARE EARTH OXIDES - CERAMIC

DETECTOR TYPE: GAS IONIZATION



EFFICIENCY OF DETECTORS- QDE

- SCINTILLATION – 95% - 100%- COMMONLY USED IN III & IV GENERATION SCANNERS

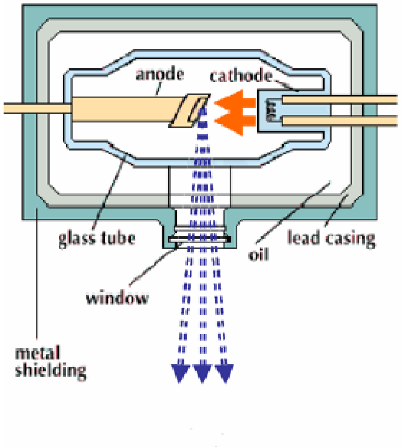
- GAS – 50% - 60%

COMPUTER SYSTEM

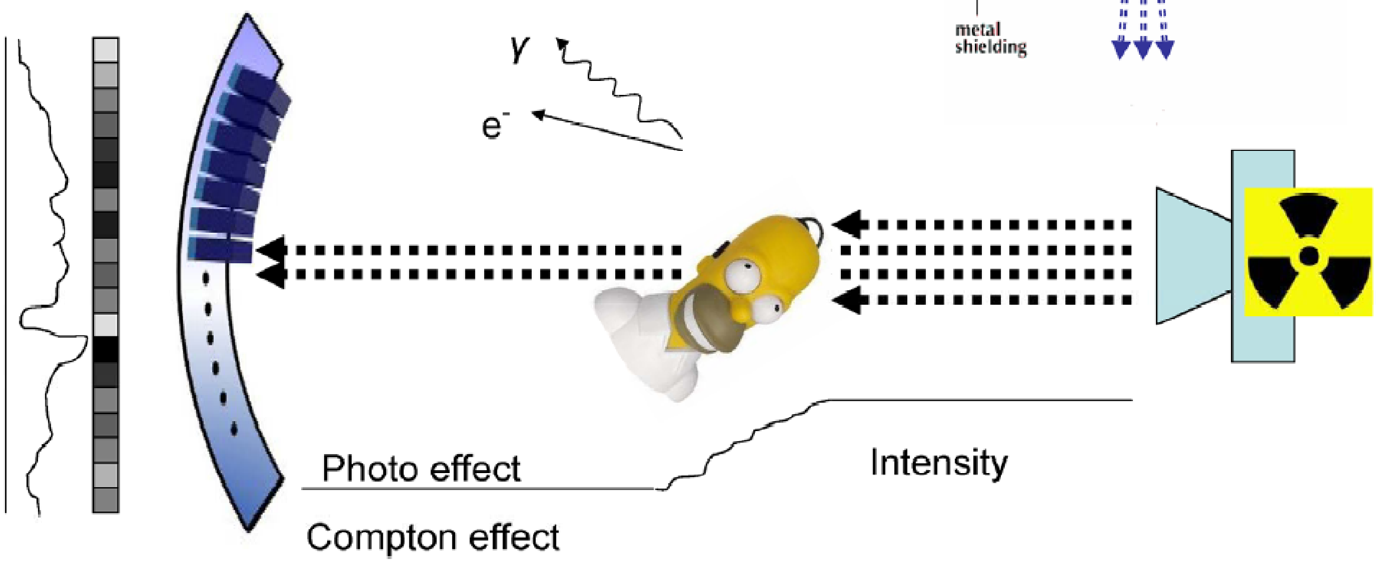
- RECONSTRUCTION AND POSTPROCESSING
- CONTROL OF ALL SCANNER COMPONENTS
- CONTROL OF DATA ACQUSITION, PROCESSING, DISPLAY.
- DATA FLOW DIRECTION

CT Data Acquisition and Image Reconstruction

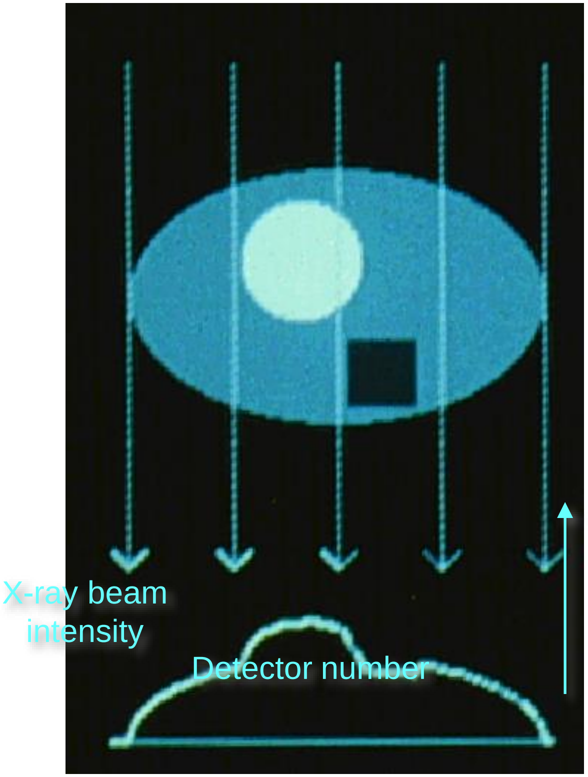
X-rays?



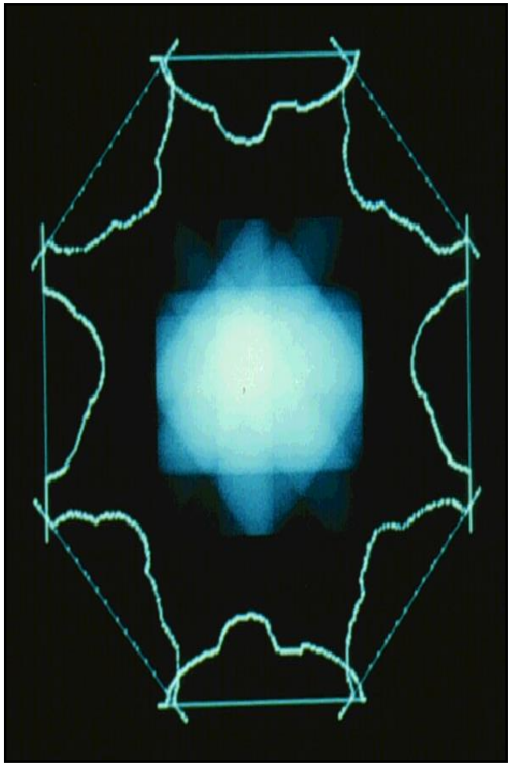
How do they interact with the object?



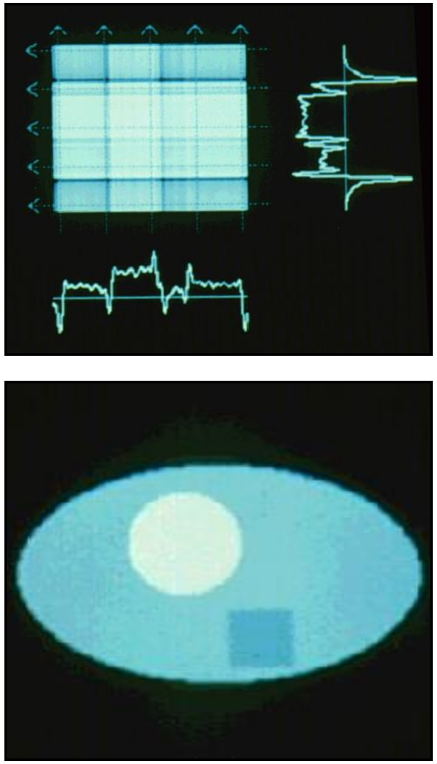
CT Data Acquisition and Image Reconstruction



A CT Projection



Simple Back-projection



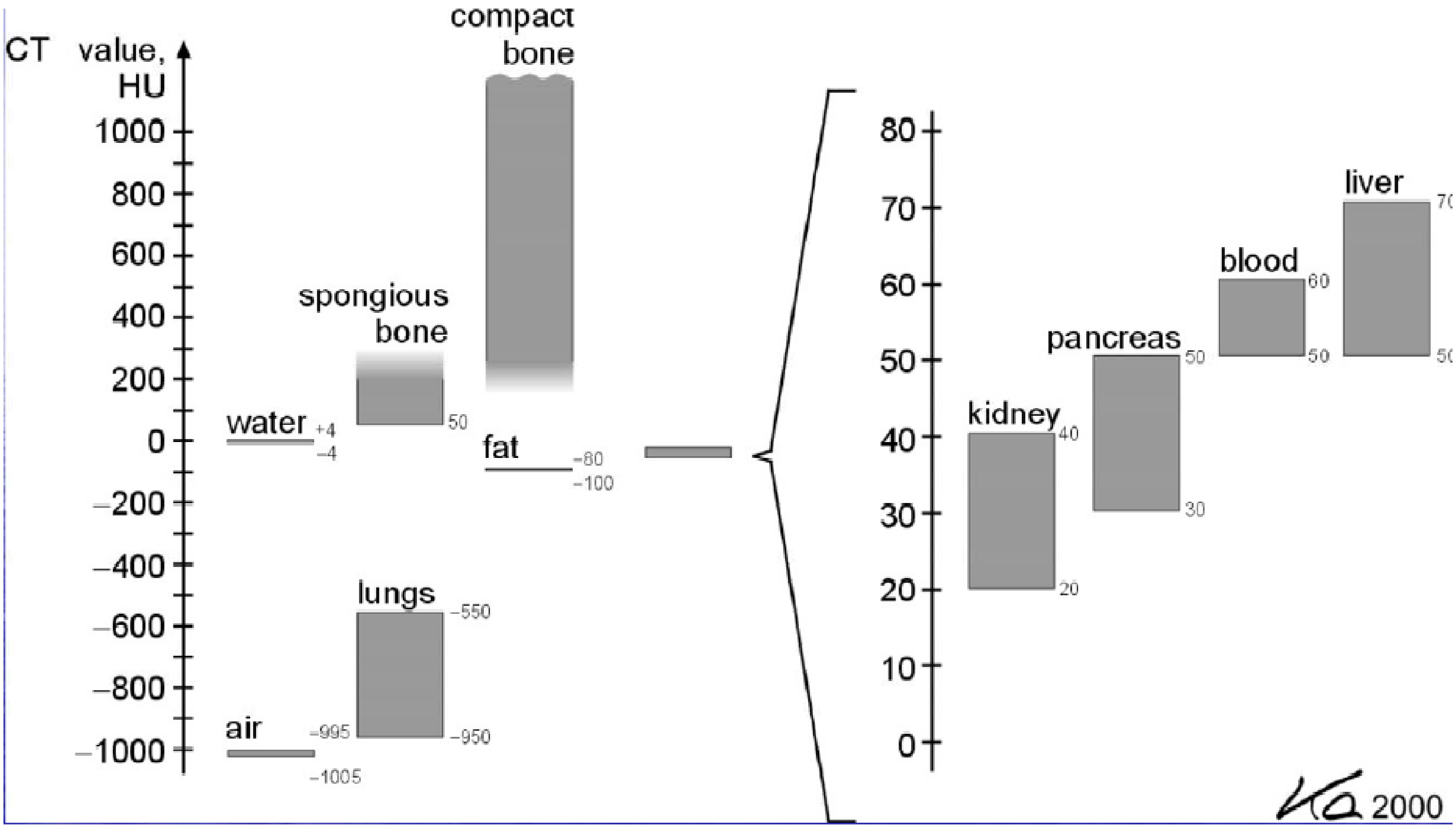
Filtered Back-projection

Computation of The CT Number

$$\text{CT Number} = \frac{K(\mu_{material} - \mu_{water})}{\mu_{water}}$$

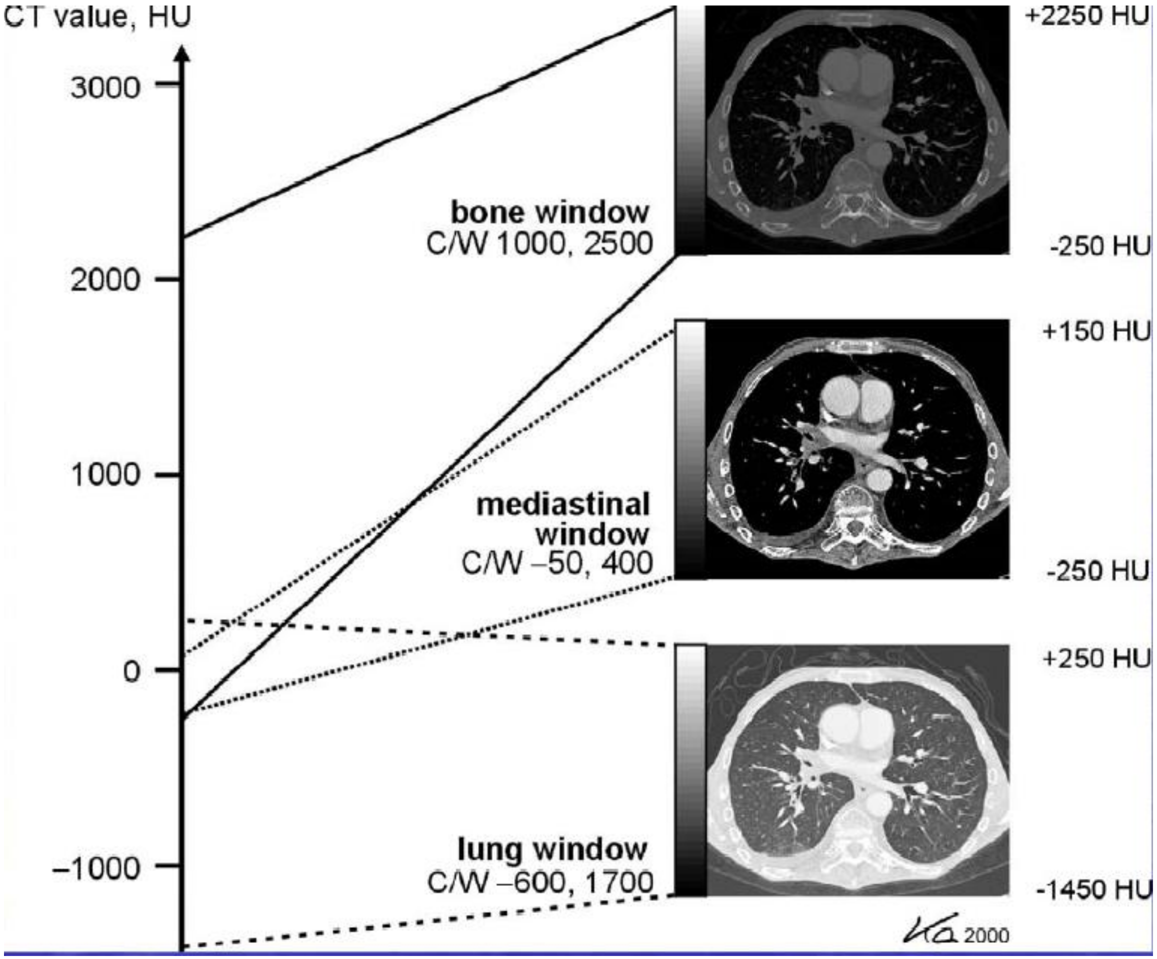
- K is a constant – the reference value for water
- One system is the Hounsfield system where K = 1000.
  - Another is the EMI where K=500

CT Number

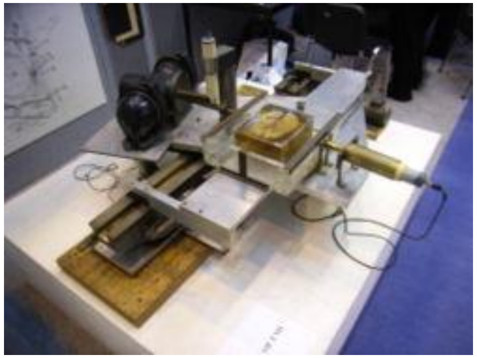


CT value, HU
1000
800
600
400
200
0
−200
−400
−600
−800
−1000
compact bone
spongious bone
50
water
+4
−4
fat
−80
−100
lungs
−550
−950
air
−995
−1005
80
70
60
50
40
30
20
10
0
kidney
40
20
pancreas
50
30
blood
60
50
liver
2000

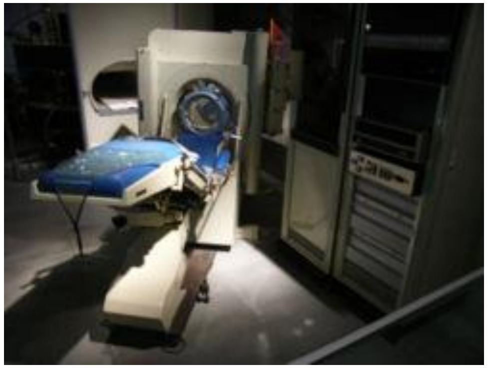
Display Window/Level



From Conventional to MDCT



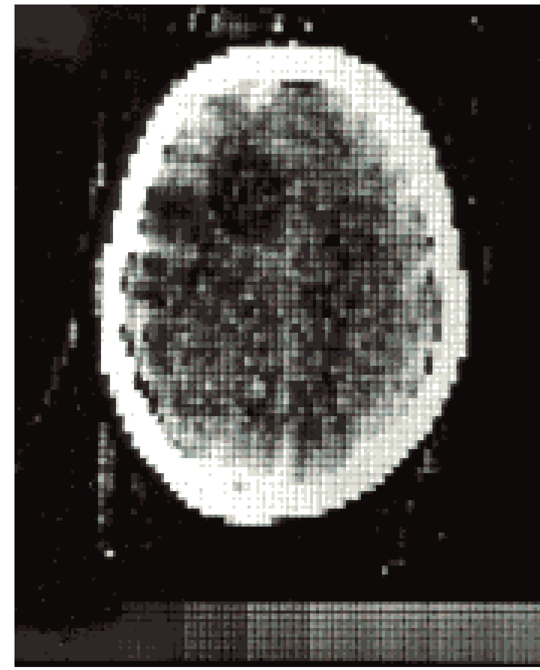
Prototype



First CT scanner

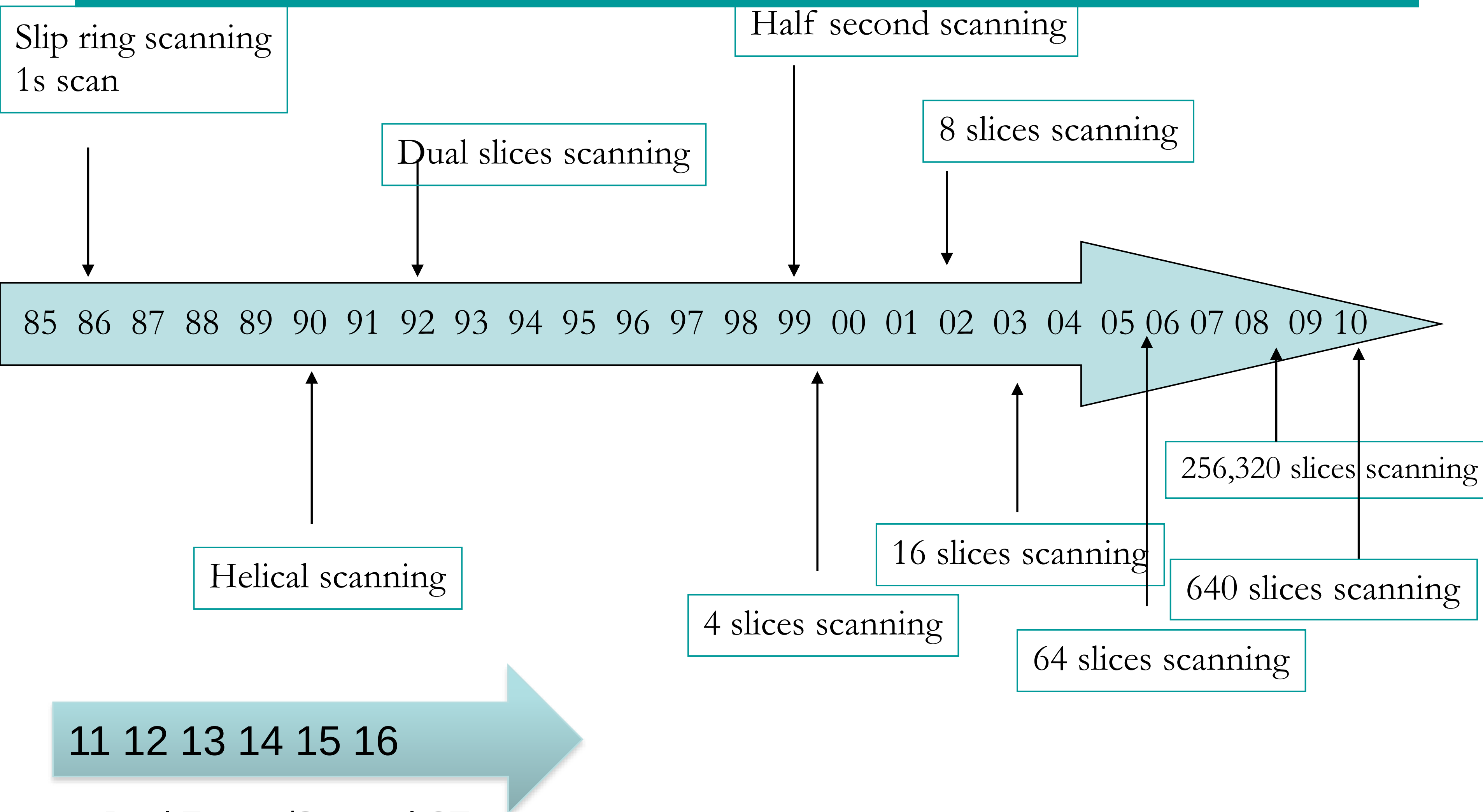
2008 Technology

First CT Scan (1972)



80 x 80 matrix size, 4 min/rotation, 8 grey level, overnight reconstruction

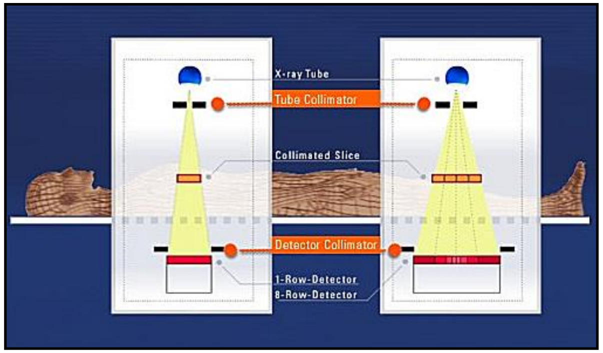
Technological advances in CT

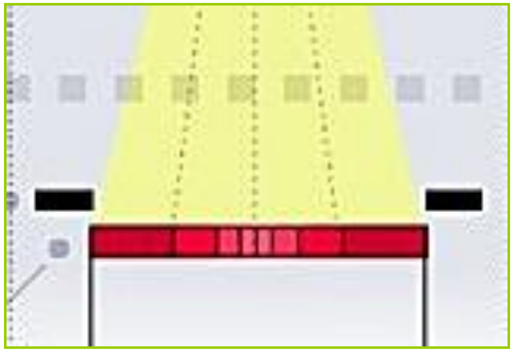


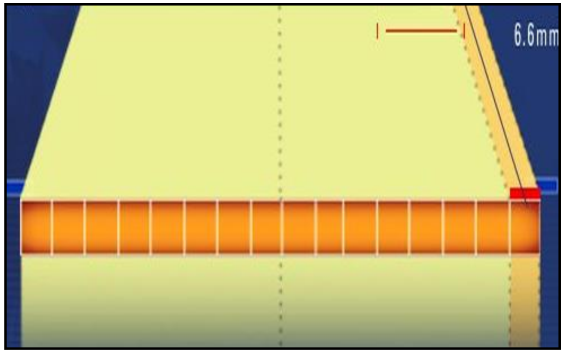
Dual Energy/Spectral CT
Low dose CT

Generations	source	Source collimation	detector	Detector collimation	Source-Detector movement	Advantages	Disadvantages
1st Gen.	single	Pencil beam	single	no	Trans.+ Rotates	No scatter	slow
2nd Gen.	single	Fan-beamlet	multiple	yes	Trans.+ Rotates	Faster than 1G	Low efficiency
3rd Gen.	single	Fan-beam	many	no	Rotates together	Faster than 2G	High cost and Low efficiency
4th Gen.	single	Fan-beam	Stationary ring	no	Source Rotates only	Higher efficiency than 3G	high scatter
5th Gen.	multiple	Fan-beam	Stationary ring	no	No movement	Ultrafast for cardiac	high cost
6th Gen.	single	Fan-beam	many	yes	3rd Gen.+ bed trans.	faster 3D imaging	higher cost
7th Gen.	single	Narrow cone- beam	Multiple arrays	yes	3rd Gen.+ bed trans.	faster 3D imaging	higher cost
8th Gen.	single	wide cone- beam	FPD	no	3rd Gen.	Large 3D	Relatively slow

Multi-Detector Technology





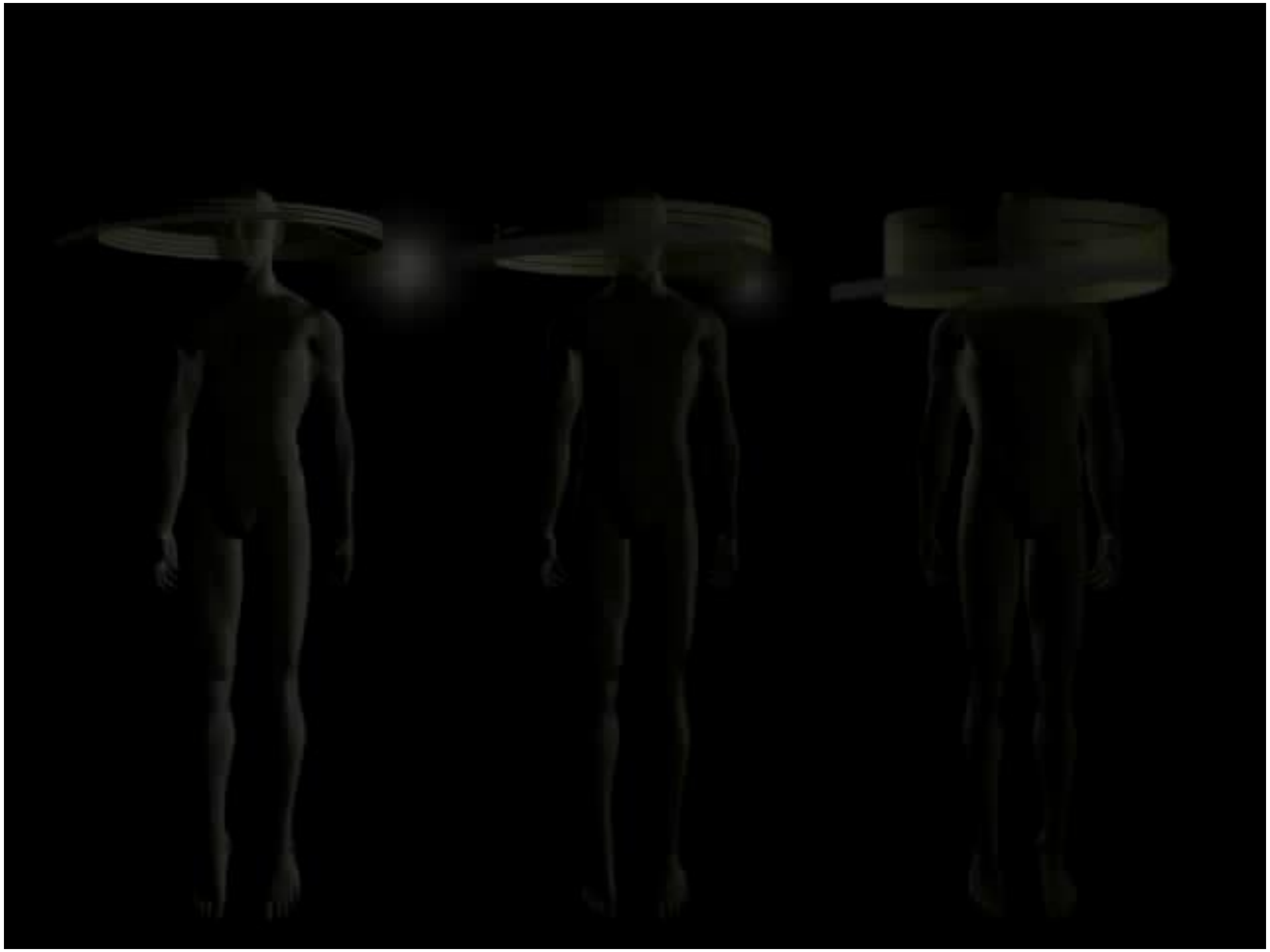


Single slice vs MDCT

Adaptive Arrays

Fixed matrix

Multi-detector Allows More Coverage with Thinner Slices



Credit: Toshiba Medical

Multi-slice Detector Design

- There are essentially two Detector designs - Fixed Matrix and Adaptive Array
- Philips,GE & Toshiba use the Fixed Matrix, Siemens use the Adaptive Array
- These two designs differ strongly

In the Market Technology

Review article: Coronary CT angiography

Table 1. 64-slice CT scanner characteristics for four main manufacturers

Manufacturer	Single-source 64-slice CT: Number of detector rows	Number of acquired slices	Maximum gantry rotation speed	Effective maximum temporal resolution	Dual-source 64-slice CT: Number of detector rows	Number of acquired slices	Maximum gantry rotation speed	Effective maximum temporal resolution	Other special features
GE Healthcare	64 × 0.625	64	350 ms	175 ms					
Philips Healthcare	64 × 0.625	64	400 ms	200 ms					
Siemens Healthcare	64 × 0.6	64	300 ms	150 ms	2 × 32 × 0.6[a]	64	330 ms	165 ms	Dual energy function
Toshiba Medical Systems	64 × 0.5	64	350 ms	175 ms					

[a] z-flying focal spot technique is used to acquire 64 slices. In addition, dual-energy CT is available with two tubes emitting X-ray spectra of different energy levels.

Sun et al, BJR, May 2012

Current Technology

Table 2. 128-, 256- and 320-slice CT scanner characteristics for four main manufacturers

Manufacturer/scanner	128-slice CT				256-slice CT				320-slice CT			
	Number of detector rows	Number of acquired slices	Maximum gantry rotation speed	Effective maximum temporal resolution	Number of detector rows	Number of acquired slices	Maximum gantry rotation speed	Effective maximum temporal resolution	Number of detector rows	Number of acquired slices	Maximum gantry rotation speed	Effective maximum temporal
GE Healthcare (Waukesha, WI)	2 × 64 × 0.625[a]	128	350 ms	44 ms								
Philips Healthcare (Best, Netherlands)	128 × 0.625	128	270 ms	135 ms	2 × 128 × 0.625[b]	256	270 ms	135 ms				
Siemens Healthcare (Erlangen, Germany) Definition AS	64 × 0.6[c]	128	300 ms	150 ms/75 ms								
Siemens Definition Flash	2 × 64 × 0.6[d] 2 × 64 × 0.6[e] 2 × 64 × 0.6[f]	2 × 128 2 × 128 2 × 128	280 ms 280 ms 280 ms	75 ms/37.5 ms 75 ms 75 ms								
Toshiba Medical Systems (Tochigi, Japan)	128 × 0.5	128	500 ms	250 ms	256 × 0.5	256	500 ms	250 ms	320 × 0.5	320	350 ms	175 ms

Sun et al, BJR, May 2012

Computation of The CT Number

$$\text{CT Number} = \frac{K(\mu_{material} - \mu_{water})}{\mu_{water}}$$

- K is a constant – the reference value for water
- One system is the Hounsfield system where K = 1000.
  - Another is the EMI where K=500

Conversion to HU Following Reconstruction

- Reconstruction yields an array of floating point numbers proportional to μt
- Calculation of HU with reference to water eliminates t
- Array of CT numbers as integer values are stored for use by image display hardware.

$$CT(x,y) = 1000\,\frac{\mu(x,y) - \mu_{water}}{\mu_{water}}$$

CT Imaging Parameters

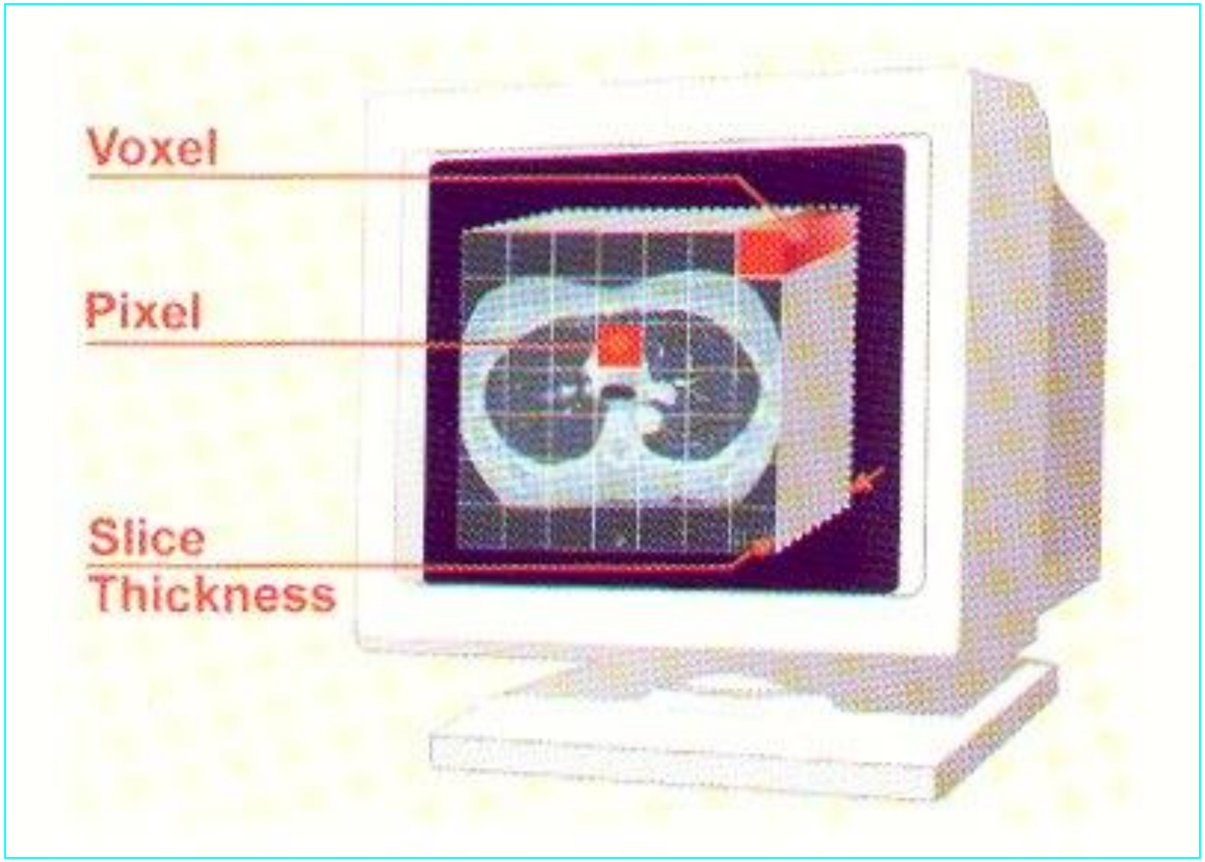
- Axial or Conventional CT
- Helical or Spiral CT
- Can vary
  - kVp
  - mA
  - time
  - algorithm

- slice thickness
  - Actual and recon.
- Field of View (Recon)
- X-ray Field Size
- Pitch

CT Image Display – Windowning and Leveling

- Each Voxel is represented on the screen as a Pixel (Picture Element)
- A voxel has a CT number that is 12 bits (4098 values), a display pixel has 8 bits (256 values)
- These values are mapped using a look up table (LUT).
- Changing the LUT is called Window width and level.

Data from Voxels are Displayed as an Array of "Pixels"



Houndsfield CT Number

$$CT\ Number = \frac{1000(\mu_{tissue} - \mu_{water})}{\mu_{water}}$$

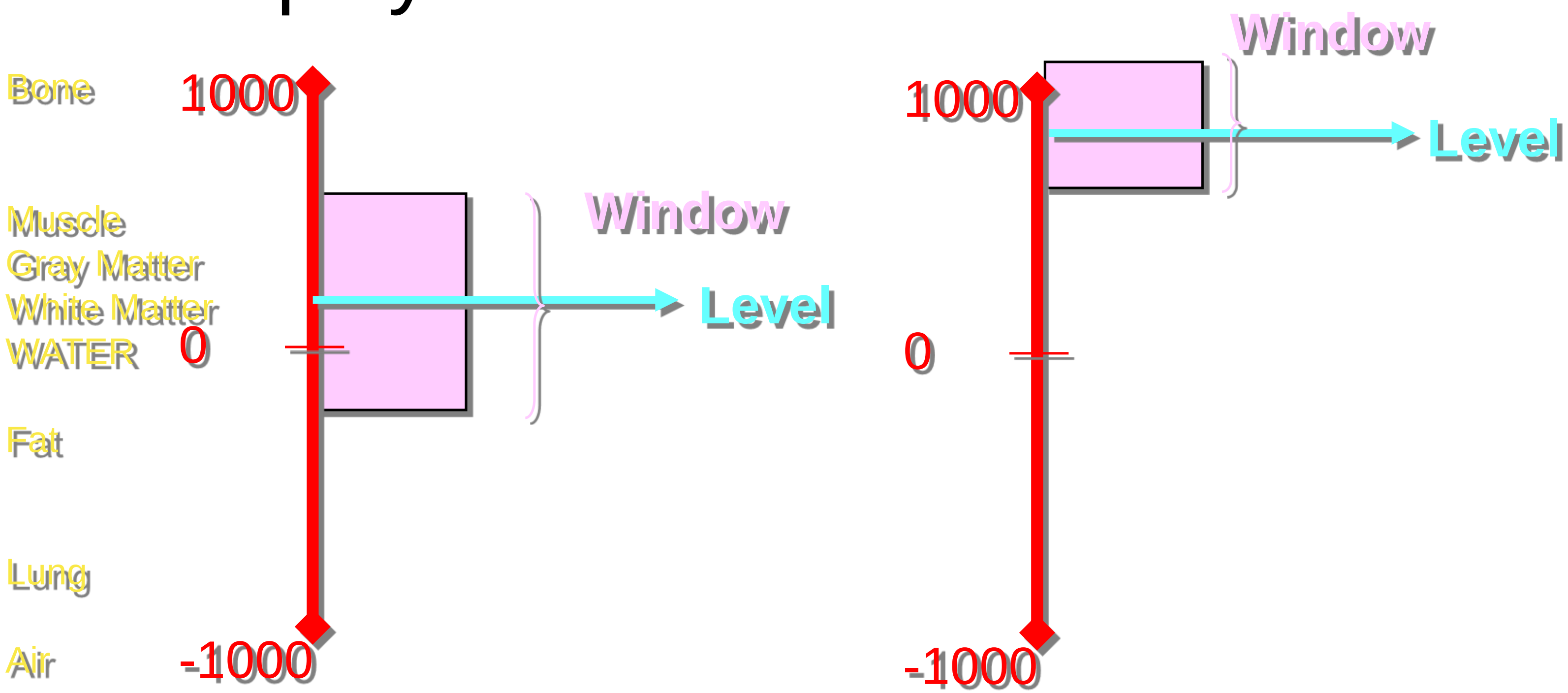
- CT number depends on attenuation coefficient of tissue
- As Compton Scatter predominates the numbers depends primarily on e^- density
  - Correlates well with mass density

Typical Hounsfield Numbers

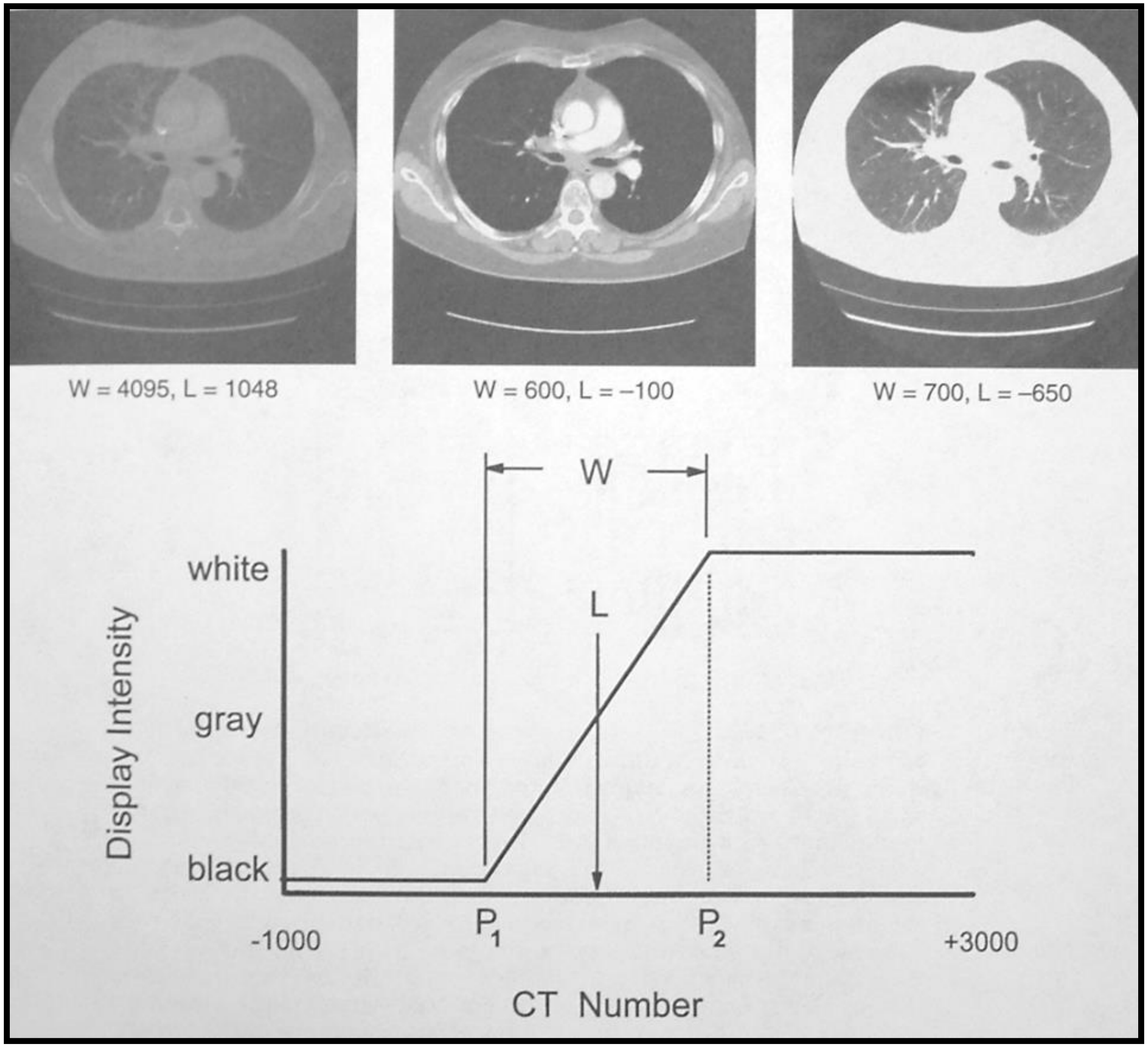
Material	HU Value	Density (g/cm3)
• **Air**	-1000	<0.01
• **Lung**	-750	0.25
• **Fat**	-90	0.92
• **Water**	0	1.00
• **White Matter**	30	1.03
• **Gray Mater**	40	1.04
• **Muscle**	50	1.06
• **Bone**	+1000	1.80

Window – Level and Width

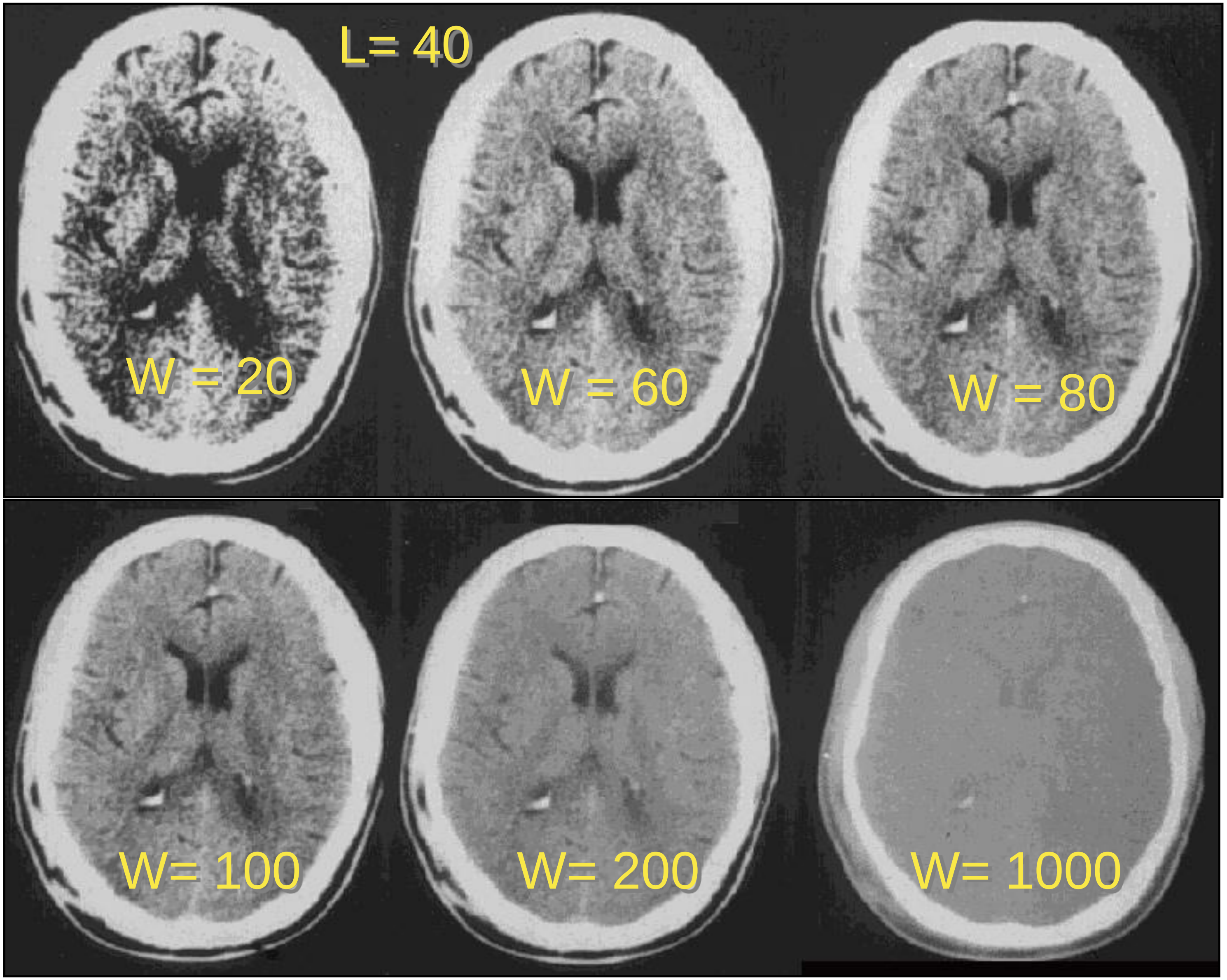
- Allows user to determine which subset of the full dynamic range of CT values to display



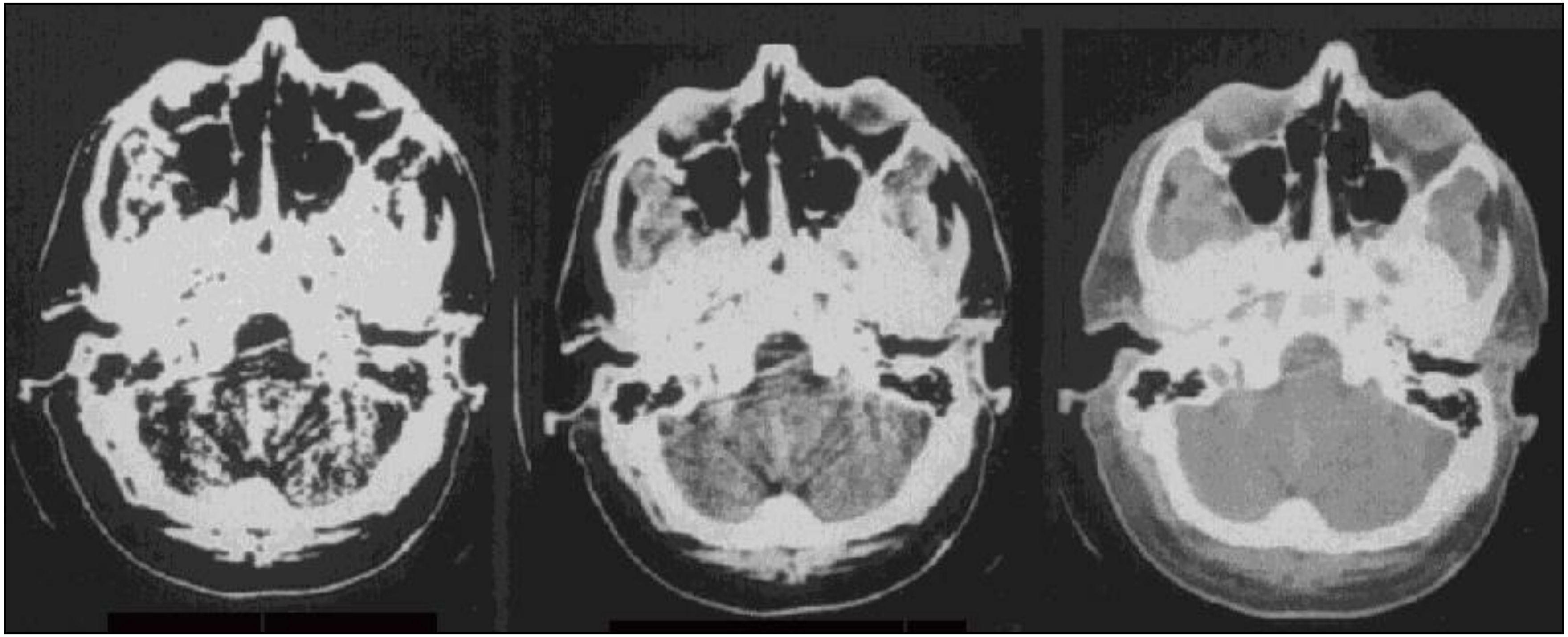
Windowing and Leveling



Soft Tissue Window Width



Bone Settings for Window - Width



L = 50, W= 20 L = 50, W =200 L=50, W =1000

CT vs MRI

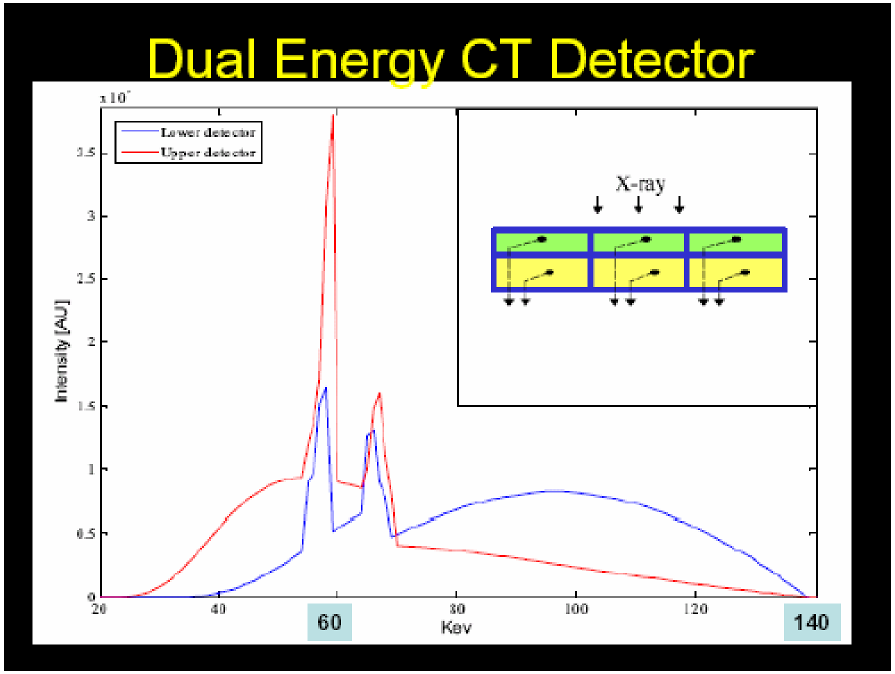
- Contrast mainly depends on HU (CT#)
- HU is energy dependent
- Fast scan with good resolution

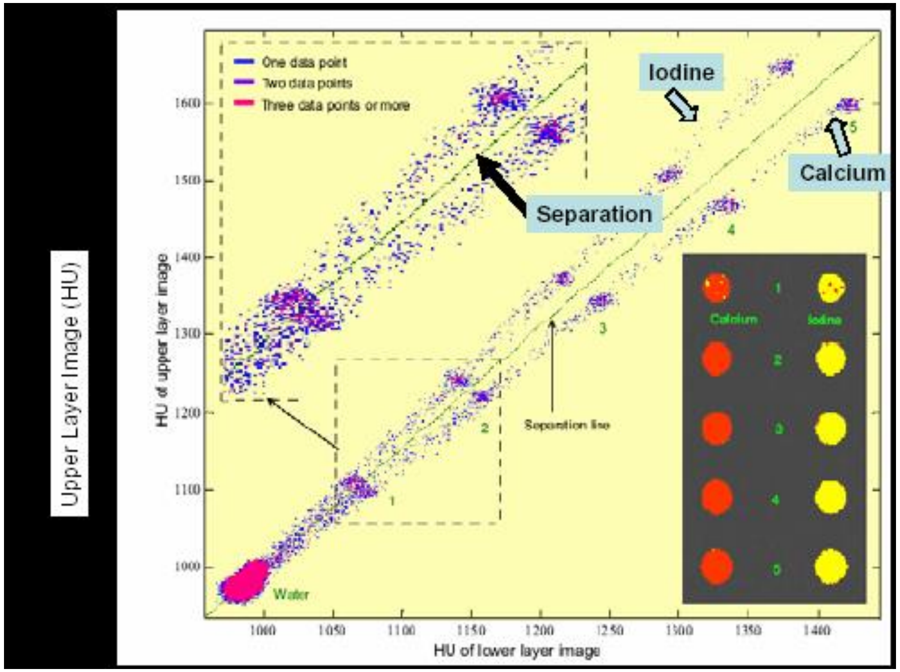
- Slower scan time, fair resolution but superior contrast mechanism
  - T1, T2, PD, Flow, Chemical shift, Susceptibility, MT, etc

Dual Energy/Spectral CT Methods

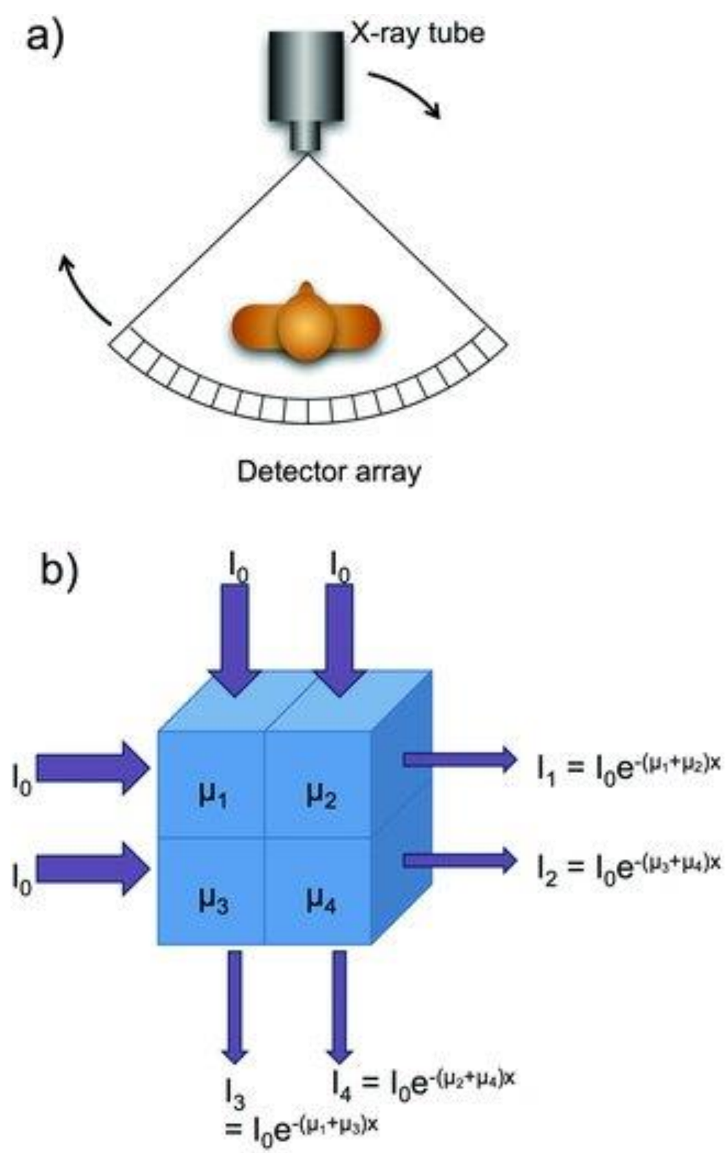
- Dual sources (Siemens)
- Energy discriminating detectors (Philips)
- kVp switching (GE)

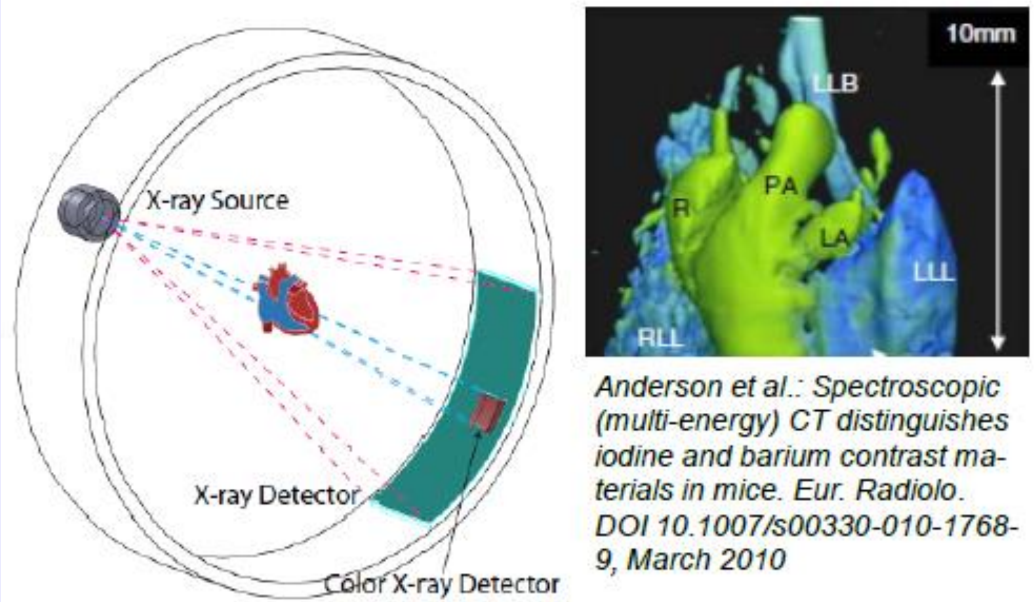
Dual Energy CT





www.aapm.org





Anderson et al.: Spectroscopic (multi-energy) CT distinguishes iodine and barium contrast materials in mice. Eur. Radiolo. DOI 10.1007/s00330-010-1768-9, March 2010

Spectra CT

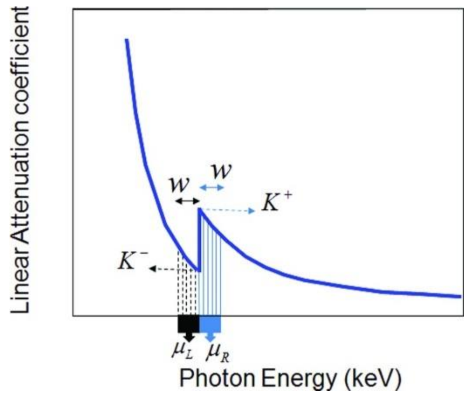
(a) Schematic drawing of third-generation CT. CT images are acquired during the rotation of an X-ray tube and an array of detectors. (b) Schematic attenuation profiles of voxels. Measured X-ray intensity can be expressed as sum of the attenuation along the path of X-ray.

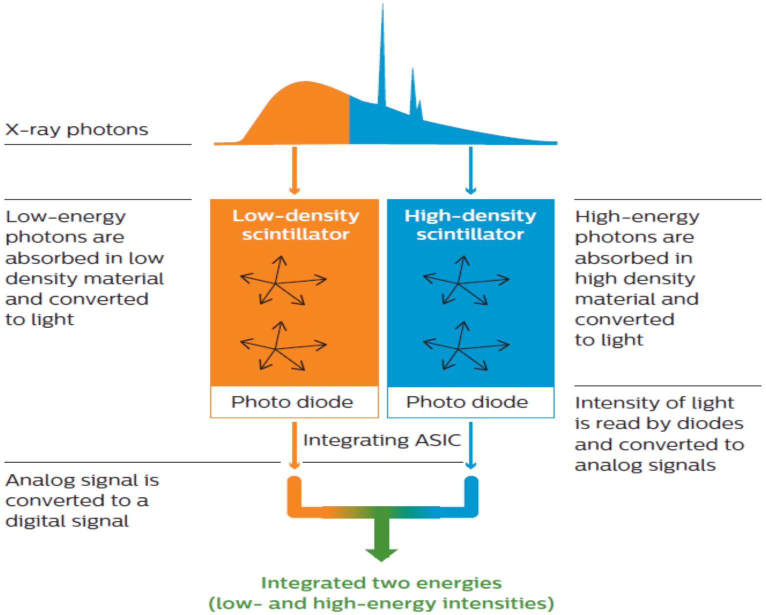
Advanced Detector Technology

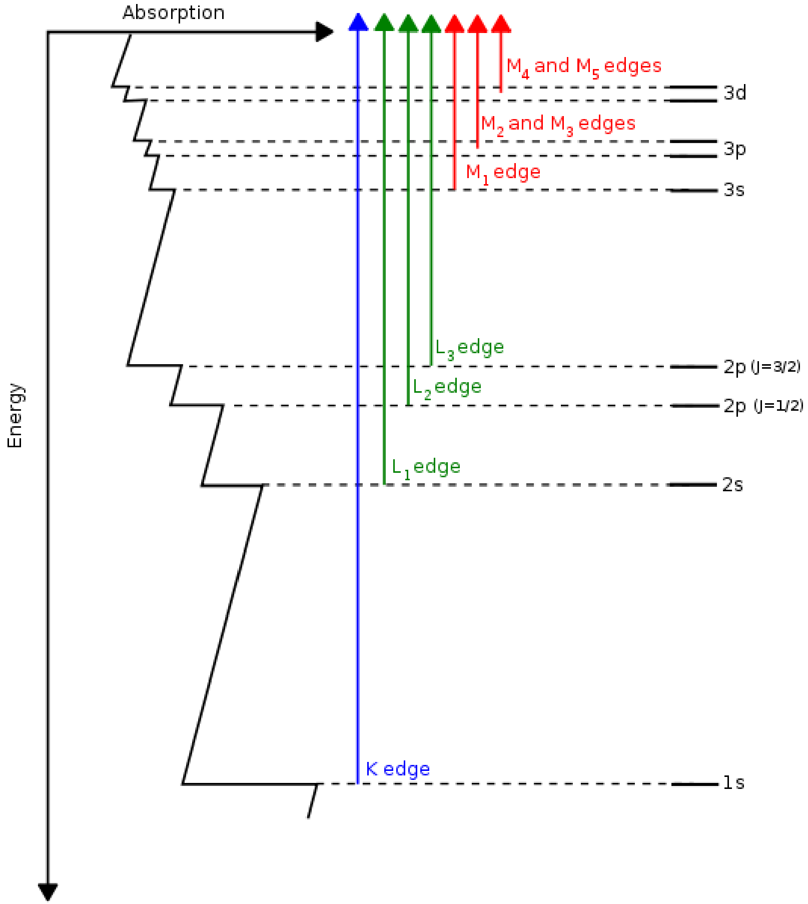
Energy discriminating photon counting detectors

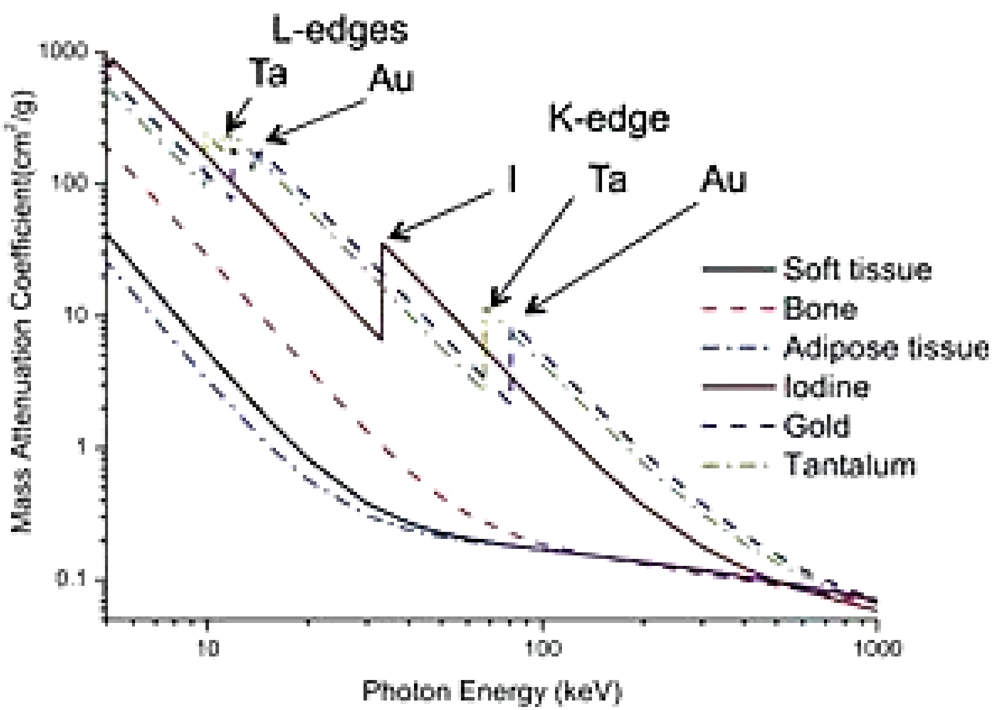
Spectral/multi energy CT has the potential to distinguish different materials by K-edge characteristics.

K-edge imaging involves the two energy bins on both sides of a K-edge



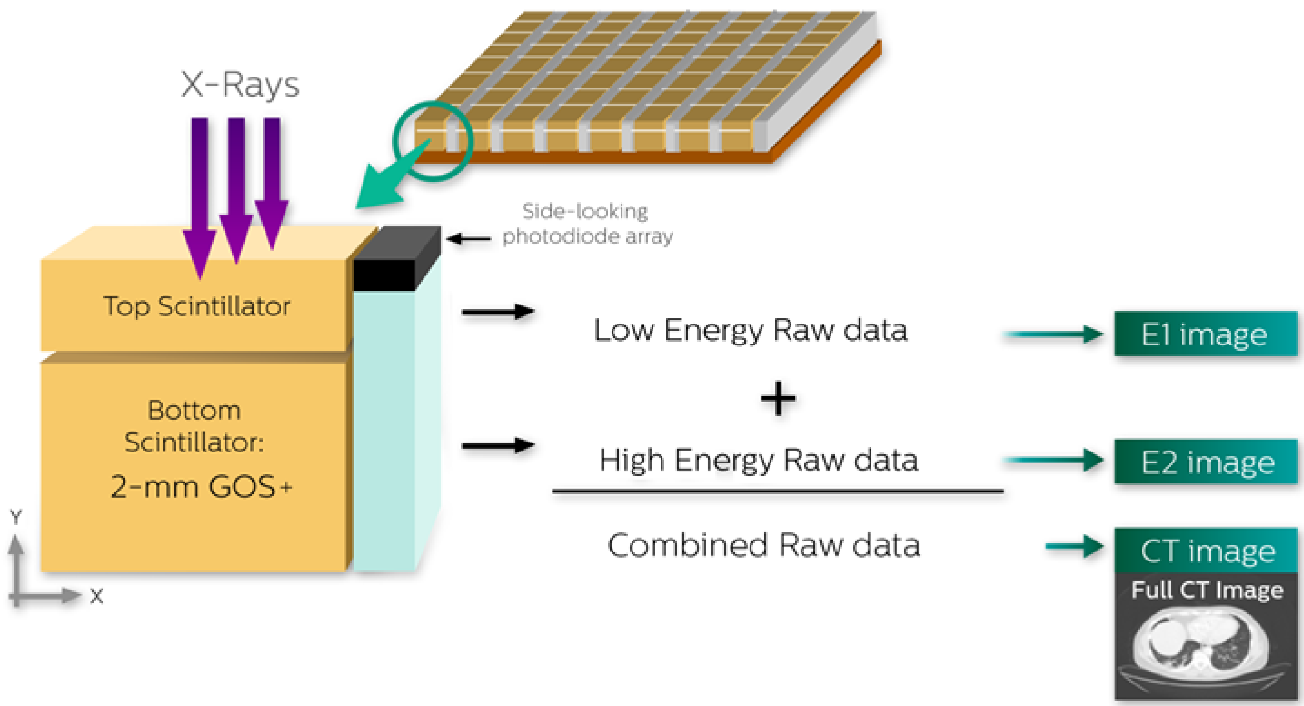






Excitation of a 1s electron occurs at the K-edge, while excitation of a 2s or 2p electron occurs at an L-edge

Understanding spectral detector technology and its impact on CT imaging

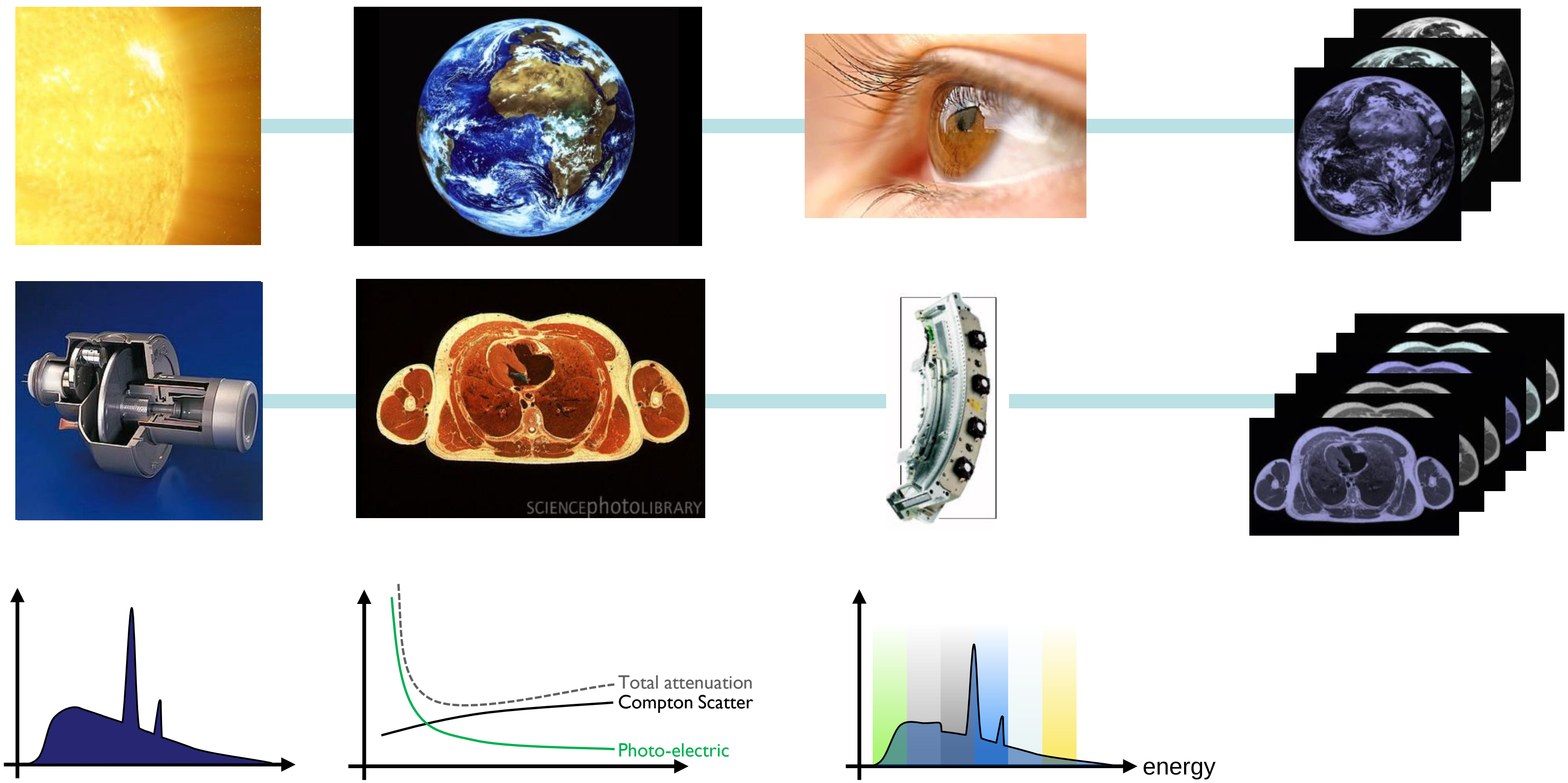


High- and low-energy data can be obtained simultaneously in time and space at the detector level

Philips

Spectral CT with Energy-Resolving Detector

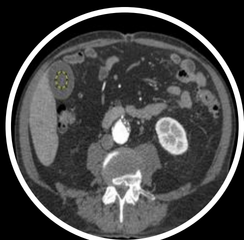
Energy-resolving detectors discriminate colors



Spectral CT with energy-resolving detector is like the **human eye at day**

Spectral Results Categories by Units

Spectral results can be roughly divided into 2 groups: **HU based** and **non-HU based**



HU

Grayscale Images

MonoEnergy

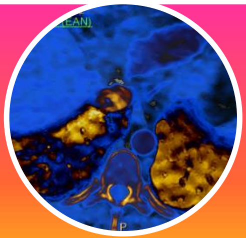
Modified HU (HU*), Processed results

VNC

Non HU, mg/ml

Images

Iodine - no water

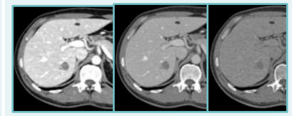
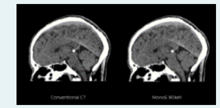
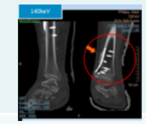
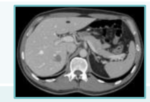
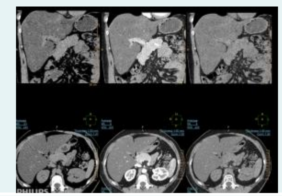
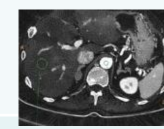
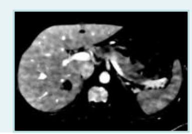
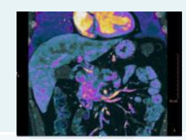
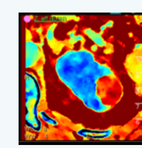


Non HU, Effective Atomic Number (Color) Images

Fused

Effective atomic number

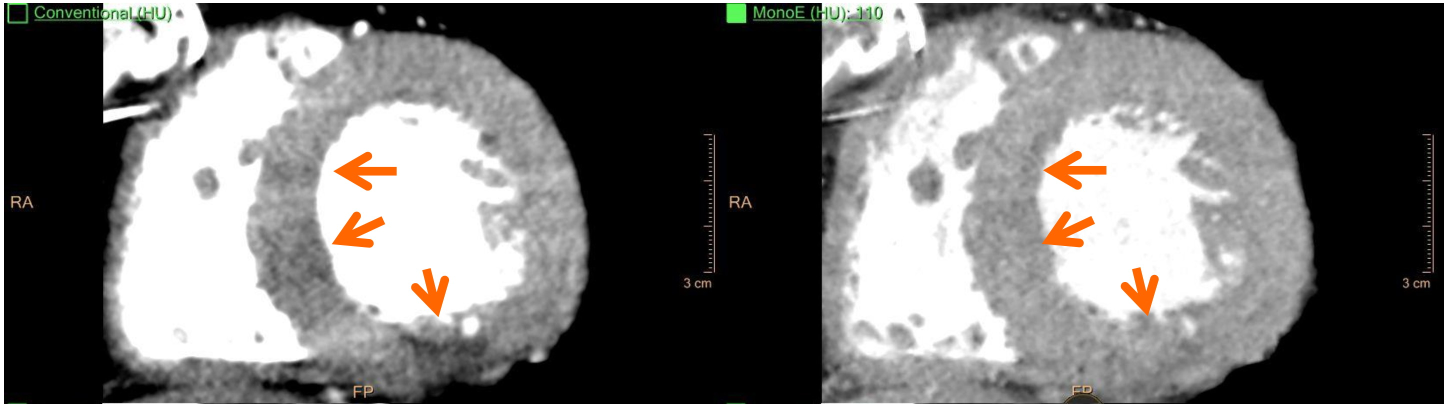
Philips

Spectral Result	Units	Clinical Objective	Examples
MonoE	■ HU	• Improved iodine conspicuity • Reduce beam hardening artifact • Metal artifact reduction	  
MonoE Equivalent *(120kVp Equivalent)*	■ HU	• Improve image quality	
Virtual Non Contrast	■ HU* (modified HU)	• Use when non-enhanced scans aren't performed • Dose management • Use existing 120 kVp protocols with dose modulation and iterative reconstruction (IMR)	
Iodine no Water	● mg/ml	• Enhancement of iodinated contrast (Ca also bright) • Quantification of Iodine	
Iodine Density	● mg/ml	• Enhancement of iodinated contrast (Ca removed) • Visualization & Quantification of Iodine	 
Effective Z	● Effective Atomic Number (EAN)	• Ability to characterize stuctures based on material content (differentiation of hemorrhage vs Ca; differentiation of kidney stones)	

Philips

Solving clinical questions:

Beam hardening reduction (Cardiac)

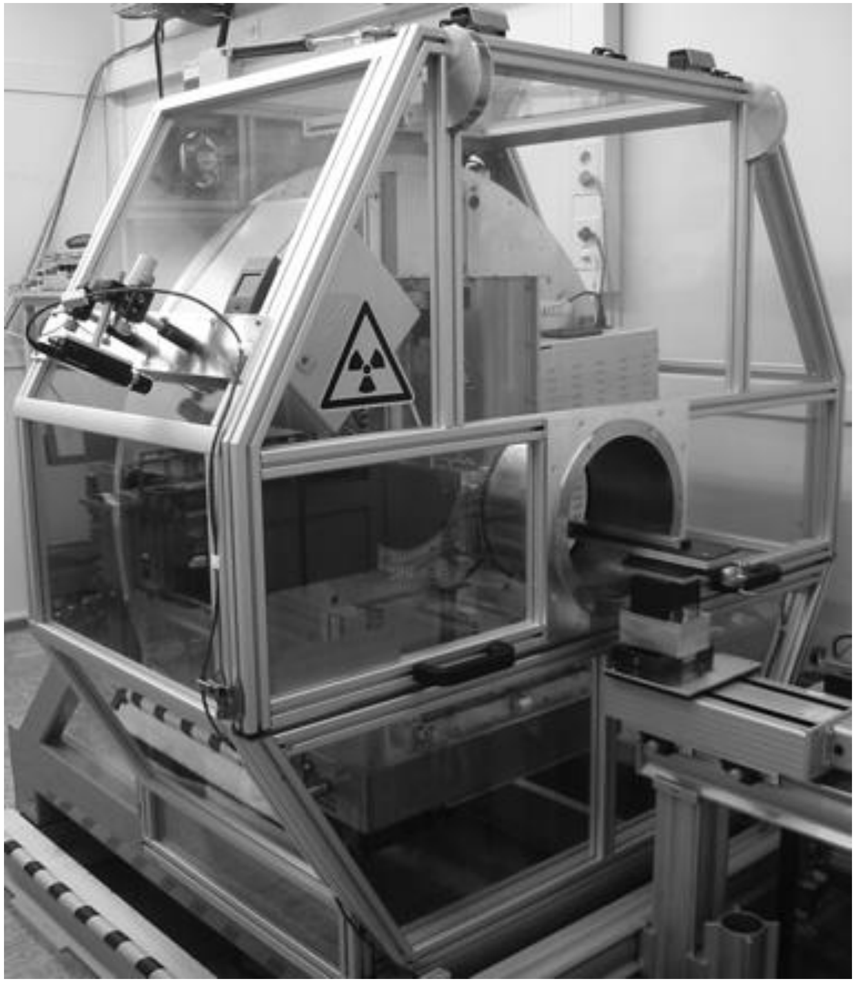


Conventional 120kVp

MonoE 110keV

Philips

Emerging Opportunities with Spectral CT



Philips Research, Hamburg, DE
Relevant Patents: US20110096892; 20110096905 (Philips)

Multicolored or spectral CT has the potential to detect and quantify intraluminal fibrin presented by ruptured plaque in the context of CT angiograms all without **calcium interference**.

Diagnosis of Chest Pain of Cardiac Origin

Diagnostic Imaging – Treatment Planning – Intervention Guidance

Symptoms

Patient presented at ER with chest pain

Early Diagnosis

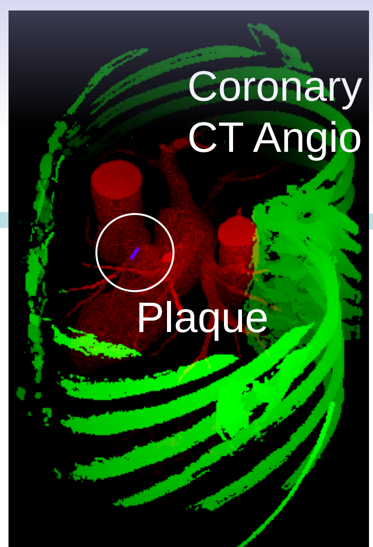
Stress Test/ Hospitalization

Diagnosis

Cardiac CT angiography (CCTA)

Surplant invasive diagnostic cardiac catheterization with a quicker, noninvasive, lower cost procedure

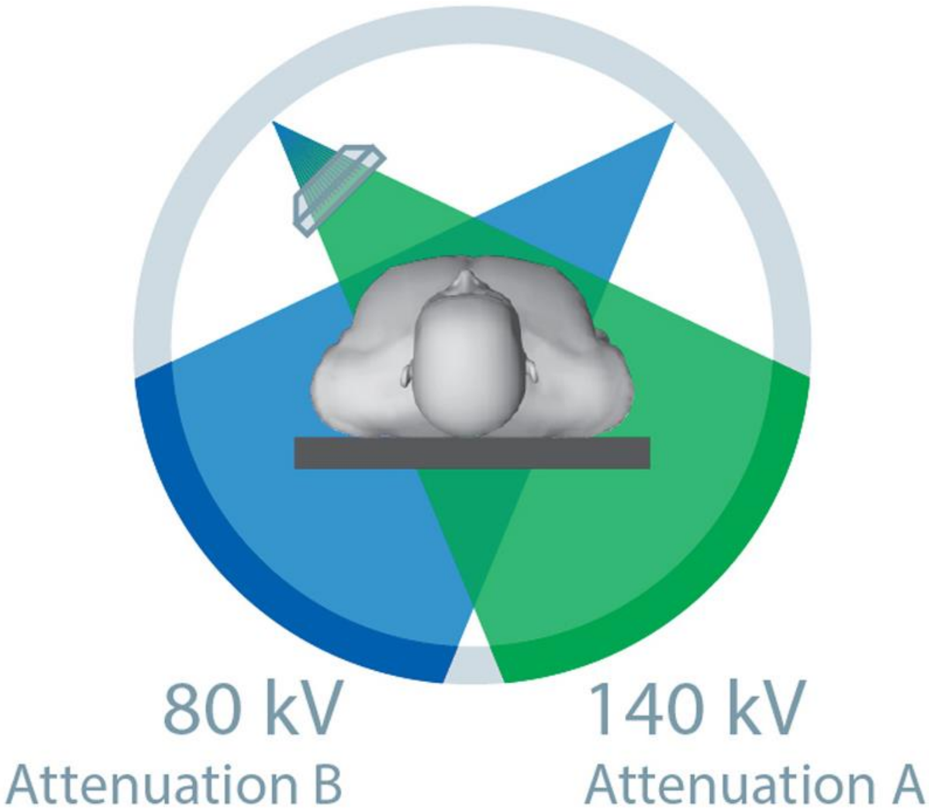
lower cost procedure
quicker, noninvasive,
catheterization with a



Detecting Atherosclerotic Plaque

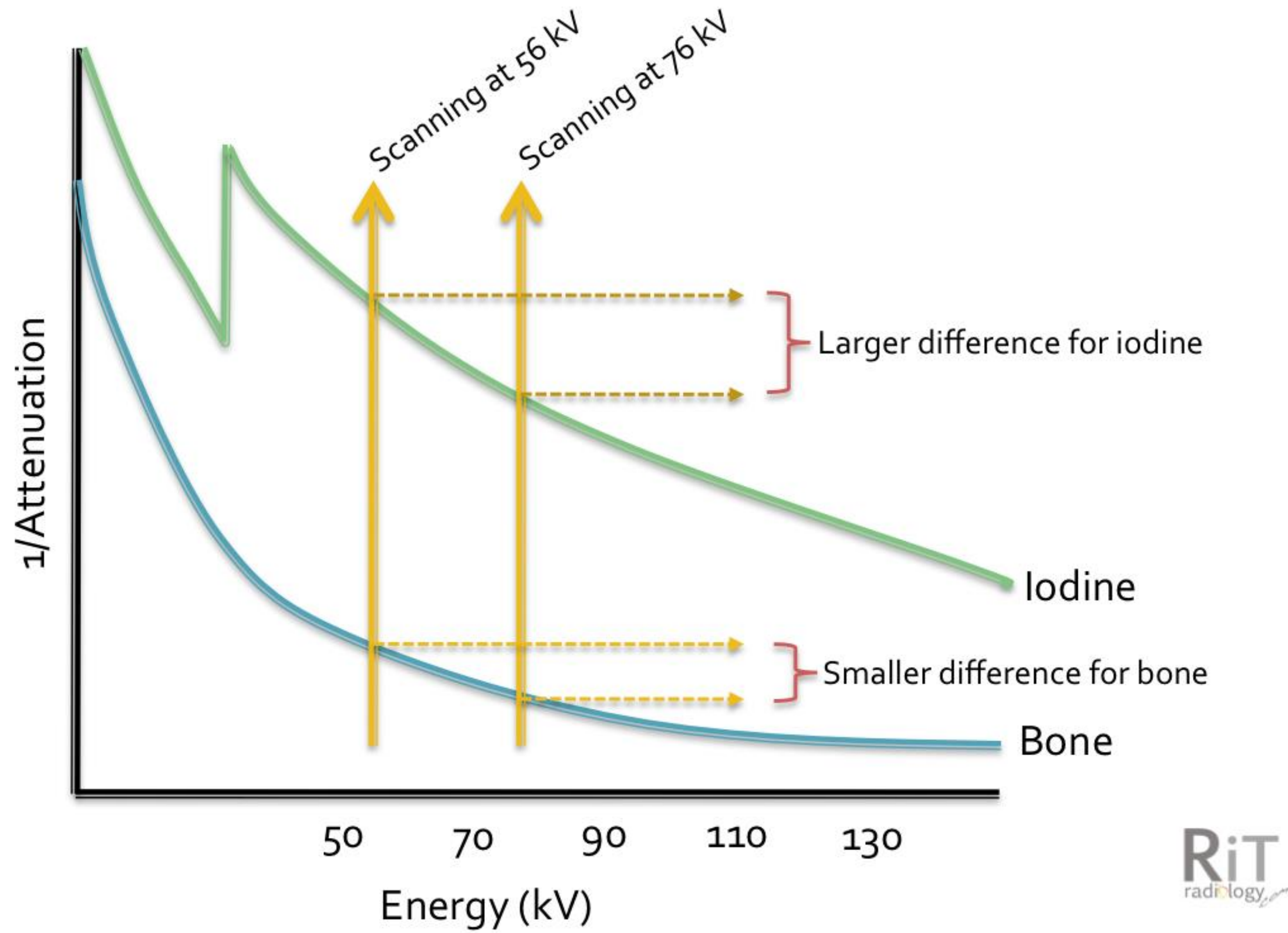
Dual Energy CT

The Selective Photon Shield ensures dose neutrality by eliminating spectral overlap. This makes Dual Energy as dose-efficient as any single 120 kV scan.



- During a Dual Source Dual Energy scan, two CT datasets are acquired simultaneously with different kV and mA levels, allowing to visualize differences in the energy-dependence of the attenuation coefficients of different materials.
- These images are combined and analyzed to visualize information about anatomical and pathological structures.

http://www.healthcare.**siemens**.com/computed-tomography/technologies-innovations/ct-dual-energy/technical-specifications


Scanning at 56 kV
Scanning at 76 kV
1/Attenuation
Larger difference for iodine
Iodine
Smaller difference for bone
Bone
50
70
90
110
130
Energy (kV)
RiT
radiology

One Basic Reason for Use of Dual Energy CT: Material Differentiation

- By scanning a patient at two different energy spectra (e.g. at 56 kV and 76 kV), the attenuation difference of the same material is different.
- Iodine has higher attenuation difference, compared to bone.
- Scanning allows the computer to process bone and iodine content on images differently.

Routine Use of Dual-energy CT for Material Differentiation

- Creation of 3D vascular images ("Direct Angio") by easy removal of bony structures
- Plaque analysis (calcified vs. soft plaques)
- Lung perfusion
- Virtual unenhanced scan (creation of unenhanced scan from enhanced images by deleting iodine content from the images)
- Calculi characterization (uric acid vs. others)

Dual Energy in Angiography

Use the spectral properties of iodine to differentiate it from other dense materials in the dataset (similar to magnetic resonance angiography (MRA)).

With Dual Energy CT, it is possible to identify bone by its spectral behavior and to erase it from an angiogram. Then, the iodine in the vessels remains the only dense material in the dataset and a MIP can be calculated from a CT angiogram to closely resemble an MRA.

Additionally, it is possible to detect those voxels that contain both calcium and iodine and add them back to the dataset.

Calcified plaques of atherosclerotic vessels can thereby be switched on and off in the dataset to visualize both the residual lumen and the plaque distribution.

http://www.dsct.com/index.php/clinical-applications-dual-energy-ct/

Dual Energy CT: Dose Reduction

- Greatest potential is replacing true non-contrast phase of single energy multi-phase CT exam with virtual non-contrast images. Examples:
  - 4 phase liver becomes a 3 phase exam
  - 3 phase pancreas becomes a 2 phase exam
  - 2-3-4 phase CT IVP becomes 1-2-3 phase
- Potential dual energy dose reduction is 25 -50% compared

Clinical significance

Radiology

Dual-Energy Head CT Enables Accurate Distinction of Intraparenchymal Hemorrhage from Calcification in Emergency Department Patients[1]

Ranliang Hu, MD
Laleh Daftari Besheli, MD
Joseph Young, MD
Markus Wu, MD
Stuart Pomerantz, MD
Michael H. Lev, MD
Rajiv Gupta, MD, PhD

Purpose: To evaluate the ability of dual-energy (DE) computed tomography (CT) to differentiate calcification from acute hemorrhage in the emergency department setting.

Materials and Methods: In this institutional review board-approved study, all unenhanced DE head CT examinations that were performed in the emergency department in November and December 2014 were retrospectively reviewed. Simulated 120-kVp single-energy CT images were derived from the DE CT acquisition via postprocessing. Patients with at least one focus of intraparenchymal hyperattenuation on single-energy CT images were included, and DE material decomposition postprocessing was performed. Each focal hyperattenuation was analyzed on the basis of the virtual noncalcium and calcium overlay images and classified as calcification or hemorrhage. Sensitivity, specificity, and accuracy were calculated for single-energy and DE CT by using a common reference standard established by relevant prior and follow-up imaging and clinical information.

Results: Sixty-two cases with 68 distinct intraparenchymal hyperattenuating lesions in which the reference standards were available were included in the study, of which 41 (60%) were confirmed as calcification and 27 (40%) were confirmed as hemorrhage. Sensitivity, specificity, and accuracy of DE CT for the detection of hemorrhage were 96% (95% confidence interval [CI]: 81%, 100%), 100% (95% CI: 91%, 100%), and 99% (95% CI: 92%, 100%) and those of single-energy CT were 74% (95% CI: 54%, 89%), 95% (95% CI: 83%, 99%), and 87% (95% CI: 76%, 94%), respectively. Six of 68 (9%) lesions were classified as indeterminate and three (4%) were misinterpreted with single-energy CT alone and were correctly classified with DE CT.

Conclusion: DE CT by using material decomposition enables accurate differentiation between calcification and hemorrhage in patients presenting for emergency head imaging and can be especially useful in problem-solving complex cases that are difficult to determine based on conventional CT appearance alone.

Online supplemental material is available for this article.

[1] From the Department of Radiology, Massachusetts General Hospital, Harvard Medical School, 55 Fruit St, GRB-273A, Boston, MA 02114. Received April 13, 2015; revision requested June 8; final revision received August 23; accepted September 22; final version accepted October 20. **Address correspondence to** R.G. (e-mail: *rgupta1@partners.org*).

R.H. and L.D.B. contributed equally to this work.

CT dose and how to reduce risks

- CT provides significant clinical information but impose higher radiation dose
- To reduce dose, numbers of approach can be done

Radiation Dose to Patients From Common Imaging Examinations

	Procedure	**Approximate effective radiation dose	Comparable to natural background radiation for	* Estimated lifetime risk of fatal cancer from examination
ABDOMINAL REGION	Computed Tomography (CT) — Abdomen and Pelvis	10 mSv	3 years	Low
	Computed Tomography (CT) — Abdomen and Pelvis, repeated with and without contrast material	20 mSv	7 years	Moderate
	Computed Tomography (CT) — Colonography	10 mSv	3 years	Low
	Intravenous Pyelogram (IVP)	3 mSv	1 year	Low
	Radiography (X-ray) — Lower GI Tract	8 mSv	3 years	Low
	Radiography (X-ray) — Upper GI Tract	6 mSv	2 years	Low
BONE	Radiography (X-ray) — Spine	1.5 mSv	6 months	Very Low
	Radiography (X-ray) — Extremity	0.001 mSv	3 hours	Negligible
CENTRAL NERVOUS SYSTEM	Computed Tomography (CT) — Head	2 mSv	8 months	Very Low
	Computed Tomography (CT) — Head, repeated with and without contrast material	4 mSv	16 months	Low
	Computed Tomography (CT) — Spine	6 mSv	2 years	Low
CHEST	Computed Tomography (CT) — Chest	7 mSv	2 years	Low
	Computed Tomography (CT) — Lung Cancer Screening	1.5 mSv	6 months	Very Low
	Radiography — Chest	0.1 mSv	10 days	Minimal
DENTAL	Intraoral X-ray	0.005 mSv	1 day	Negligible
HEART	Coronary Computed Tomography Angiography (CTA)	12 mSv	4 years	Low
	Cardiac CT for Calcium Scoring	3 mSv	1 year	Low
MEN'S IMAGING	Bone Densitometry (DEXA)	0.001 mSv	3 hours	Negligible
NUCLEAR MEDICINE	Positron Emission Tomography — Computed Tomography (PET/CT)	25 mSv	8 years	Moderate
WOMEN'S IMAGING	Bone Densitometry (DEXA)	0.001 mSv	3 hours	Negligible
	Mammography	0.4 mSv	7 weeks	Very Low

*Risk Level	Negligible	Minimal	Very Low	Low	Moderate
Estimated additional risk of fatal cancer for an adult from examination	Less than 1 in 1,000,000	1 in 1,000,000 to 1 in 100,000	1 in 100,000 to 1 in 10,000	1 in 10,000 to 1 in 1,000	1 in 1,000 to 1 in 500

Note: These risk levels represent very small additions to the 1 in 5 chance we all have of dying from cancer.

Important: Pediatric patients vary in size. Doses given to pediatric patients will vary significantly from those given to adults.

RadiologyInfo.org RSNA ACR

For the most current information, visit radiologyinfo.org.

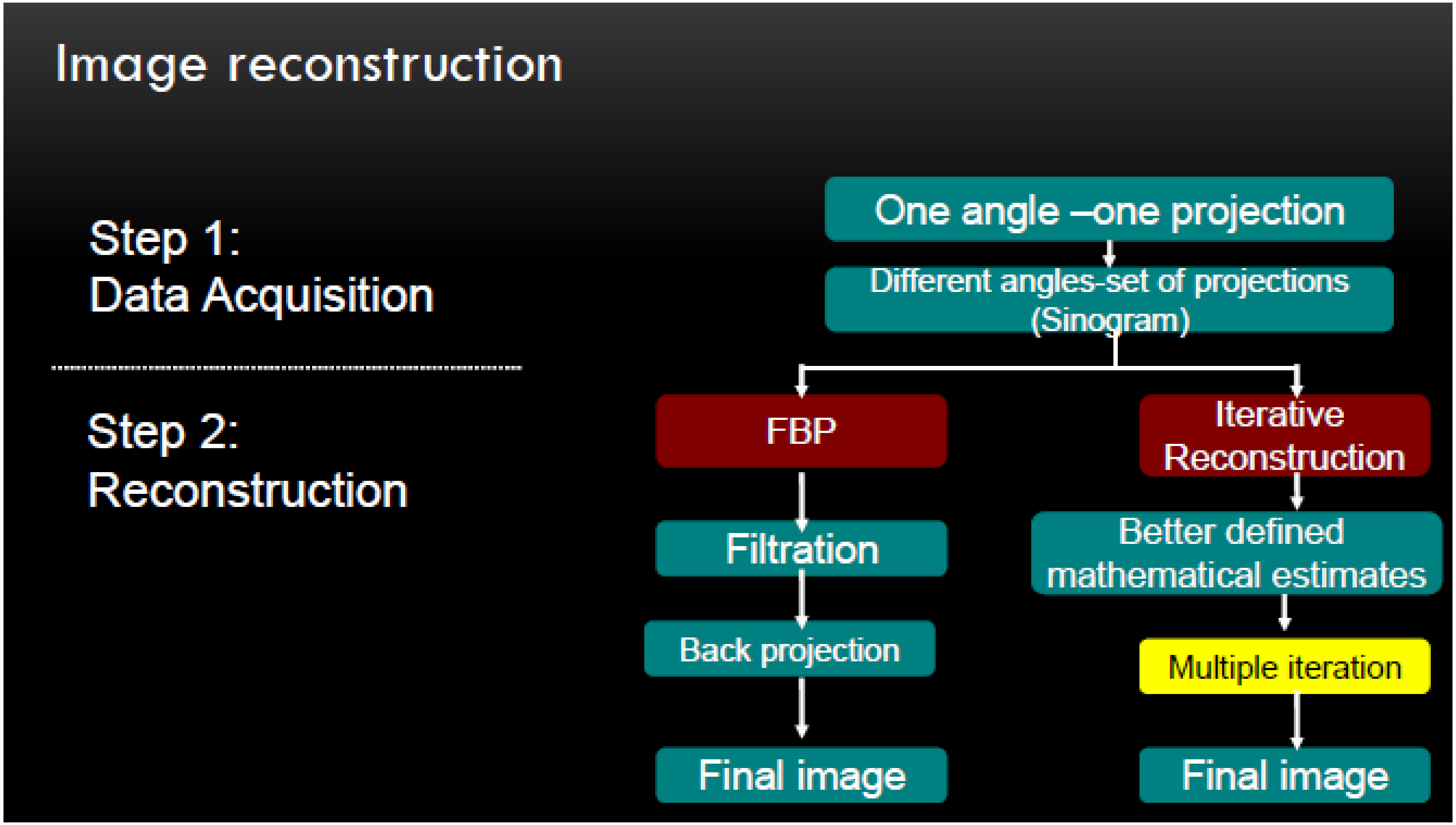
** The effective doses are typical values for an average-sized adult. The actual dose can vary substantially, depending on a person's size as well as on differences in imaging practices.[1]
[1]Mettler, F.A., et al. "Effective doses in radiology and diagnostic nuclear medicine: a catalog." *Radiology*, July 2008: 248(1):254–263.

06.15

Different approaches

- Detector technology
  - Gemstone, NanoPearl, Stellar, etc
- Post-patient collimation
- Iterative reconstruction
- Automatic kV selection
- Organ sensitive dose reduction
- Automatic Tube Current Modulation

Approach to reduce dose through iterative based reconstruction



Vendors approach

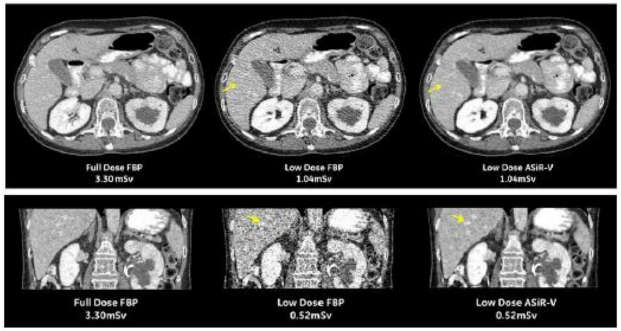
	GE	**Siemens**	**Philips**	**Toshiba**
Name	Veo - ASIR	Iris/ Safire	IMR/ iDose	AIDR

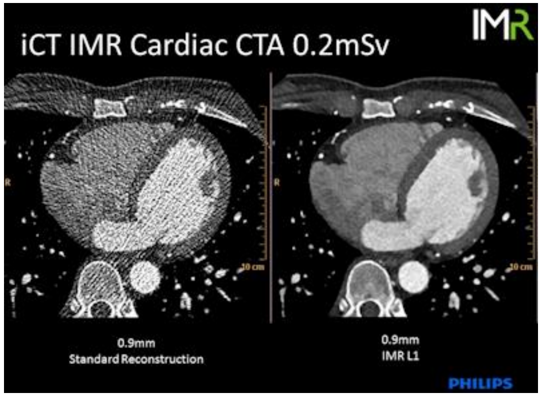
Iterative Reconstruction

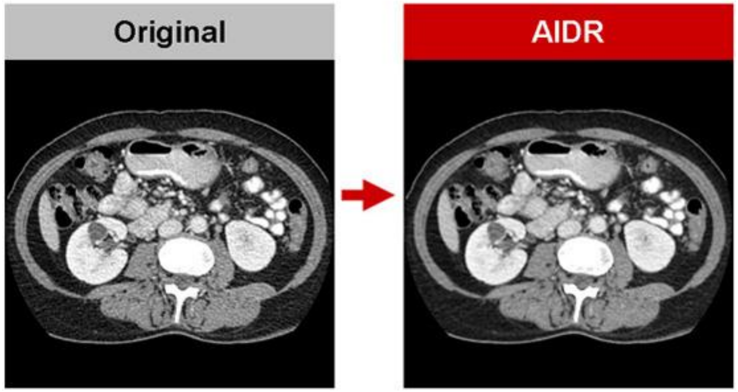
SIEMENS – ADMIRE

Five image noise and sharpness levels

ASiR-V







Dental X-rays

Topics

- ✓ Dental X-ray equipment
- ✓ Radiation protection in dental radiology
- ✓ Quality control for dental equipment

Overview

- To be able to apply the principle of radiation protection to dental radiology system including design and Quality Control.

Part 22: Optimization of protection in dental radiology

Topic 1: Dental x-ray equipment

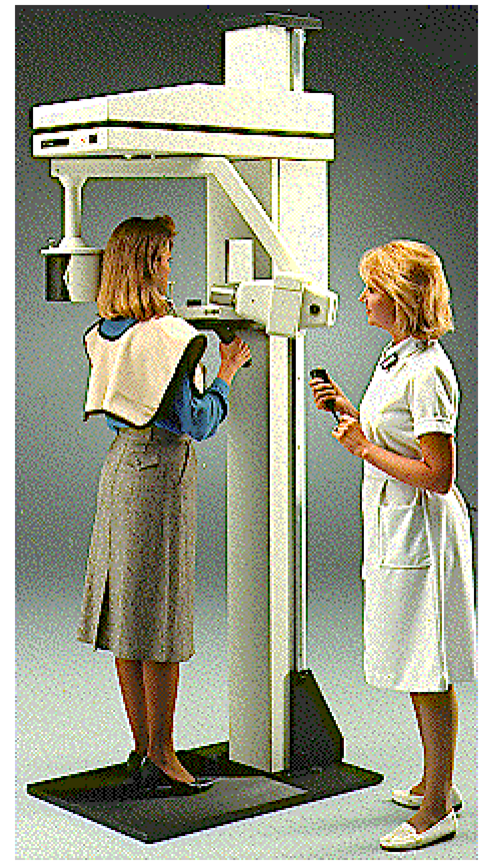
Types of units

- “Intra-Oral” units
  - Standard dental tube
  - Uses an intra-oral image receptor and extra-oral x-ray tube
- Panoramic (Orthopantomography, OPG)
- Cephalometric (Ceph)

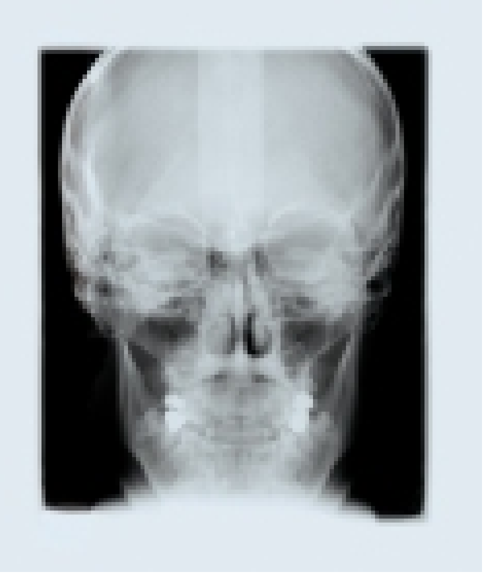
Intra-Oral Dental X-Ray Equipment

Modern Dental X-Ray Unit

Panoramic X-Ray Equipment

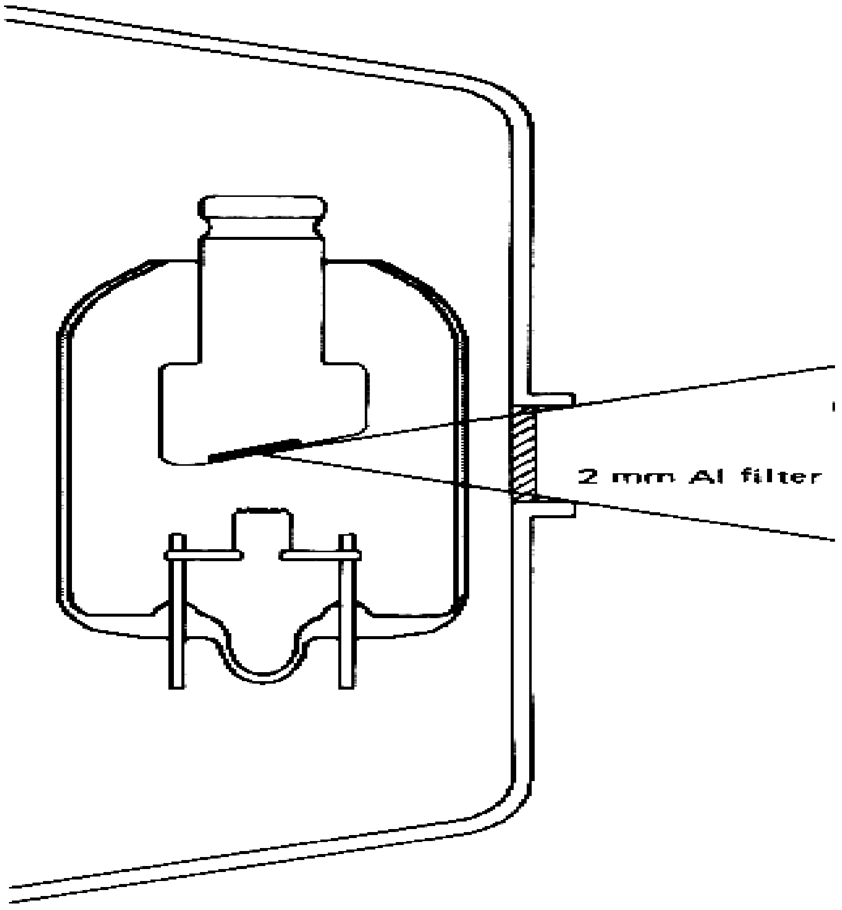


Cephalometric X-Ray Equipment

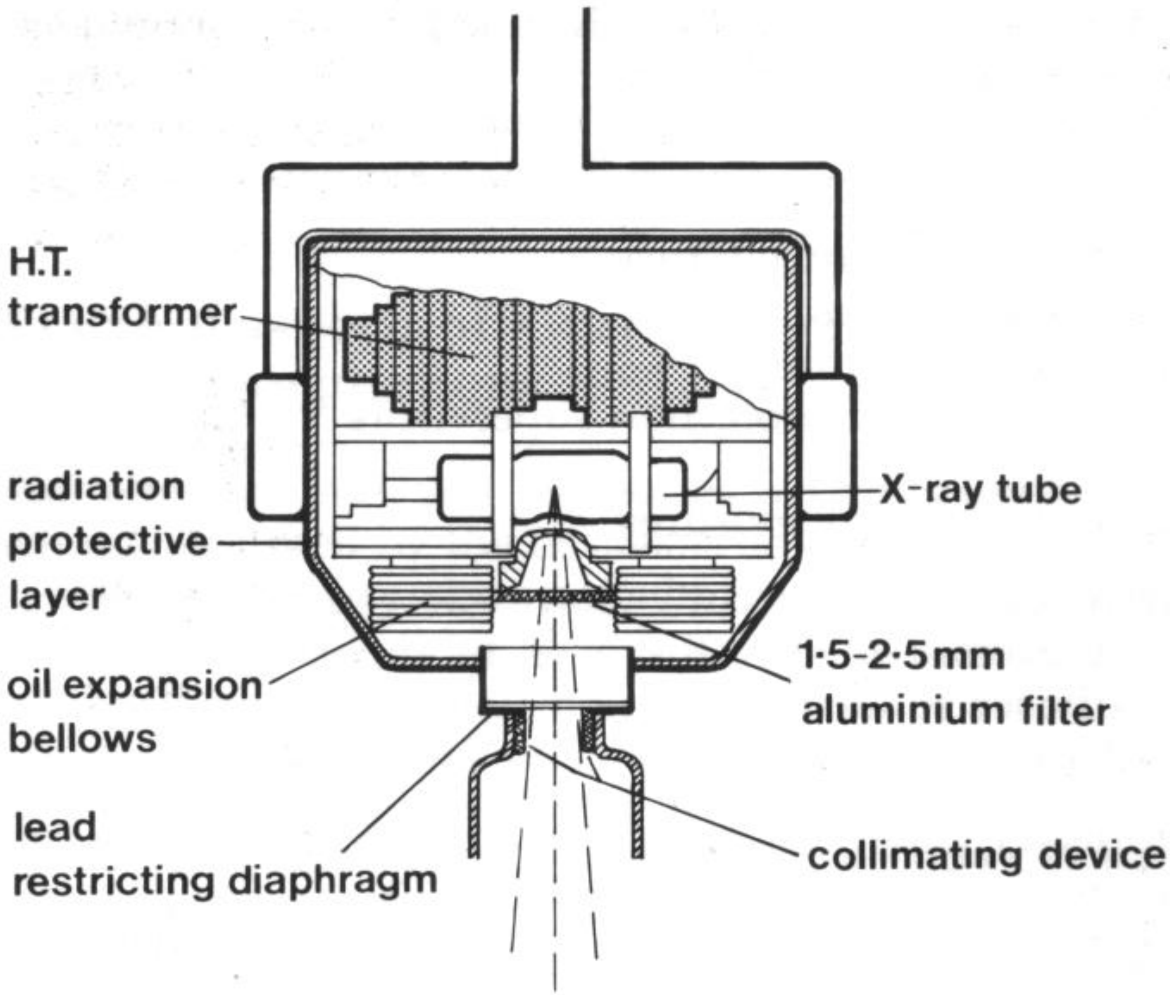


X-Ray Tube

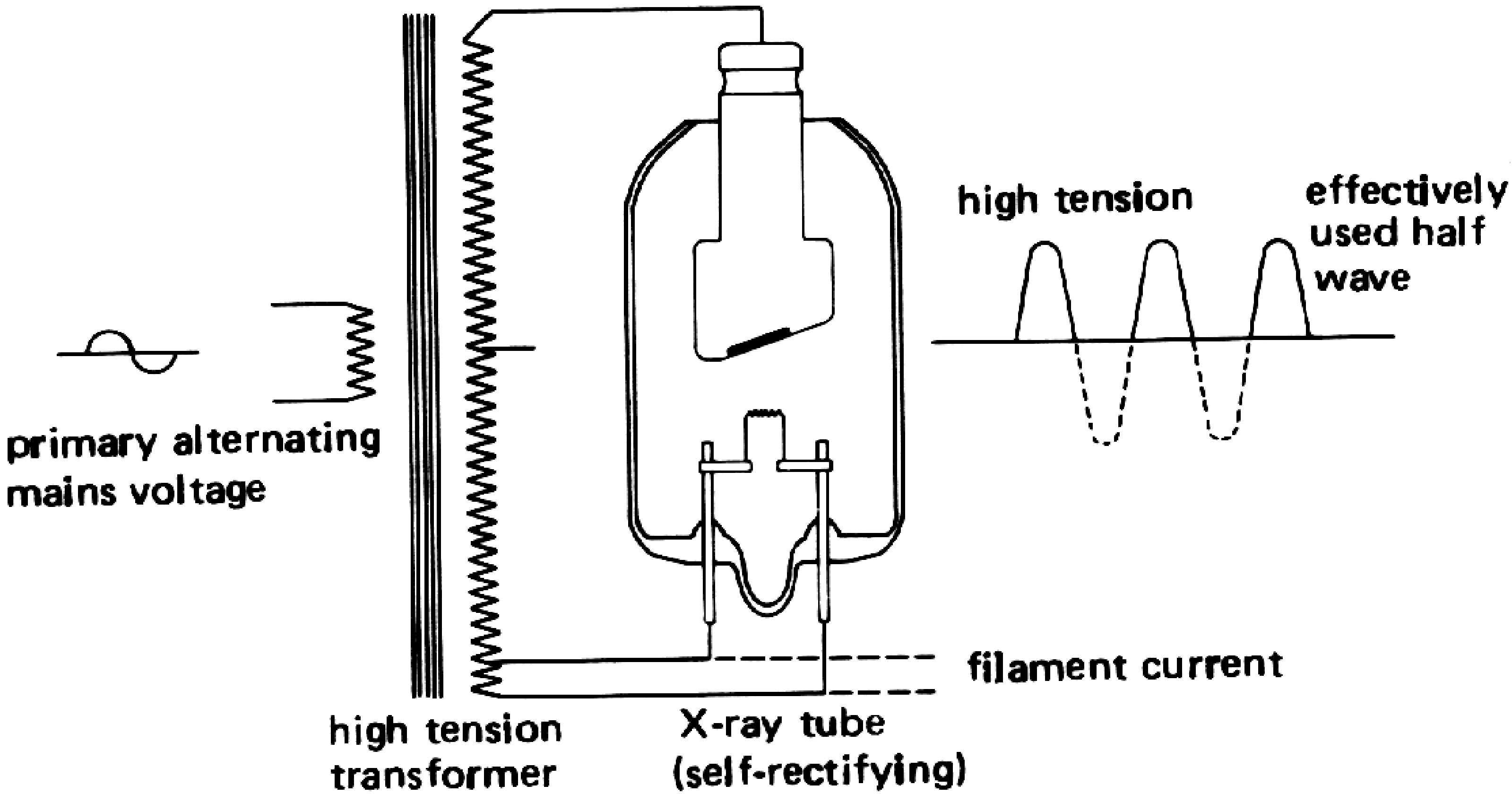
- stationary Anode
- avoid overheating
- tube duty cycle:
  - typical: 1:30 intaroral
  - 1:10 OPG
  - 420 mAs/hr intraoral



Tube Head


H.T.
transformer
radiation
protective
layer
oil expansion
bellows
lead
restricting diaphragm
X-ray tube
1·5-2·5mm
aluminium filter
collimating device

Generator Circuit

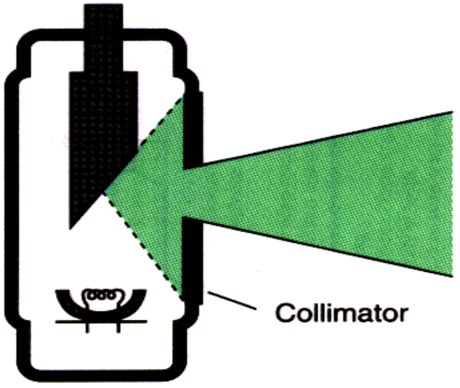


Generators & Pre-Heat

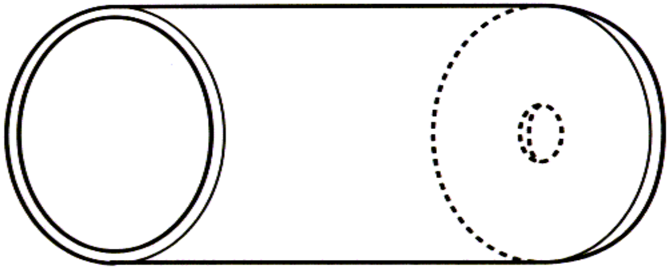
- ***Medium frequency*** - stable waveform
- ***Single phase (SP)*** - pulsed
- ***Pre-Heat***: separate circuit for heating filament
- Single Phase units without a pre-heat circuit
  - initial pulses of variable kV

Collimator

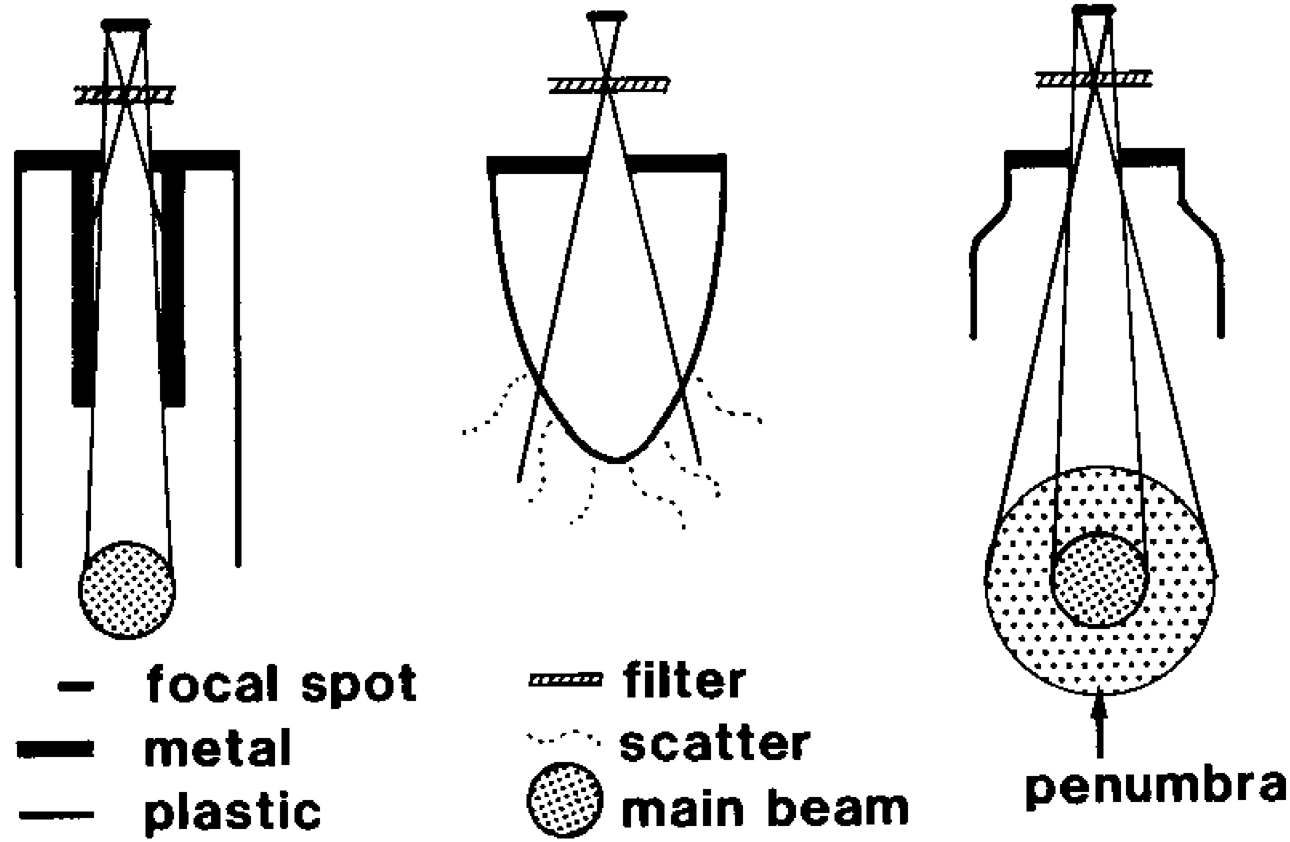
1. Lead Collimator with central hole



2. Spacer Tube (cone, position indicating device or PID)



Cones

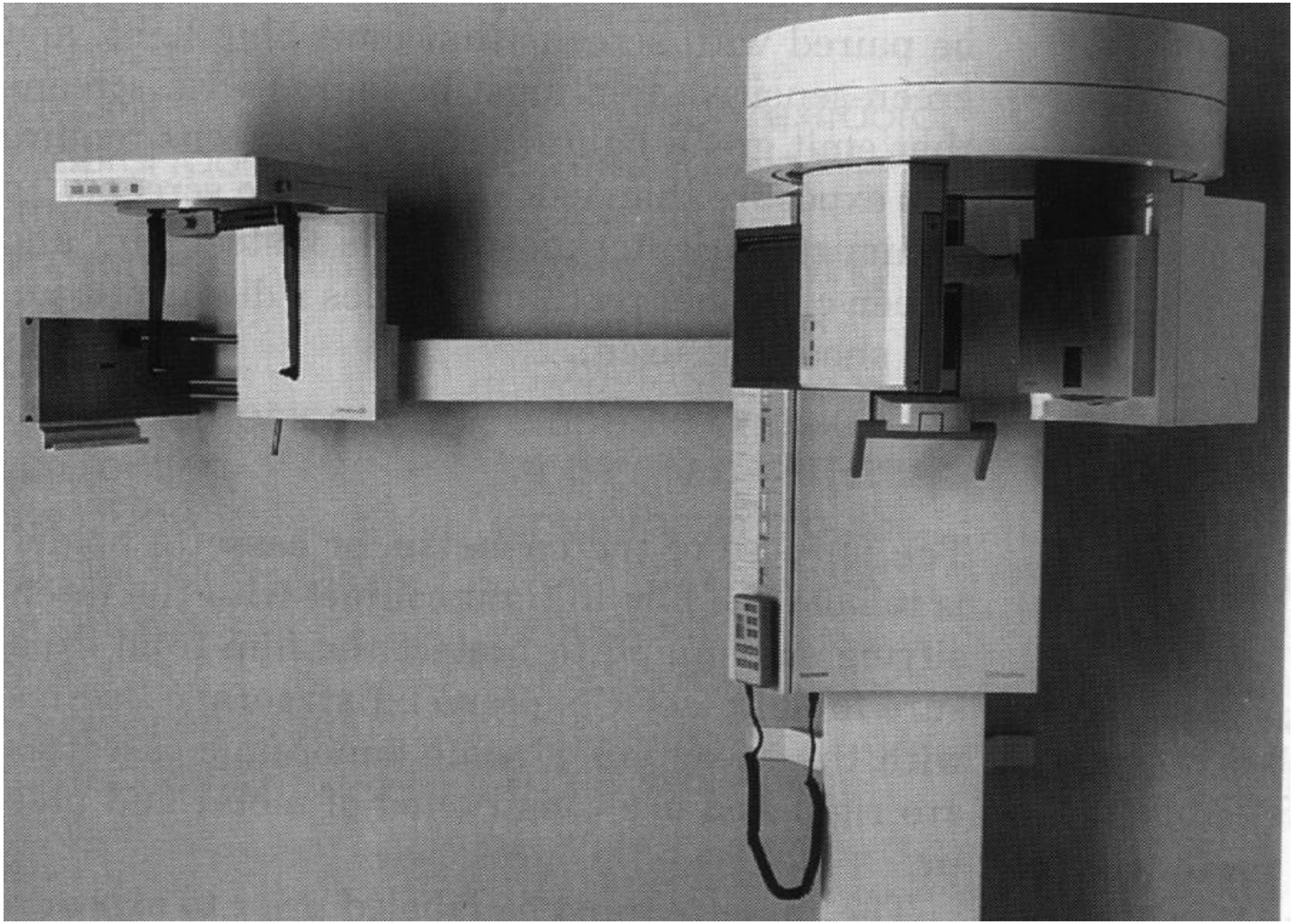


Good *Bad* *Bad*

Cone (PID) Length and Collimation

- Three cone (source-to-skin) distances– 8”, 12”, and 16”
  - Longer distance improves image sharpness, reduces dose
- Circular vs rectangular collimation
  - Rectangular– smaller field irradiated
    - Results in lower dose
    - Less scattered radiation
    - Increased contrast
    - But more difficult to position

Cephalometric Holder



Intra-Oral Dental X-Ray Equipment (technical data)

Exposure time	from 30 ms to 2.5 s
Tube	Min. 50 kV, ~7mA, Typically 70 kV
Focal spot size	0.4 to 0.7 mm
Inherent filtration	~2 mm Al equivalent
Focus-skin distance	20, 30, or 40 cm
Irradiated field	28 cm^2 with round section, 6 cm diameter collimator Rectangular also available

Panoramic X-Ray Equipment (technical data)

Focal spot	0.5 mm
kV	60 - 80 kV in 2 kV steps
mA	4 - 10 mA steps 4, 5, 6, 8, 10
Exposure time	12 s (standard projections) 0.16 - 3.2 s (cephalometric projections)
Flat panoramic cassette	15x30 cm (Lanex Regular screens))

Image Receptors in Dental Radiology

Intraoral Radiology

- **Small films (2 x 3 or 3 x 4 cm) in light-tight envelopes (no screen)**
- **Digital intraoral sensors - compared with category F film, the radiation dose is reduced by 60%.**

Panoramic Radiology and Cephalometry

- **Screen-film combination**
- **Digital sensors - compared with screen-film sensitivity class 200, the radiation dose is reduced by 50-70%.**

Dental Radiology Film Types

Sensitivity class D

- Good spatial resolution
- Typical delivered dose: about 0.5 mGy
- Typical exposure times: 0.3 - 0.7 s

Sensitivity class E, E-F, or F

- Good spatial resolution
- Typical delivered dose: about 0.25 mGy
- Typical exposure times: 0.1 - 0.3 s

Image quality of D, E, E-F, F films similar

Thank you