Introduction to Medical Imaging

Radiography

Klaus Mueller

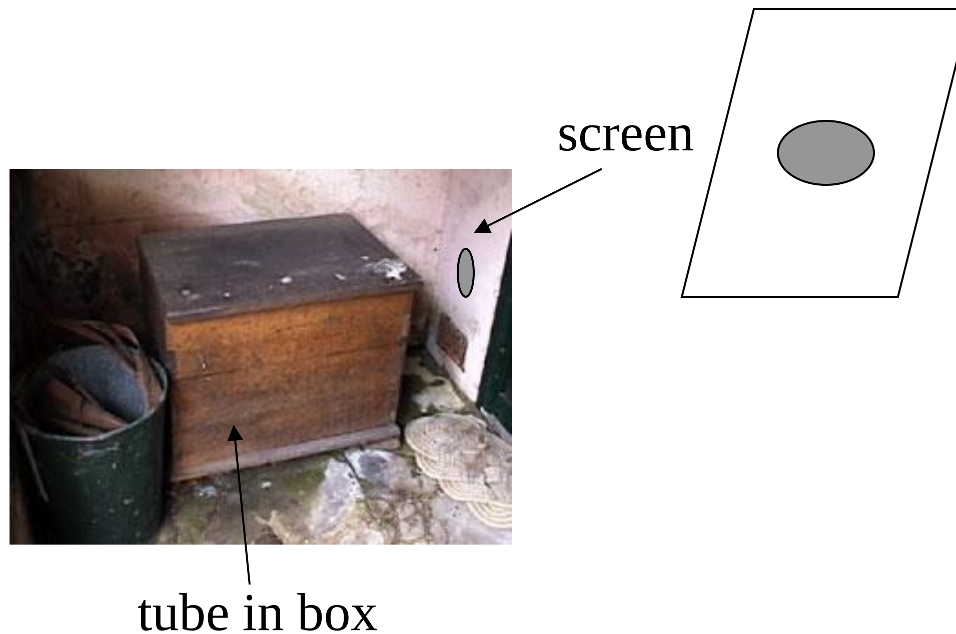
Computer Science Department

Stony Brook University

X-Ray Discovery

Discovered by Wilhelm Röntgen in 1895

- accidentally, when performing experiments with cathode tubes and fluorescent screens
- the “light” even illuminated the screen when the tube was placed into a box
- he called this new type of radiation *X-rays* (*X* for unknown)
- these X-rays could travel through all kinds of materials, at different material-specific attenuations

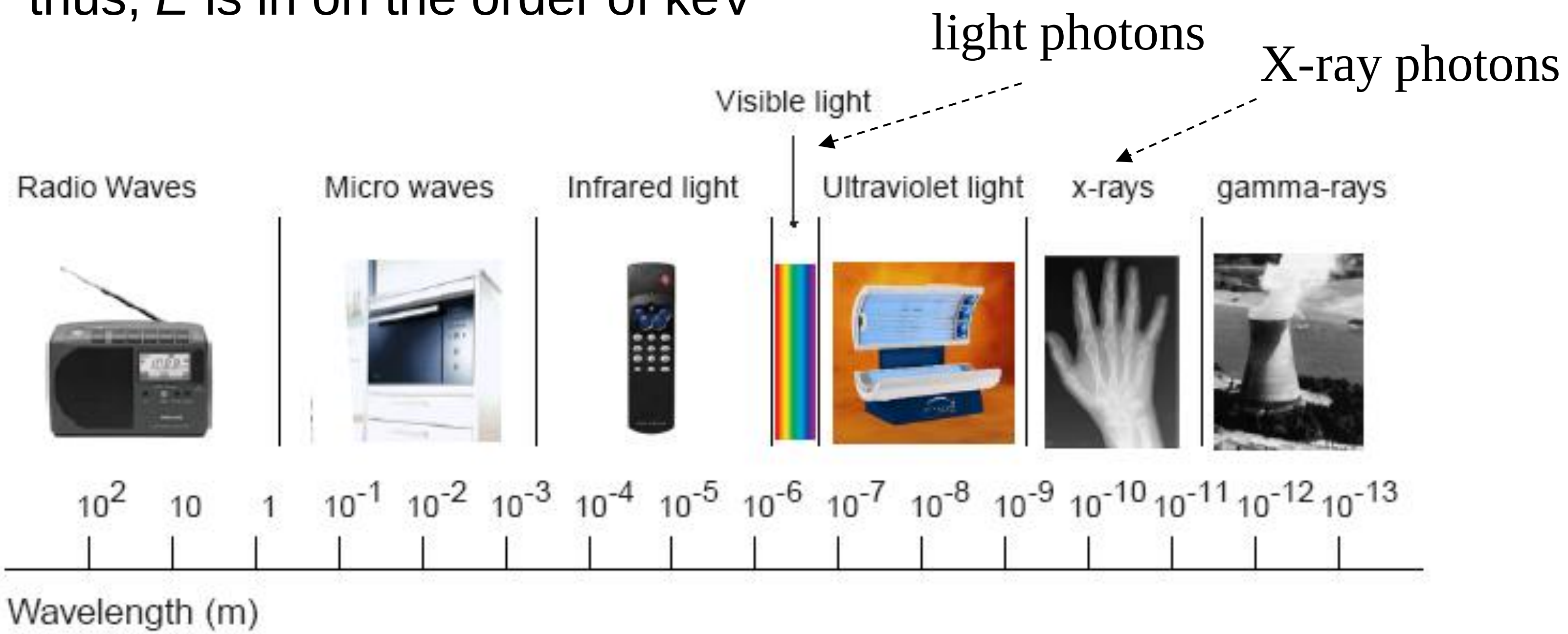


X-Ray Physics

X-rays are electromagnetic waves, consisting of *photons*

- energy is given by: $$E = h \cdot f = \frac{hc}{\lambda}$$

h: Plank's constant (4.135 10^{-15} eVs)
c: speed of light (300 10^{6} m/s)
λ: wavelength (on the order of 10^{-10} m)
thus, *E* is in on the order of keV



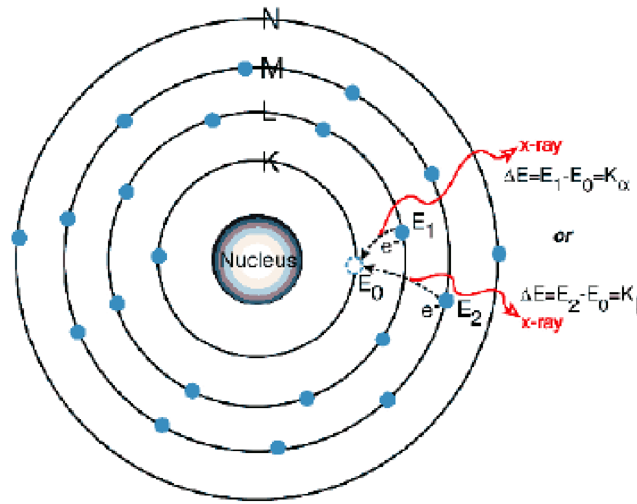
X-Ray Generation

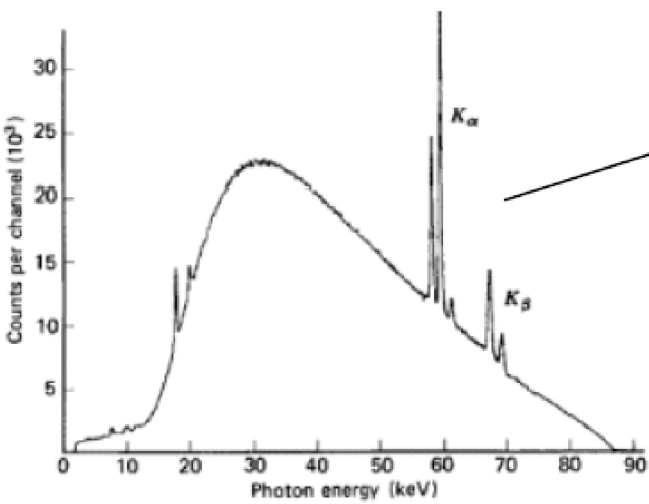
Electrons hitting anode release their energy via *Bremsstrahlung*

- gives rise to a continuous spectrum
- specific peaks arise at specific orbital shell energies (*characteristic radiation*) when anode L-electrons drop back into the K-shell

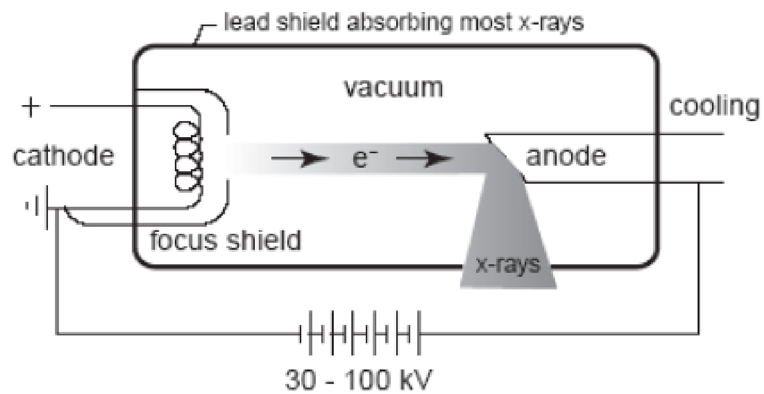
Important X-ray tube parameters:

- amount of emitted photons (6-100 mA)
- energy of emitted photons (determined by $V_{cathode\text{-}anode}$, 50-125 kV)





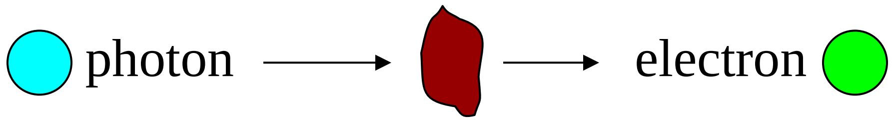
The spectrum from a tungsten target x-ray tube operated at 87 kVp.



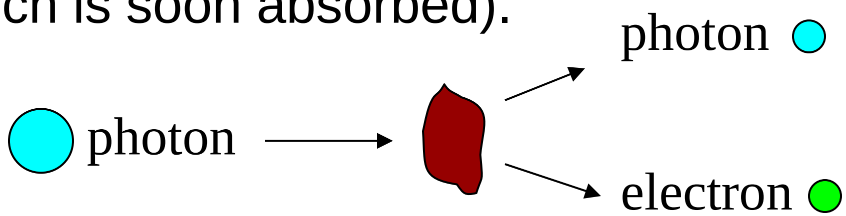
X Ray Interaction With Matter

Three types of interaction with matter:

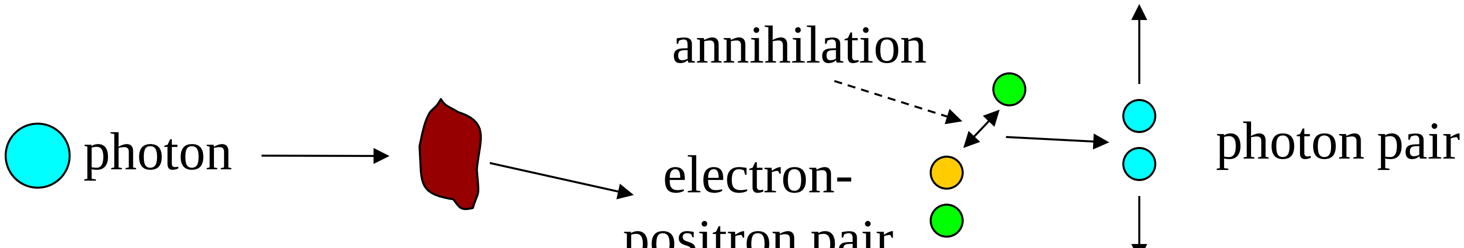
- *photo-electric absorption*: absorption of a photon by an atom and release of an electron along the same direction (which is soon absorbed)



- *Compton scattering*: only partial absorption of photon energy. The photons changes direction (at lower energy) and an electron also gets released (which is soon absorbed).



- *Pair production*: when photon energy > 1.02 MeV, an electron-positron pair may form. Soon, the positron annihilates with another electron. Two photons form, flying in two opposite directions (used in nuclear imaging)



Notes on X-Ray Interaction

Electrons soon after recombine with other atoms in tissue

- will NOT be detected in image generation (on the X-ray detector)

Photo-electric effect most desirable in radiography

- absorbs photon completely → weakens the energy along that ray
- denser tissue (such as bone) absorbs more photons → less energy arrives at the detector
- less dense tissue (such as muscle or air) absorbs less photons → more energy arrives at the detector
- this controls image formation and contrast

Compton effect less desirable

- emitted photon traveling along diverted path may get detected on detector → non-linear ray
- since we assume linear rays this is problematic
- the photons due to Compton scattering are perceived as noise

Pair production only in high-energy X-ray

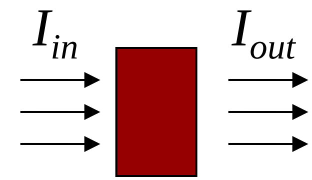
- desirable in function imaging (see later)

X-Ray Interaction With Tissue

Basic attenuation equation:

$\mu(x)$: attenuation at location x

$$I_{out} = I_{in} e^{-\int_{x_{in}}^{x_{out}} \mu(x)dx}$$



In practice, an X-ray beam comprises photons at a spectrum of energies:

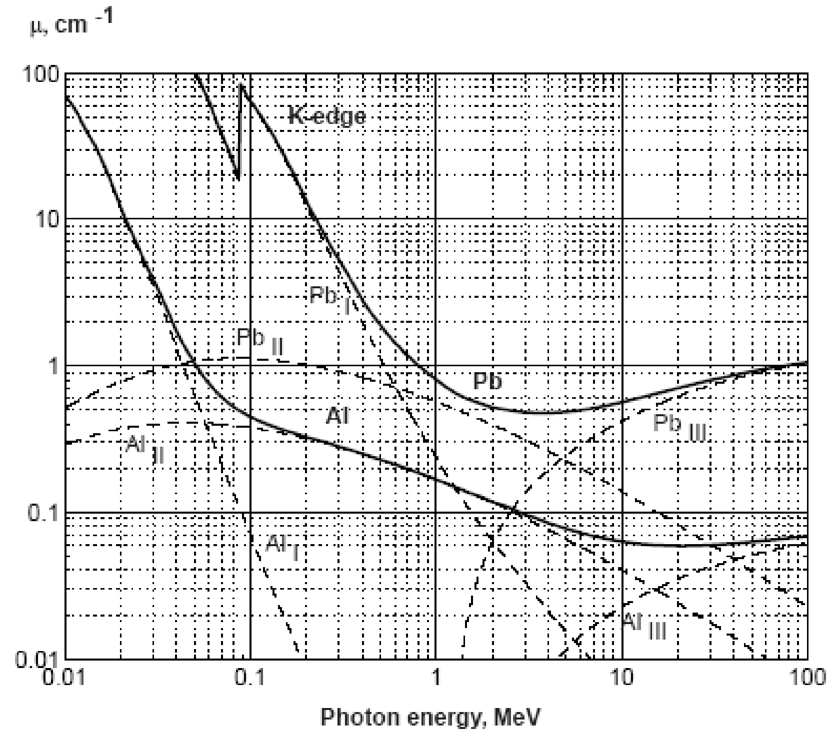
$$I_{in} = \int_0^\infty \sigma(E)dE \qquad I_{out} = \int_0^\infty \sigma(E) e^{-\int_{x_{in}}^{x_{out}} \mu(E,x)dx} dE$$

- and the attenuation equation becomes:

Interaction effects at different energies:

- low: photo-electric (I) dominates
- intermediate: Compton (II)
- high: pair production (III)

- Al: aluminum
- Pb: lead



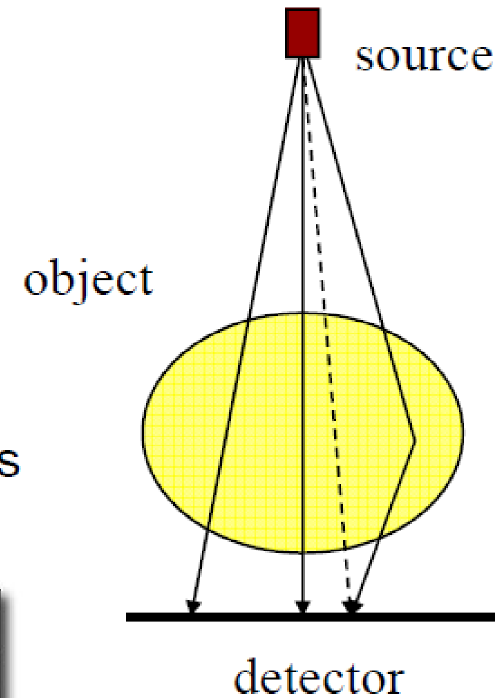
Scattered Radiation

Scattered radiation is due to Compton scattering

- dominates effects at energies >26keV (at 26keV photo-el = Compton)
- dense materials (such as bone) threshold higher

Scattered photons are detrimental to imaging

- they violate the straight ray assumption
- tend to under-estimate attenuation



Quantified by SPR: Scatter/Primary Ratio

- detected radiation to primary vs. scattered photons
- low SPR diminishes contrast

without scatter

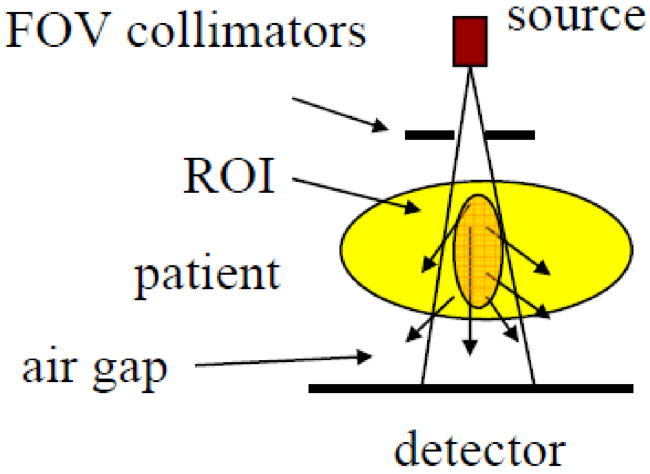
with scatter

Scattered Radiation

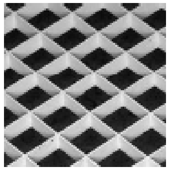
But scattered radiation has less energy than direct radiation

Depends on

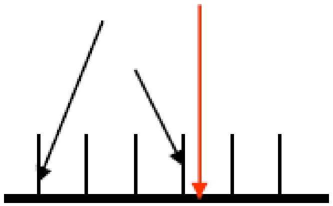
- energy of the x-rays (↑)
- patient thickness (↑) → in abdominal imaging SPR>3
- field of view FOV (↑) → want to reduce FOV to region of interest (ROI) as much as possible
- air gap between patient and screen (↓)
  but, air gap reduces resolution and FOV



Anti-scatter grid:



- fixed on detectors
- shields off scattered photons
- longer teeth provide:
  - more scatter reduction, but also…
  - fewer true photons → less SNR



X-Ray Detectors: Screen-Film

Quantum Efficiency (QE):

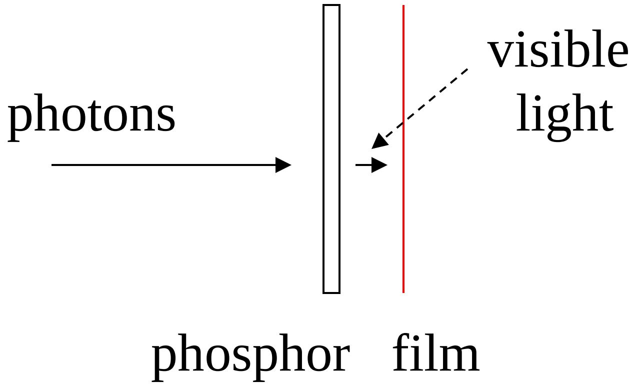
$$QE = \frac{\text{detected photons}}{\text{incoming photons}} 100 \quad \text{percent}$$

Photographic film: very inefficient (QE=2%)

- would require huge patient doses

Phosphor-based: Place film between two intensifying fluorescent screens

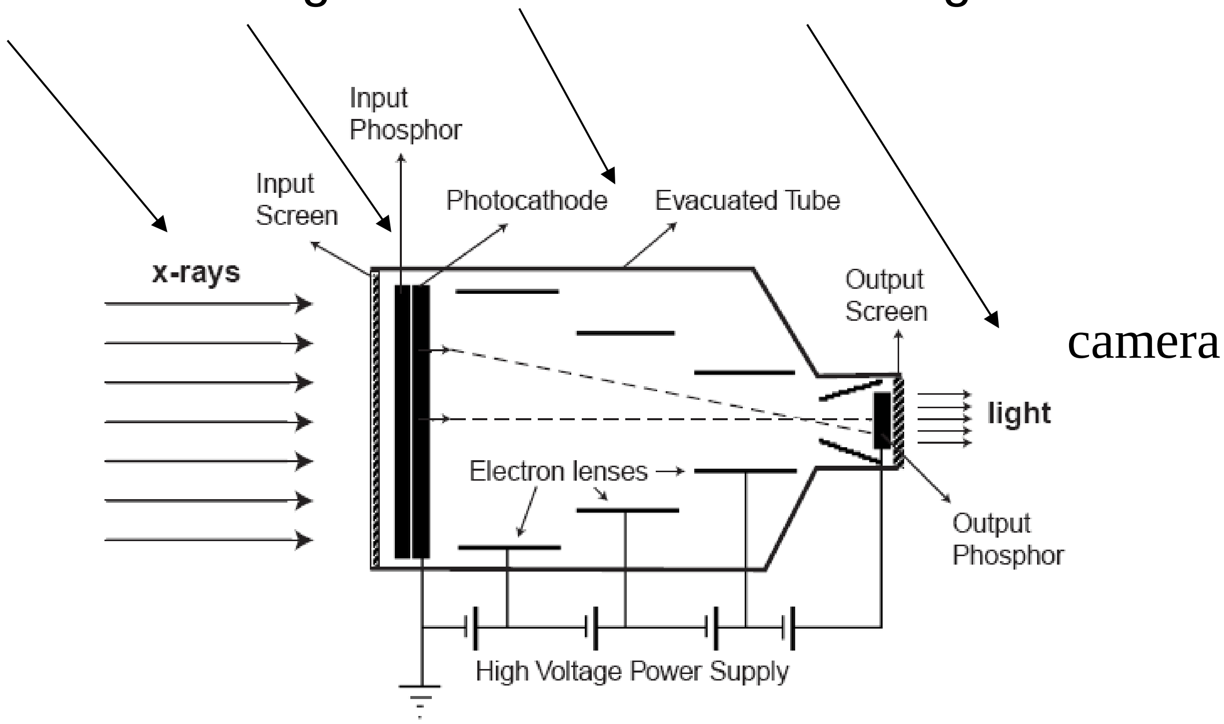
- made out of rare earth phosphors (gadolinium oxysulfide Gd_2O_2S)
- phosphor converts X-rays to scattered visible light
- light directed toward film is recorded (QE=25%)



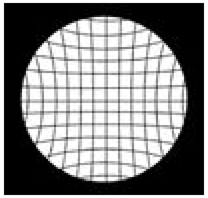
X-Ray Detectors: Image Intensifier

Image intensifier:

- photons → visible light → electrons → visible light



- limited spatial resolution due to limited camera resolution
- elevated noise due to additional conversions
- geometric distortions (*pin-cushion* distortion)



X-Ray Detectors: Storage Phosphors

Exposure: trap electrons in the conduction band (electrons cannot fall back into valence band and emit light)

Readout:

- pixel-wise scanning with a laser beam (electrons fall back into valence band, light is emitted)
- capture light with optic array
- transmit to photo-multiplier (converts light into electrical signal)
- direct analog signal to an A/D converted (generates bit-stream)
- Digital image is now available for storage, further processing

Clear: subject plate to strong light source

Advantages:

- linear detector response (while film follows an S-curve)
- allows efficient digital mass storage
- allows use in *Picture Archiving and Communication Systems* (PACS)

X-Ray Detectors: Direct Radiography

Shortcomings of image intensifier detectors

- camera was made out of Si-crystal technology, restricting its size to a small area (just like CCDs)
- this required the long chain from photons to camera (see before)

Newer (scintillator) technology: hydrogenated amorphous silicon detectors (a-SI:H)

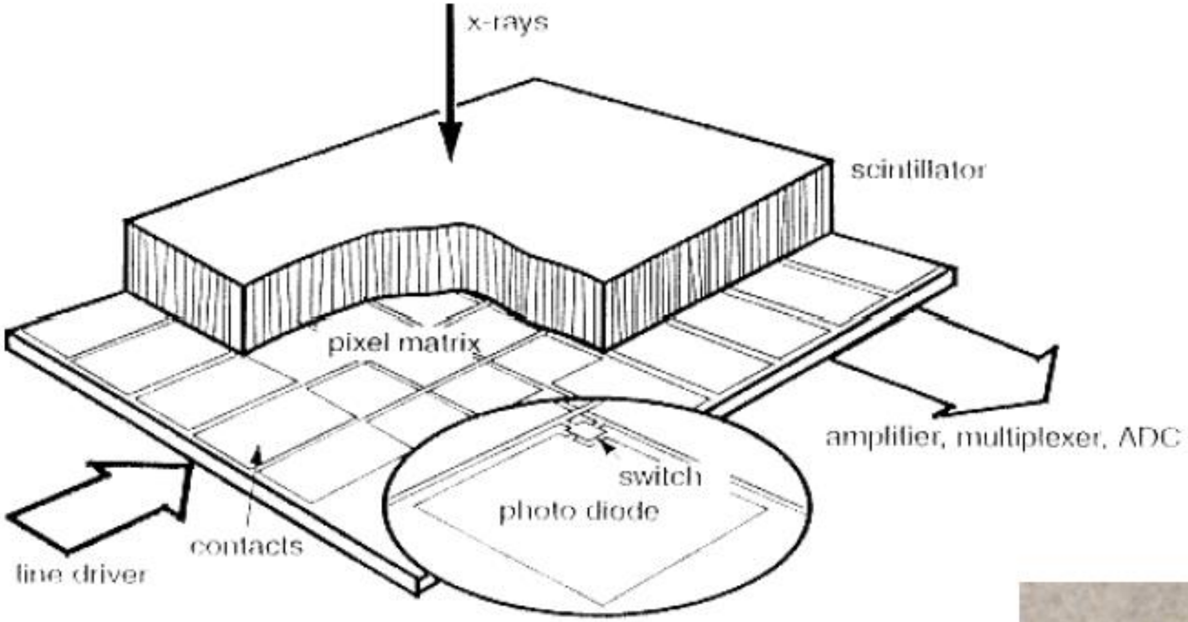
- can be manufactured in flat, large sheets
- can be coupled directly with the phosphor plate
- but still need to convert photons to visible light, affecting resolution

Latest technology: amorphous selenium (a-Se)

- a photo-conducting layer (not a phosphor)
- a-Se electrical conductivity proportional to radiation energy
- before exposure: a homogenous charge is applied to Se-surface
- during exposure: photons are absorbed in the Se-layer, setting free electrons → electrons neutralize charge locally (pixels)
- resulting image can then be read by a photo-conductor matrix
- high QE and resolution (11-13 lp/mm, lp=line pairs=half-pixels)

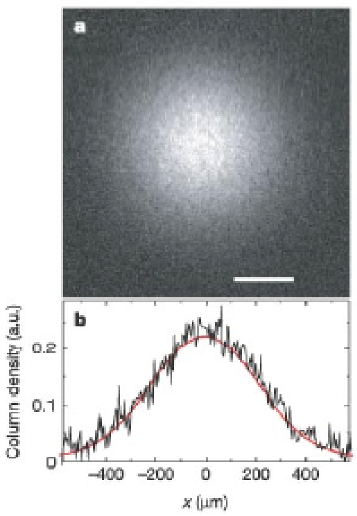
Flat Panel Amorphous Silicon Detectors

Has become the standard detector technology:



- resolution: 120-140 µm
- high sensitivity enables near-real-time imaging
- low noise

Quantum Noise



X-ray beam has a *quantum structure*

- each photon carries a specific energy quantum

Photons in a beam are independent and distributed in a random manner

- just like individual rain drops, they form clusters
- but as more drops gather, the distribution becomes more uniform

Quantum noise follows the statistical law: $\sigma = \sqrt{N}$

Thus,

$$SNR = \frac{N}{\sigma} = \frac{N}{\sqrt{N}} = \sqrt{N}$$

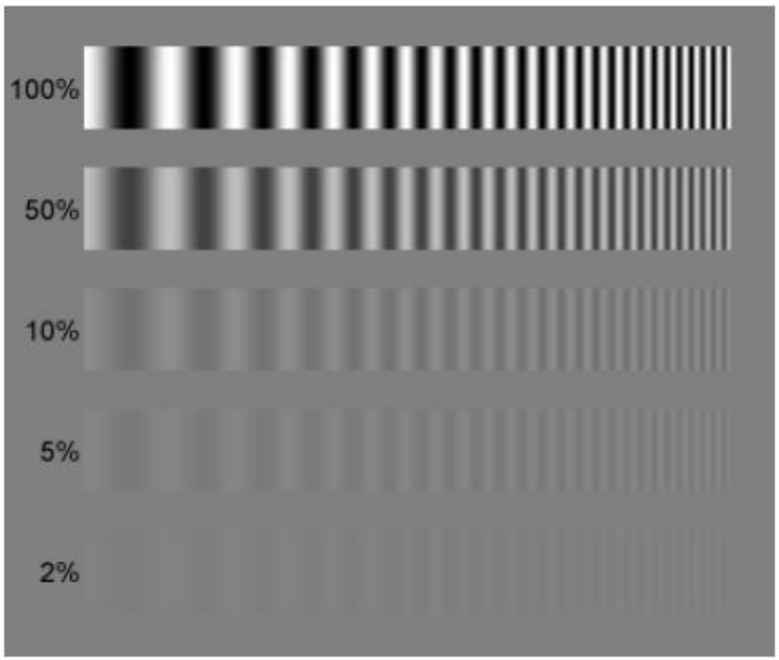
SNR improves as the number of photons N increases

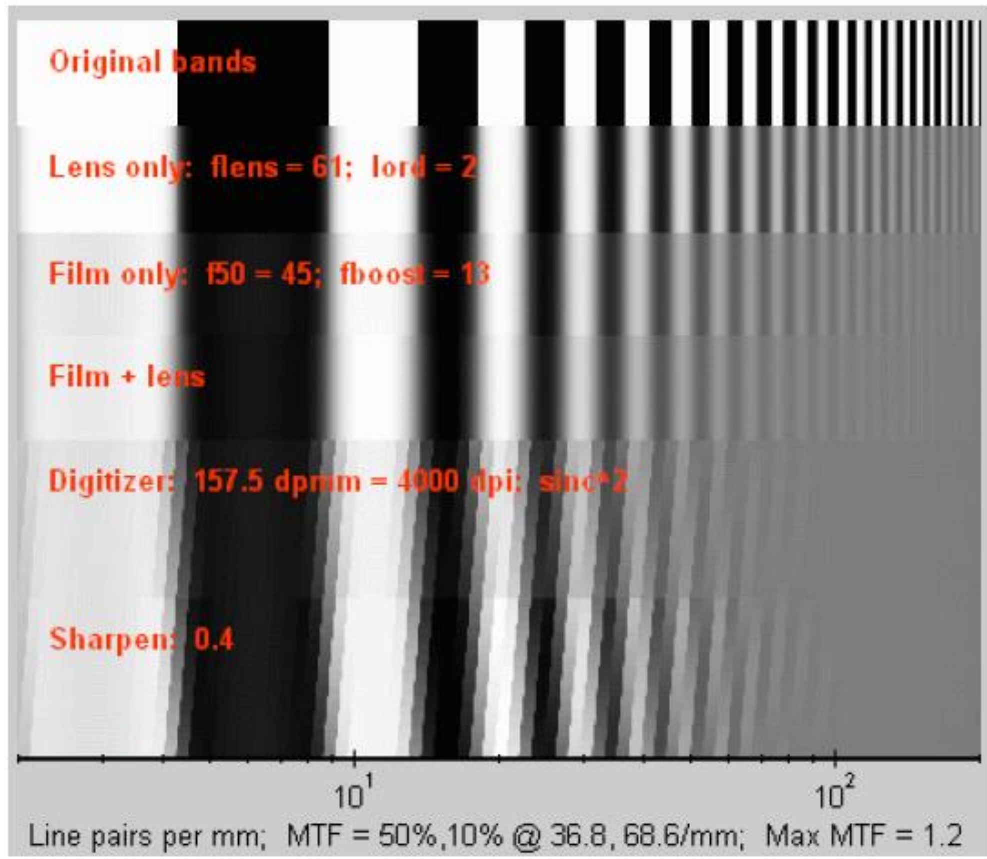
- however, this also increases patient dose
- so there is a trade-off
- doubling SNR increases dose by a factor of 4

Interlude: Modulation Transfer Function

Measures the ability of a sensor to resolve (detect, provide contrast with) signals at different frequencies

- frequency measured in line pairs (lp) / mm
- detectability measured in %





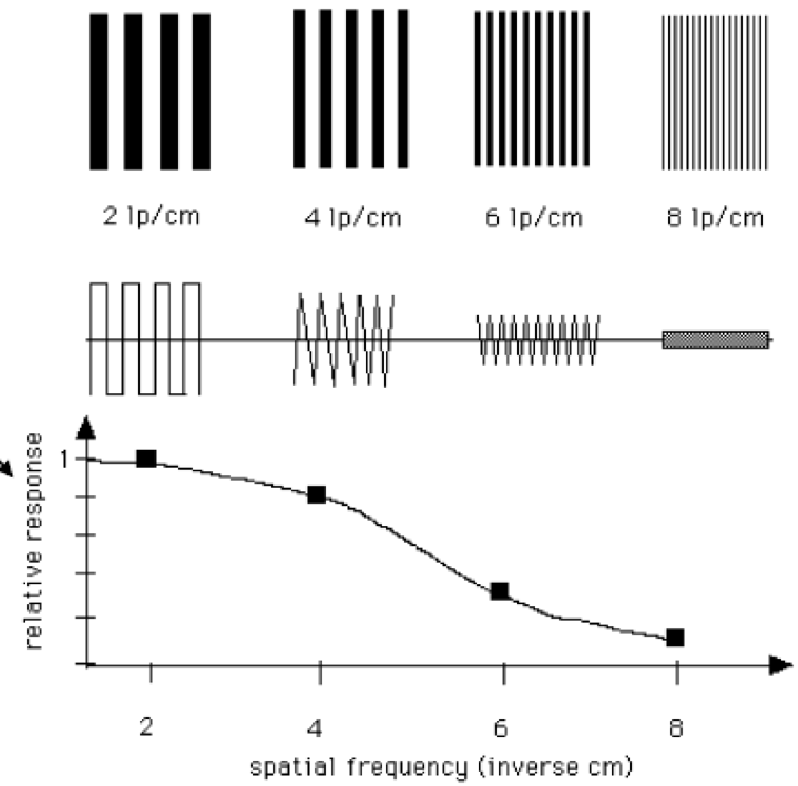
DQE: Detective Quantum Efficiency

More recent metric to rate a detection system:

- compares contrast at different frequencies with noise at that frequency

Measure contrast with the MTF

MTF: Modulation Transfer Function



Measure noise with variance

So, DQE is then (k=constant)

$$DQE(f) = \frac{k \cdot [MTF(f)]^2}{[\sigma(f)]^2}$$

Thus, DQE is an excellent metric to express dose efficiency

- want high contrast for given noise (and N)

Image Quality

Depends on:

- quality of the anode tip (finer tips give better focus)
- patient thickness (thicker patients cause more scattering, which deteriorates resolution)
- light scattering properties of the phosphor (for phosphor-based systems)
- film resolution (for film-based systems)
- sampling procedure (for systems with digital read-outs)
- spot size of the read-out laser (for systems with digital read-outs)

Resolution:

- screen-film combinations: usually the spatial resolution is sufficient (in the range of 5-15 lp/mm, 100-33μm)
- storage phosphors: sufficient for most applications, except digital mammography (in the range of 2.5-5 lp/mm, 200-100μm)
- direct radiography: needed for digital mammography

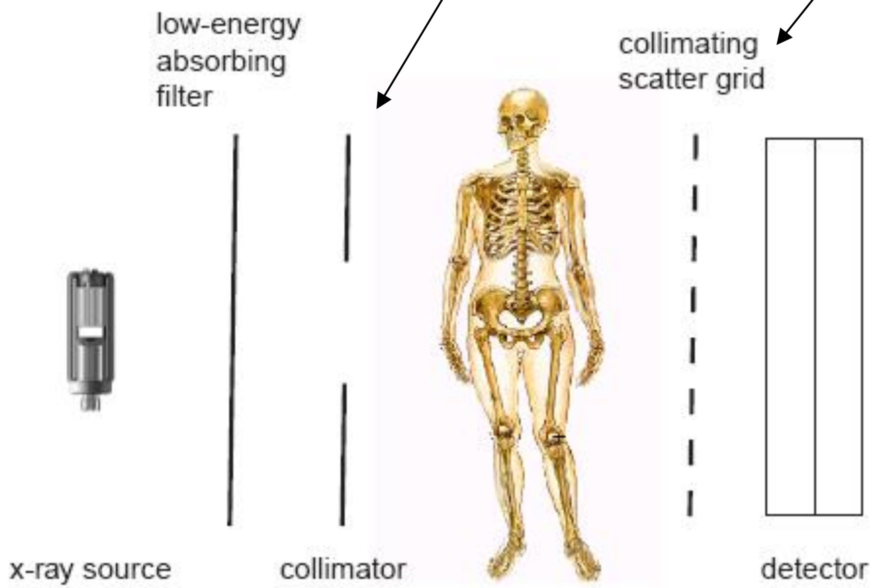
Required resolution indicates that image size $\geq 2000^2$ pixels

Available Technology: Summary

remove photons that would be absorbed in the body anyway (beam hardening)

limit radiation to patient volume of interest

remove scattered photons (tungsten)



- film-screen based
- camera + image intensifier
- storage phosphor plate
- amorphous Si flat panel
- amorphous Se direct detector

Clinical Use

Majority of clinical radiographic examinations are now digital

Mammography is somewhat behind because it requires resolutions that exceed that of storage phosphors

- direct radiography with amorphous Si is being developed

X-ray images can be static or dynamic

- static X-ray can be performed with any of the modalities
- dynamic X-ray uses image intensifier, viewed in real-time on a TV monitor

Radiographic images are made for all parts of the body

- skeletal, chest (thorax, heart), mammography (breast), dental

Fluoroscopic image sequences are produced in real time

- used in applications where motion is the subject of investigation
- intra-operative fluoroscopy (surgery, patient setup, positioning)
- guidance for minimally invasive procedures
- angiography (coronary imaging, vessels)

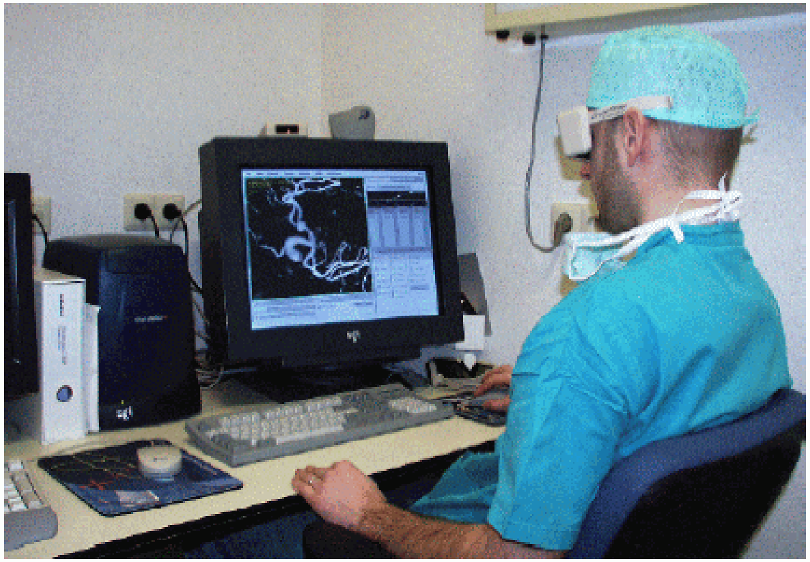
Case Studies (1)

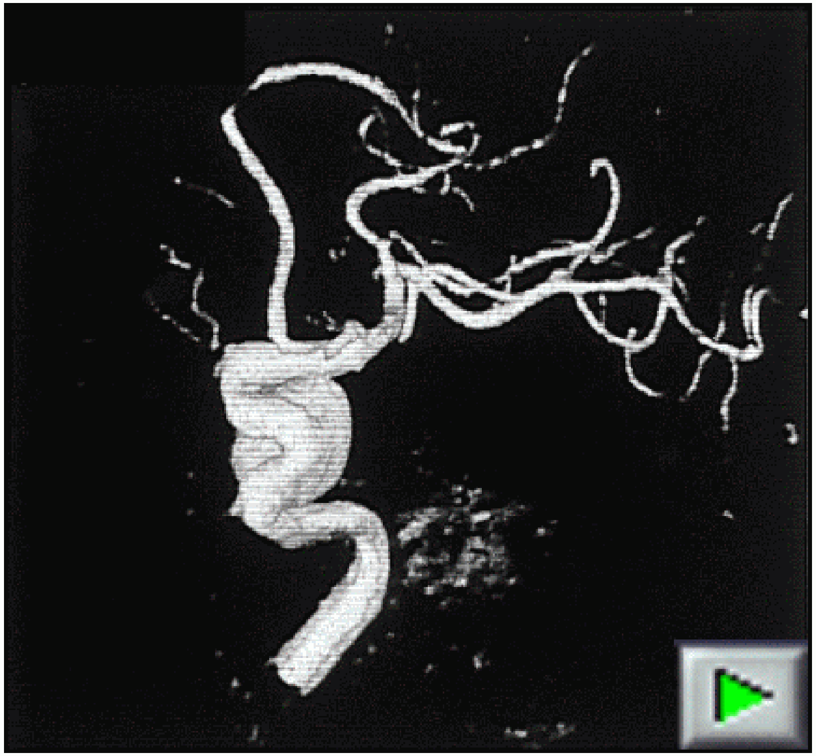
Multi-purpose radiographic room. The table can be tilted in any orientation. Both a storage phosphor and an image intensifier are available.

Case Studies (2)

3D-angiographic room: C-arm with x-ray tube and image intensifier at both ends. By rotating the C-arm on a circle around the patient a series of radiographic images are acquired that are subsequently used to compute a 3D image of the blood vessels.

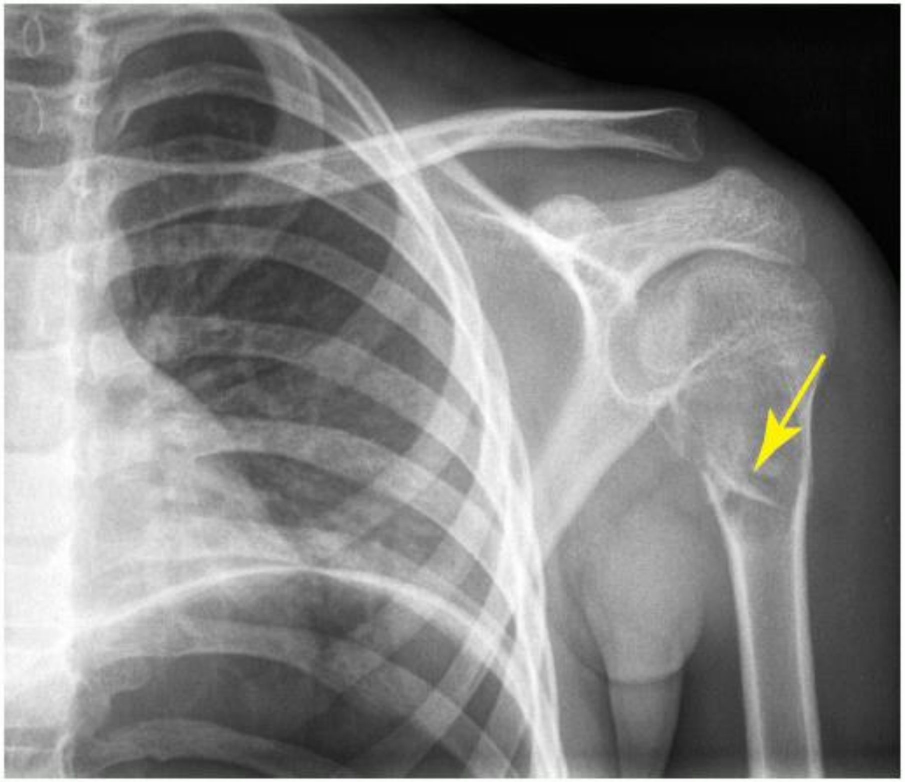
Case Studies (3)





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

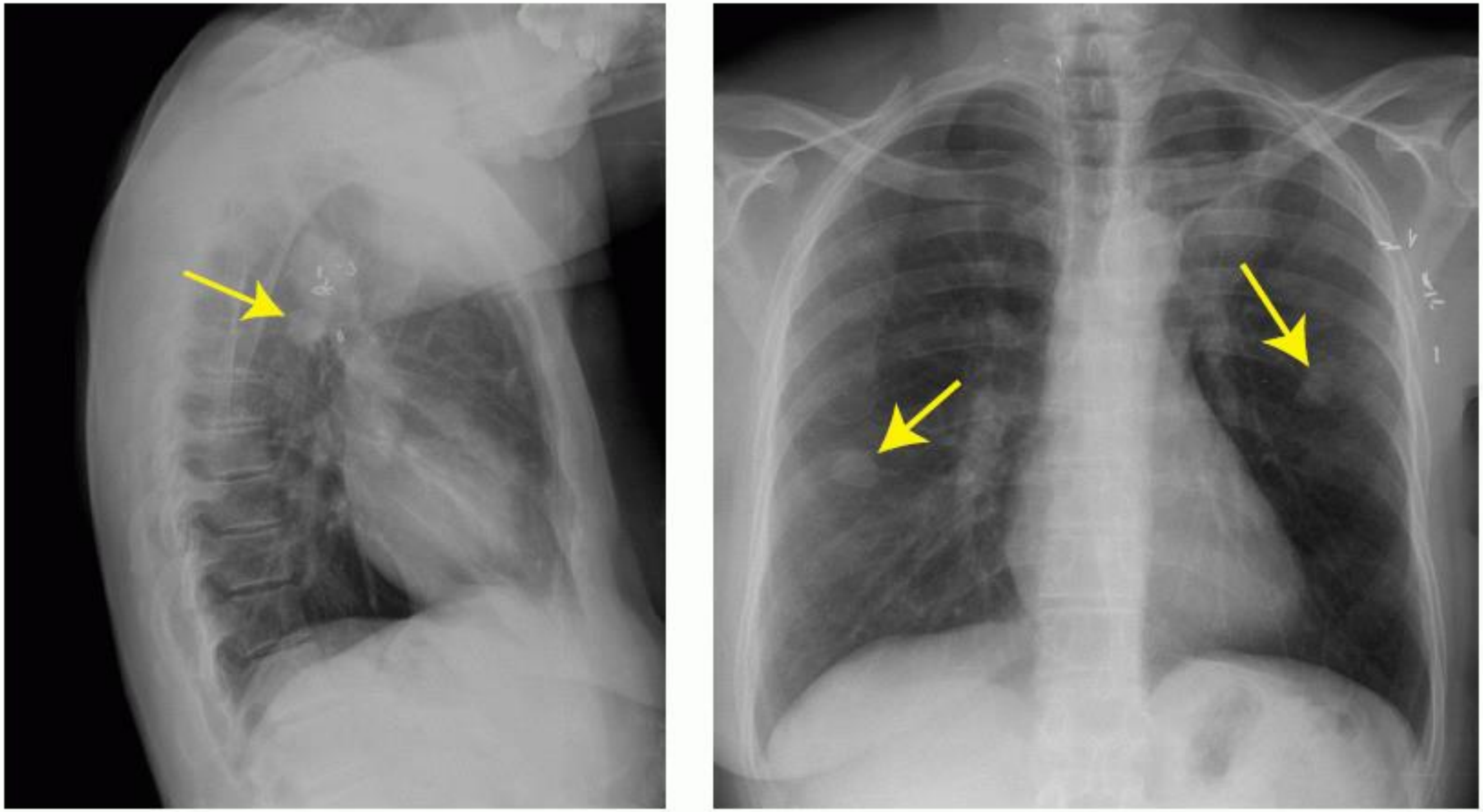
Case Studies (4)



Double mandibular fracture with strong displacement to the left.

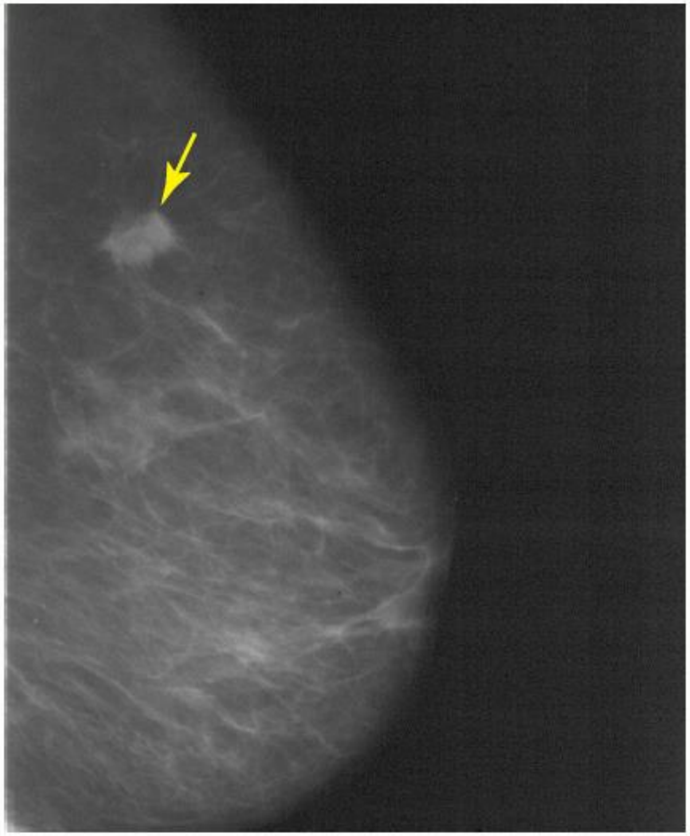
Solitary humeral bone cyst known as ”fallen leaf sign”

Case Studies (5)



Radiographic chest image showing multiple lung metastases

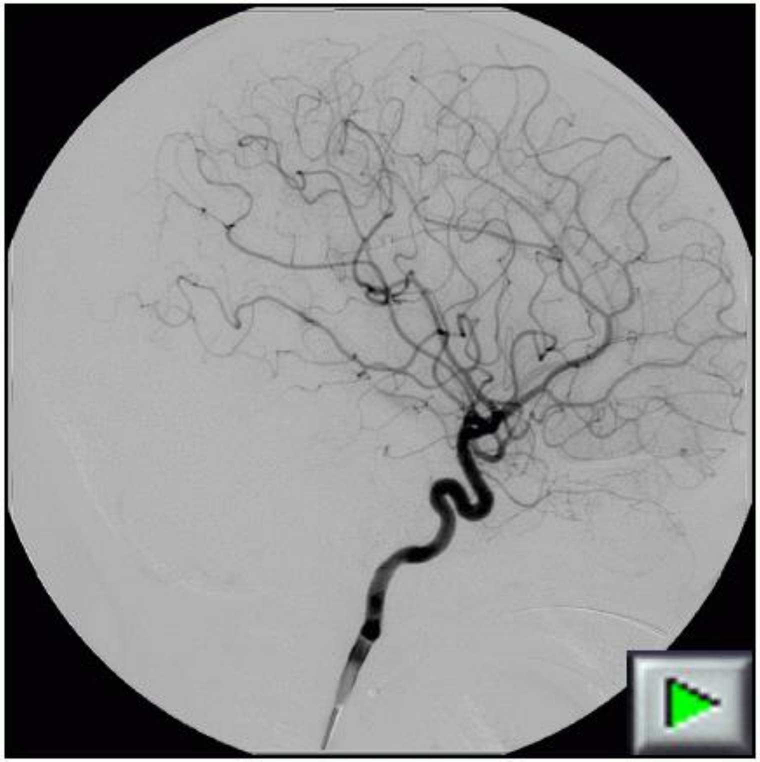
Case Studies (6)



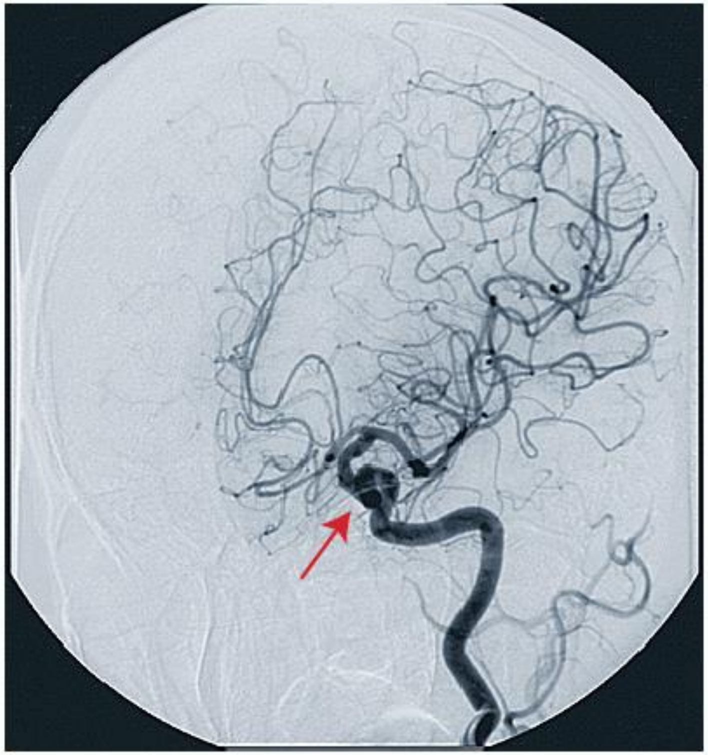
Dense opacity with spicular borders in the left breast, which suggests a malignant lesion

Postoperative fluoroscopic control of bone fixation with plate and screws after a complete fracture of the humerus

Case Studies (7)

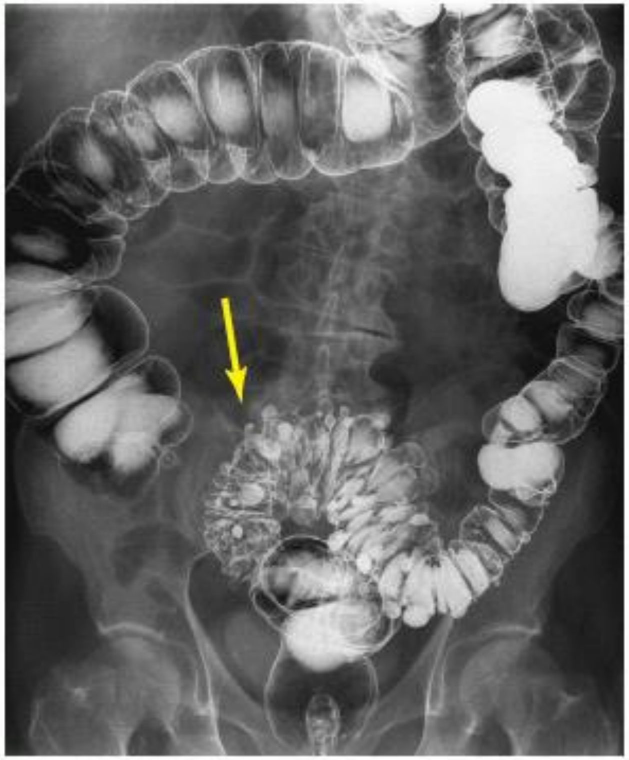


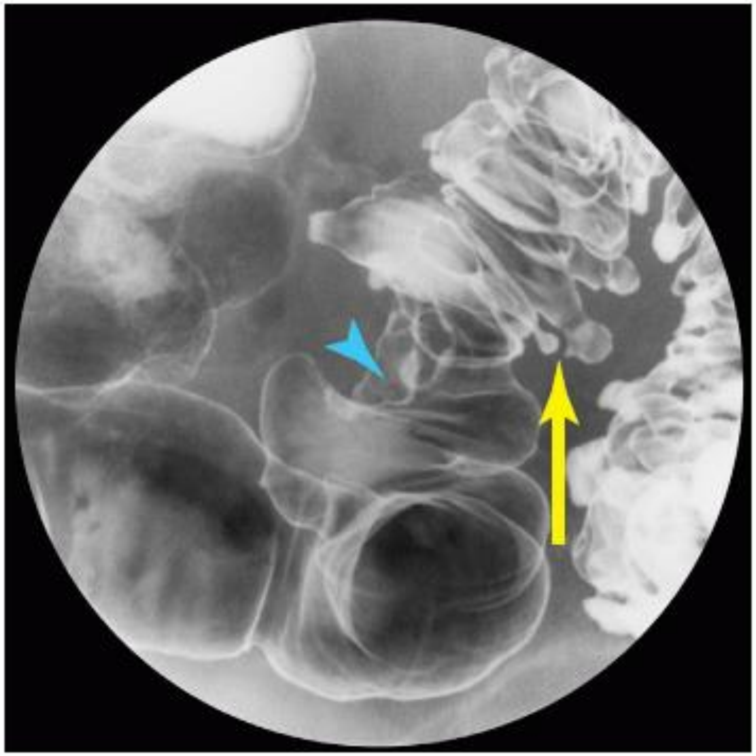
Cerebral angiogram obtained by injecting a iodine containing fluid into the arteries. The contrast dye subsequently fills the cerebral arteries, capillaries and veins.



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

Case Studies (8)





Double contrast (barium + gasinsufflation) enema with multiple diverticula in the sigmoid colon (yellow arrows). Polypoid mass proliferating intraluminal (blue arrowhead, only visible on the spotview).

Biological Effects and Safety (1)

When X-rays pass through tissue, they deliver energy

- the ionization (the removal of electrons from their nuclei) causes chemical changes to the irradiated cell

This can cause biological damage:

- destruction of the cell
- cell may lose its ability to divide
- cell may divide in uncontrolled ways (malignant growth)
- damage may be sufficiently small to enable self-repair

Absorbed radiation dose is measured in Gray (Gy)

- one Gy is an absorbed dose of 1 J/kg of irradiated material
- each organ has a specific dose: the *organ dose*

Absorbed dose also dependent on radiation weighting (quality)

- example: radioactive isotopes also emit harmful particles
- weighting is expressed as *equivalent dose*, measured in Sieverts (Sv)
- factors are 1 (X-ray) to 20 (α-particles formed by heavy isotopes)

Biological Effects and Safety (2)

Harm of dose depends on the irradiated organ

- for this, tissue weighting factors have been developed
- the *effective dose* (measured in Sv) is calculated by multiplying the equivalent dose by the tissue weighting factor
- effective dose for a patient is then the sum of all effective dose

Some tissue weighting factors

- 0.01 for skin and bone
- 0.2 for the gonads
- the sum of all weights is 1 (for a uniform dose over the whole body: effective dose = equivalent dose)

Specific effective dose examples for typical radiographic examinations

- dental X-ray: 0.01 - 002 mSv; chest X-ray: 0.01 - 0.05 mSv
- skull: 0.1 - 0.2 mSv; pelvis: 0.7 - 1.4 mSv
- lumbar spine: 1.3 - 2.0 mSv; mammography: 1.0 - 2.0 mSv
- note: many times more than one image is taken, multiplying the dose

Biological Effects and Safety (3)

Dynamic X-ray increases effective dose significantly

- increase by order 10 for diagnostic procedure
- increase by order 100 for interventional procedure

Examples:

- arteriography of the lower limbs: 6.2 mSv
- abdominal angiography: 8.2 mSv
- nephrostomy (kidney, urinary tract): 13.6 mSv
- embolization of spermatic vein: 17.3 mSv
- biliary drainage (digestive system): 38.2 mSv

Risk for cancer:

- a conservative estimate for the lifetime risk is 0.05 per Sv

Recommendations by ICRP (Intern’l Comm. Rad. Prot.) panel:

- the equivalent dose due to natural sources is 2 mSv/year
- limit additional background and indirect radiation to 1 mSv/year
- limit for personnel in medical imaging departments is 20mSv/year

Dose Quantification

Incident dose = pure dose without body

Surface dose = incident dose + scattered radiation from body

Radiation in the body = surface dose - exit dose

Image receptor dose = exit dose - loss in detector