Intelligent Analysis of Biomedical Images

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Courtesy: Some slides are adopted from CSE 377 Stony Brook University and CS 473 U. Waterloo

ChestX-ray14

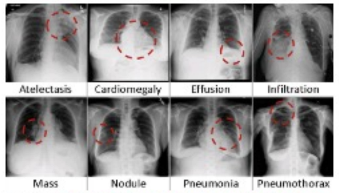
Edit

Introduced by Wang et al. in ChestX-ray8: Hospital-scale Chest X-ray Database and Benchmarks on Weakly-Supervised Classification and Localization of Common Thorax Diseases

ChestX-ray14 is a medical imaging dataset which comprises 112,120 frontal-view X-ray images of 30,805 (collected from the year of 1992 to 2015) unique patients with the text-mined fourteen common disease labels, mined from the text radiological reports via NLP techniques. It expands on ChestX-ray8 by adding six additional thorax diseases: Edema, Emphysema, Fibrosis, Pleural Thickening and Hernia.

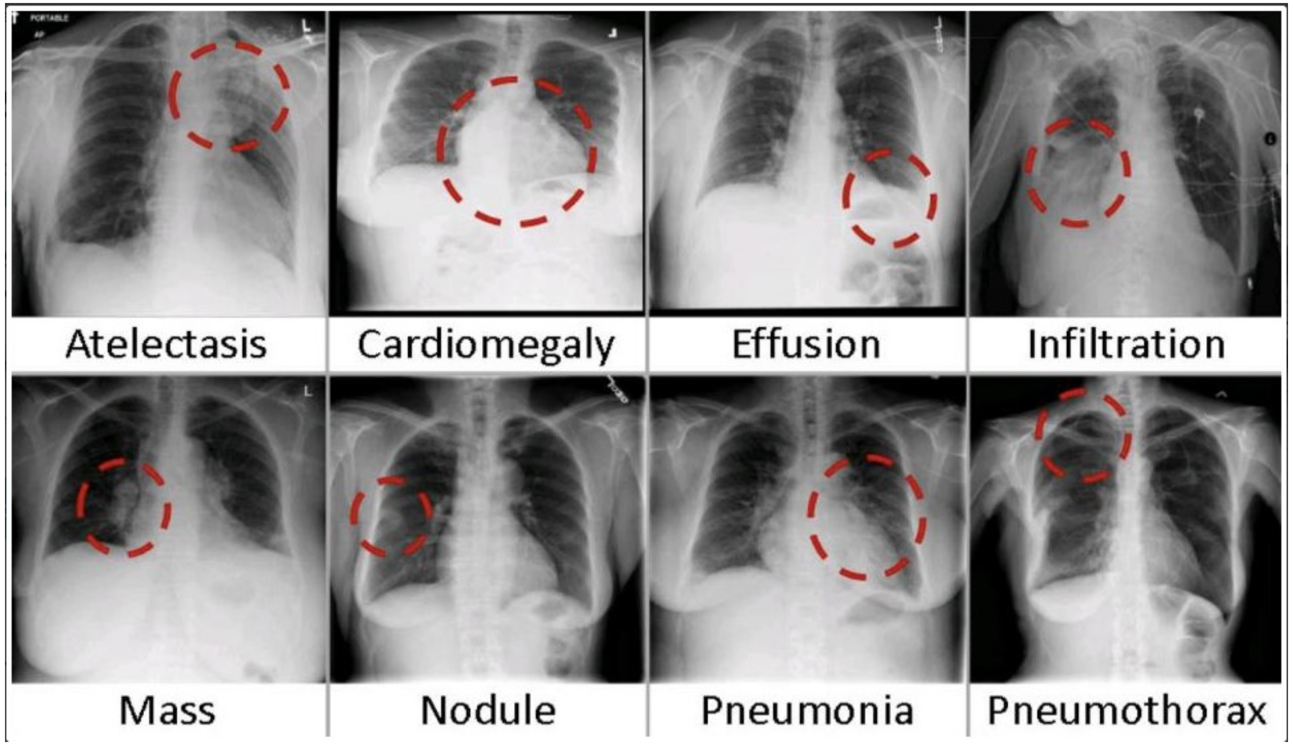
Source: https://nihcc.app.box.com/v/ChestXray-NIHCC/file/220660789610

Homepage

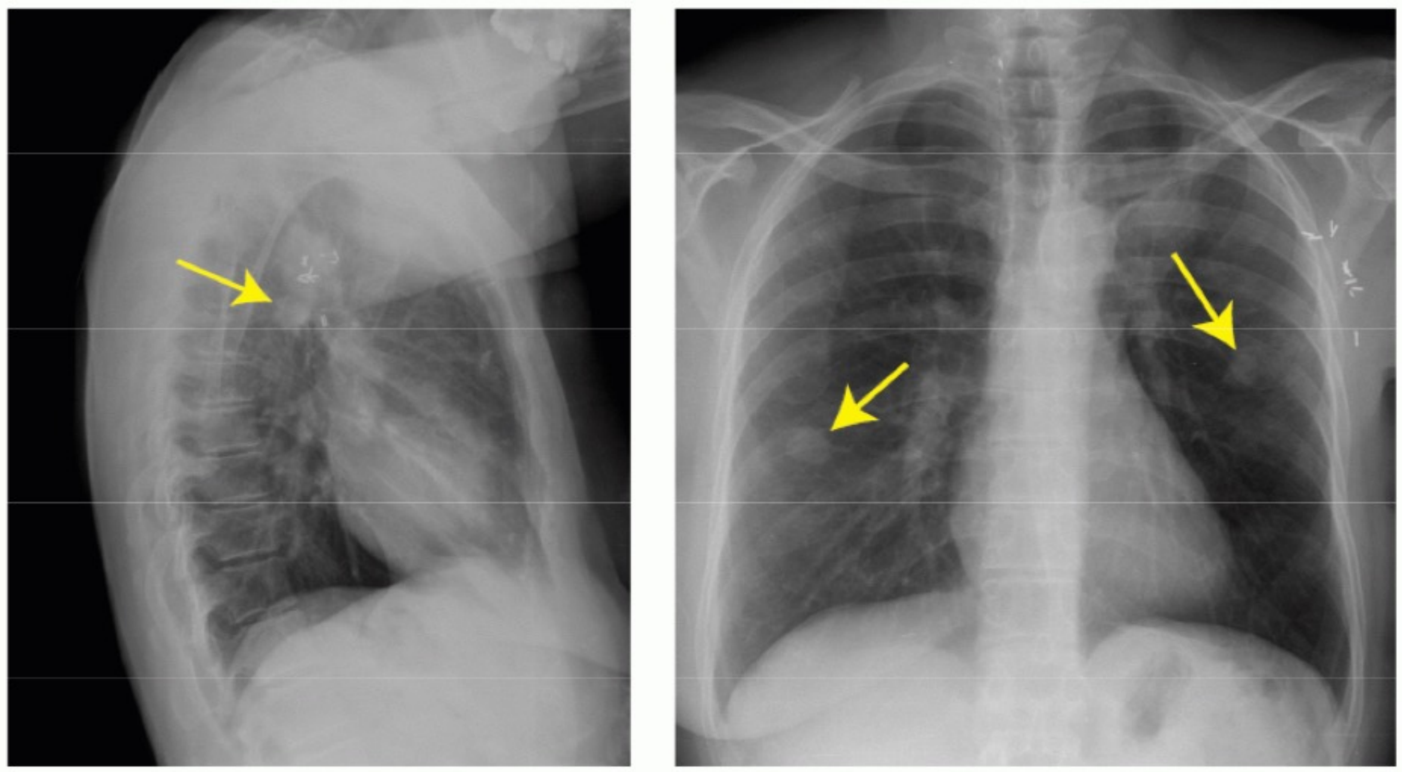


Source: https://nihcc.app.box.com/v/Ches...

Usage



Case Studies



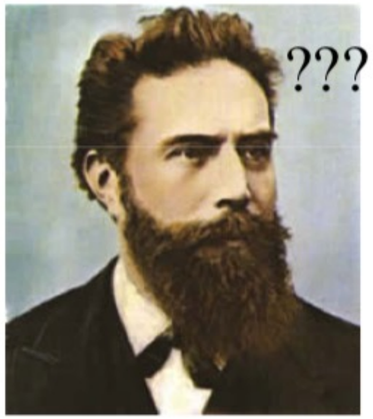
Radiographic chest image showing multiple lung metastases

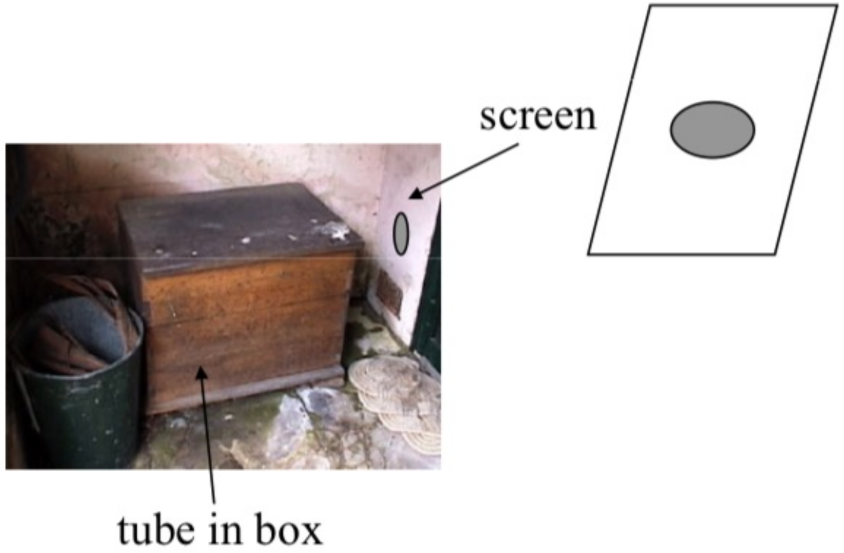
Radiography

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



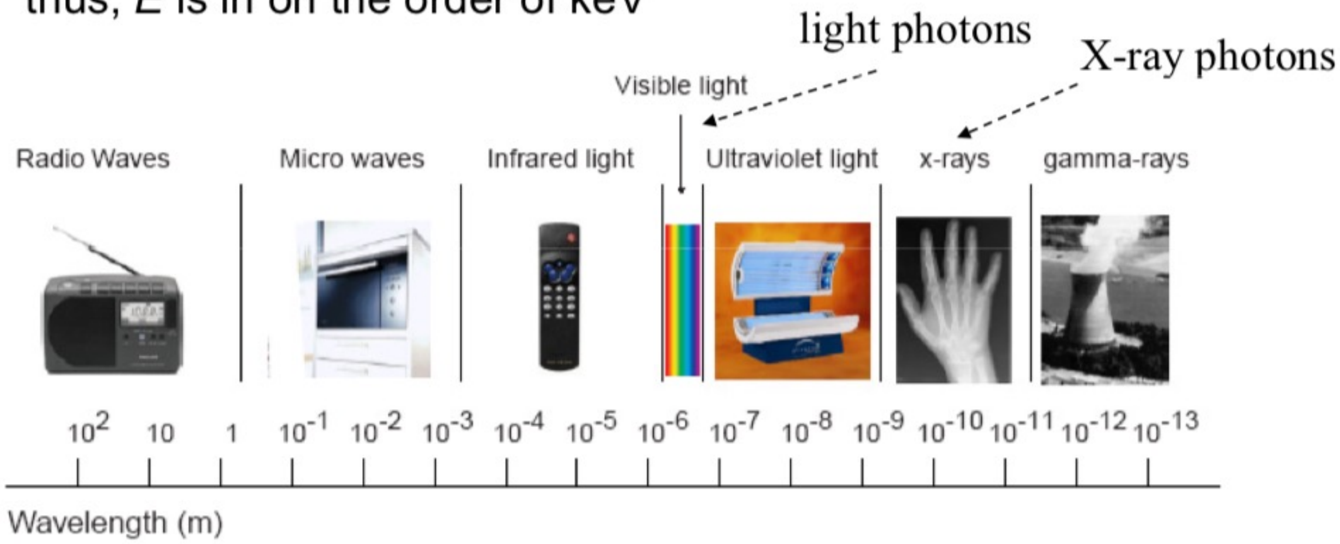


X-Ray Physics

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



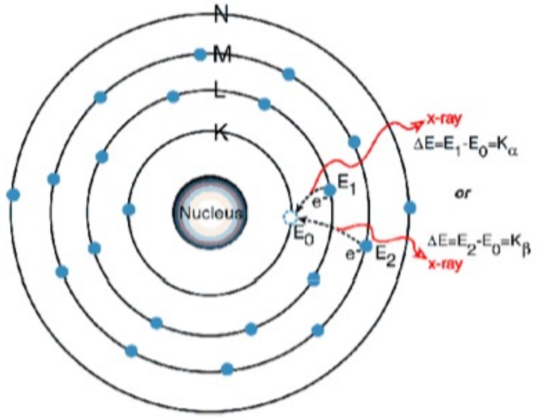
X-Ray Generation

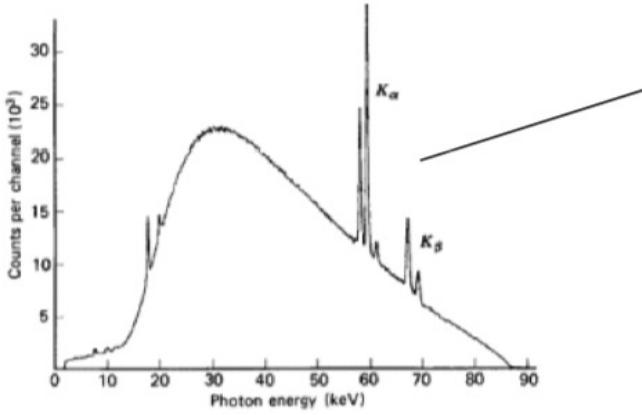
Electrons hitting anode release their energy via *Bremsstrahlung*

- 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

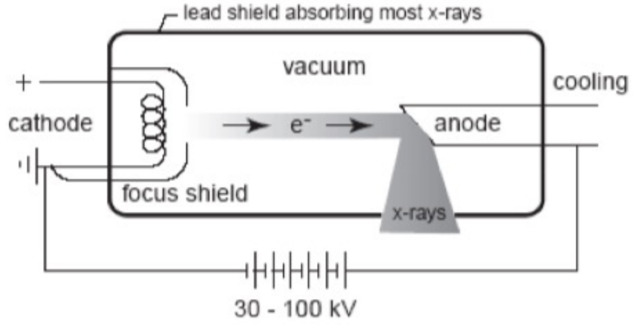
Important X-ray tube parameters:

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





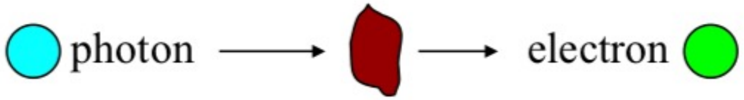
The spectrum from a tungsten target x-ray tube operated at 87 kVp.



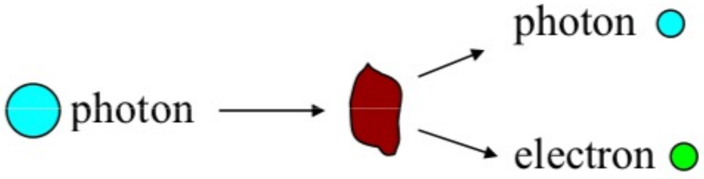
X-Ray Interaction with matter

Three types of interaction with matter:

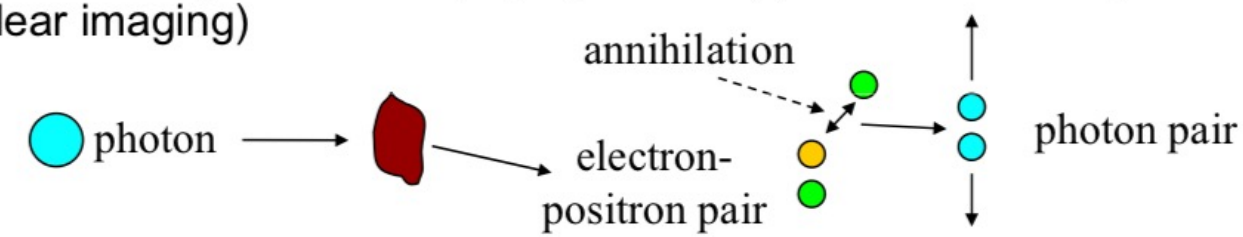
- *photo-electric absorption*: absorption of a photon by an atom and release of an electron along the same direction (*ionizes*)



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



- *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

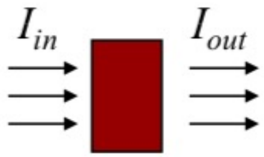
- desirable in function imaging (see later)

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:

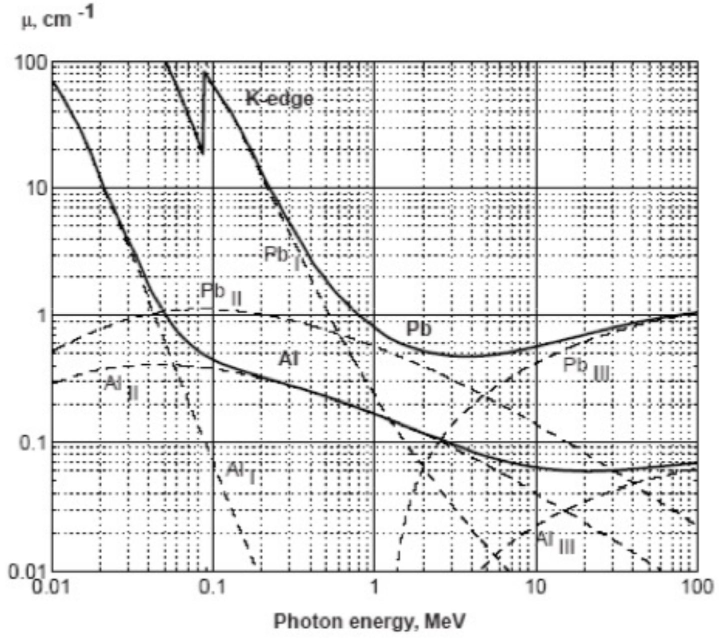
$$I_{in} = \int_0^{\infty} \sigma(E)dE \qquad I_{out} = \int_0^{\infty} \sigma(E)e^{-\int_{x_{in}}^{x_{out}} \mu(E,x)dx} dE$$

- 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

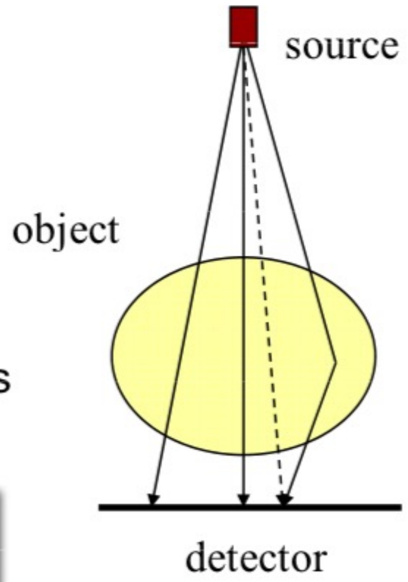
- dominates effects at energies >26keV (at 26keV photo-el = Compton)
- dense materials (such as bone) threshold higher

Scattered photons are detrimental to imaging

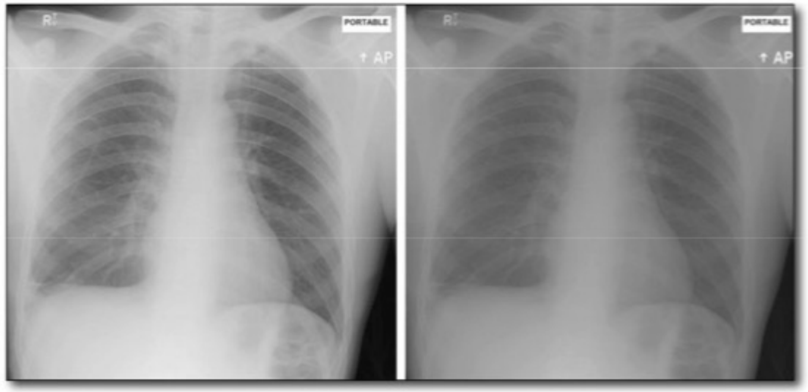
- they violate the straight ray assumption
- tend to under-estimate attenuation

Quantified by SPR: Scatter/Primary Ratio

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



without scatter



with scatter

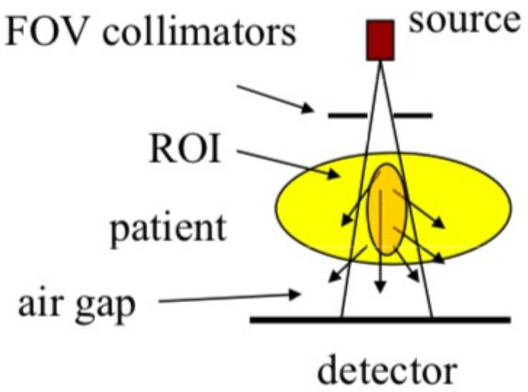
Scattered Radiation

Depends on

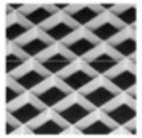
- energy of the x-rays (↑)
- patient thickness (↑) → in abdominal imaging SPR>3

But scattered radiation has less energy than direct radiation

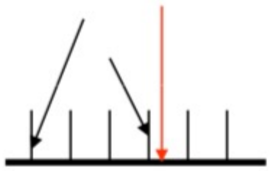
- field of view FOV (↑) → want to reduce FOV to region of interest (ROI) as much as possible
- air gap between patient and screen (↓)
  but, air gap reduces resolution and FOV



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):

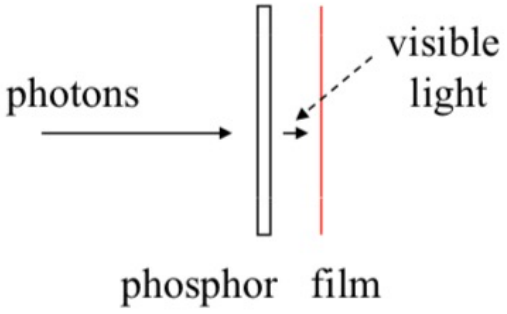
$$QE = \frac{\text{detected photons}}{\text{incoming photons}} 100 \ \text{percent}$$

Photographic film: very inefficient (QE=2%)

- would require huge patient doses

Phosphor-based: Place film between two intensifying fluorescent screens

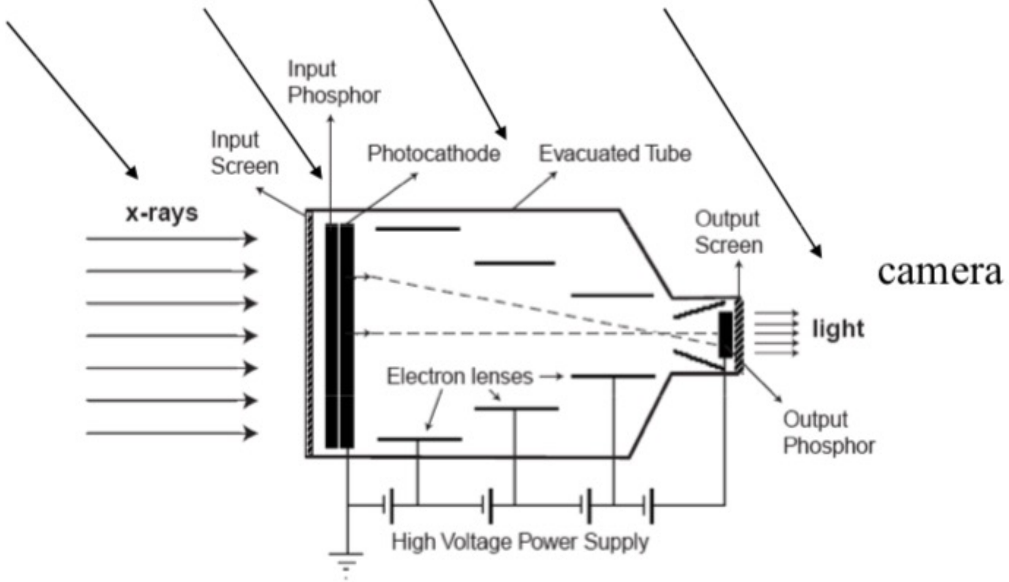
- made out of rare earth phosphors (gadolinium oxysulfide Gd_2O_2S)
- phosphor converts X-rays to scattered visible light
- light directed toward film is recorded (QE=25%)



X-Ray Detectors: Image Intensifier

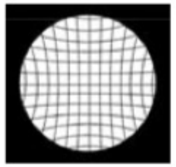
Image intensifiers produces images at high speeds (unlike film)

- 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)

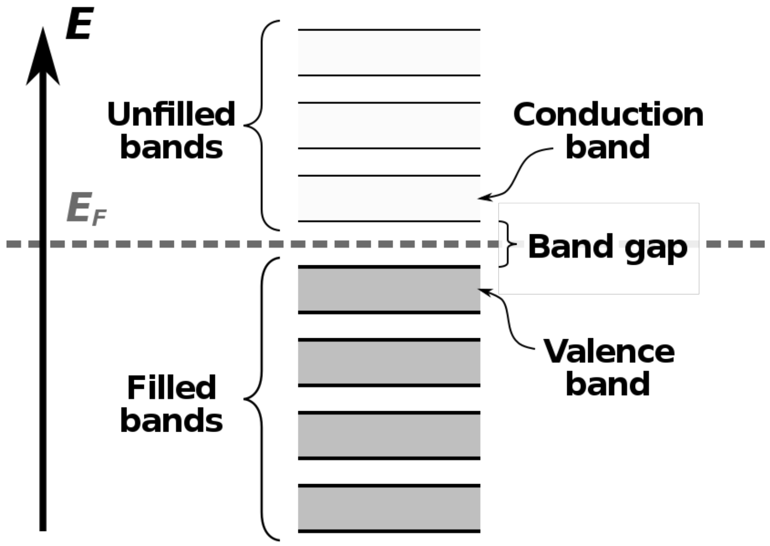


X-Ray Detectors: Storage Phosphors

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

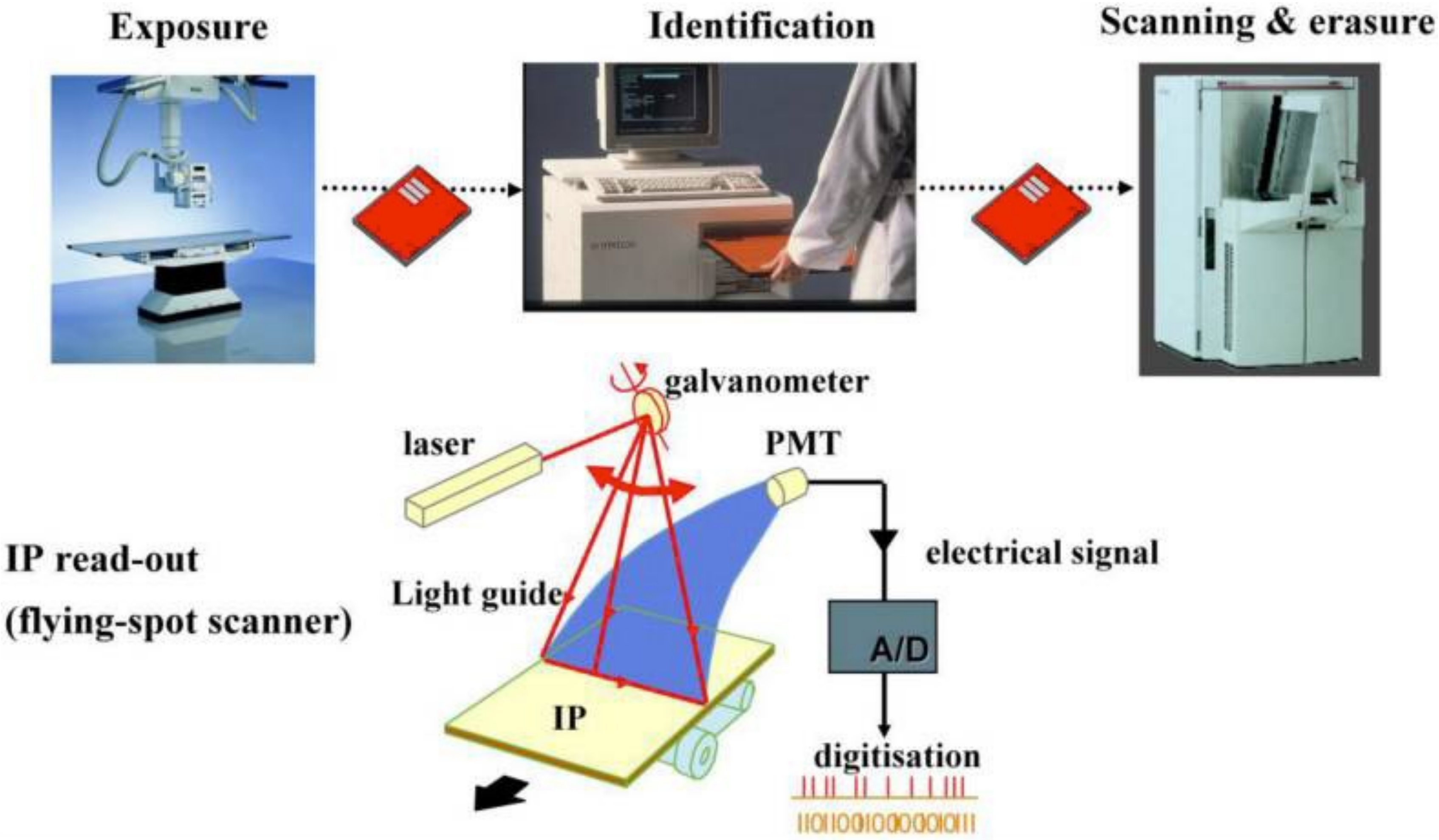


Clear: subject plate to strong light source

Advantages:

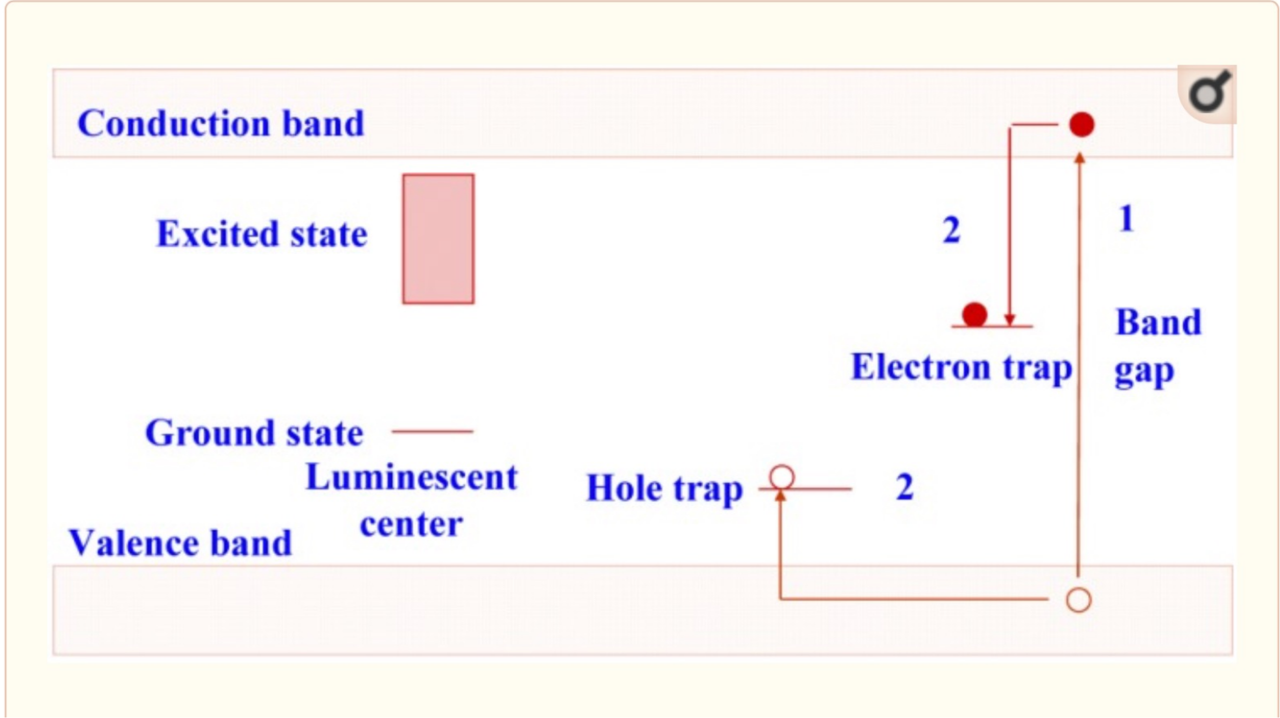
- linear detector response (while film follows an S-curve)
- allows efficient digital mass storage
- allows use in *Picture Archiving and Communication Systems* (PACS)

X-Ray Detectors: Storage Phosphors



X-Ray Detectors: Storage Phosphors

In CR, the conventional X-ray phosphor screen + film detector is replaced by a storage phosphor plate. Evidently, the storage phosphor is responsible for substantial X-ray absorption. Unlike what happens in conventional phosphors, in a storage phosphor, part of the electron/hole pairs do not recombine to transfer their energy to a luminescent center. A considerable fraction is trapped in metastable states. A simplified non-detailed general model of the storage process is shown in Figure 2. As long as the phosphor is not exposed to light or heat, recombination is not possible and the electrons and holes remain trapped. The spatial distribution of the trapped charges in a plate containing storage phosphor crystals makes up the latent image in CR.



Paul Leblans, et. al, "Storage Phosphors for Medical Imaging," Materials, 2011.

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: high-energy x-rays→ photons*) technology: hydrogenated amorphous silicon detectors (a-SI:H)

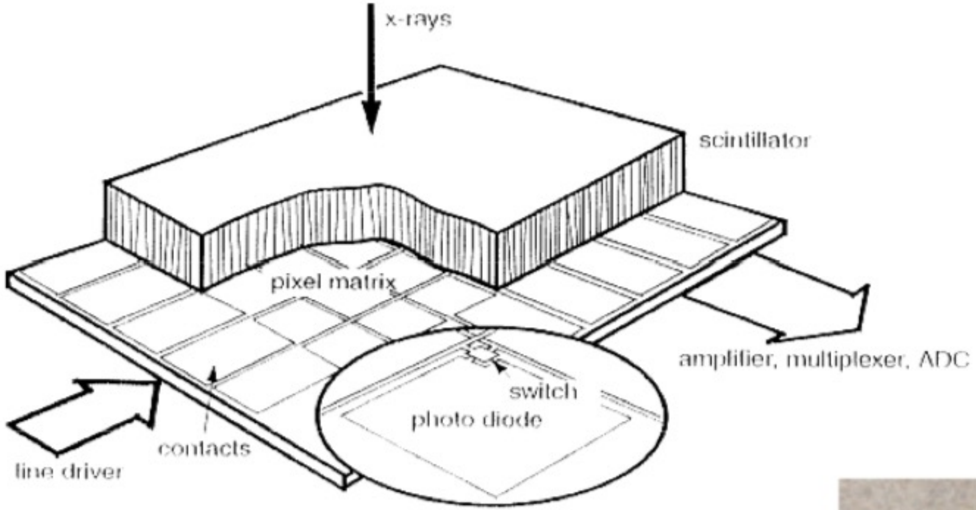
- can be manufactured in flat, large sheets
- can be coupled directly with the phosphor plate
- but still need to convert photons to visible light, affecting resolution

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)

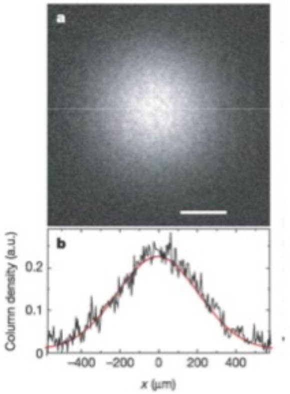
Flat Panel Amorphous Silicon Detector

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,

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

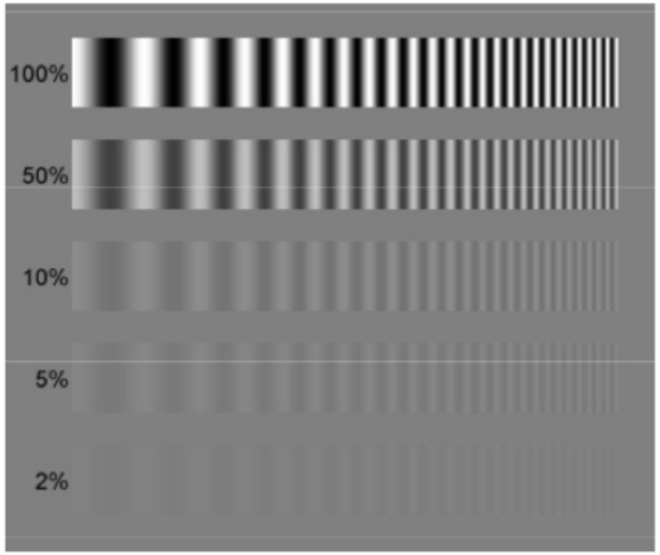
SNR improves as the number of photons N increases

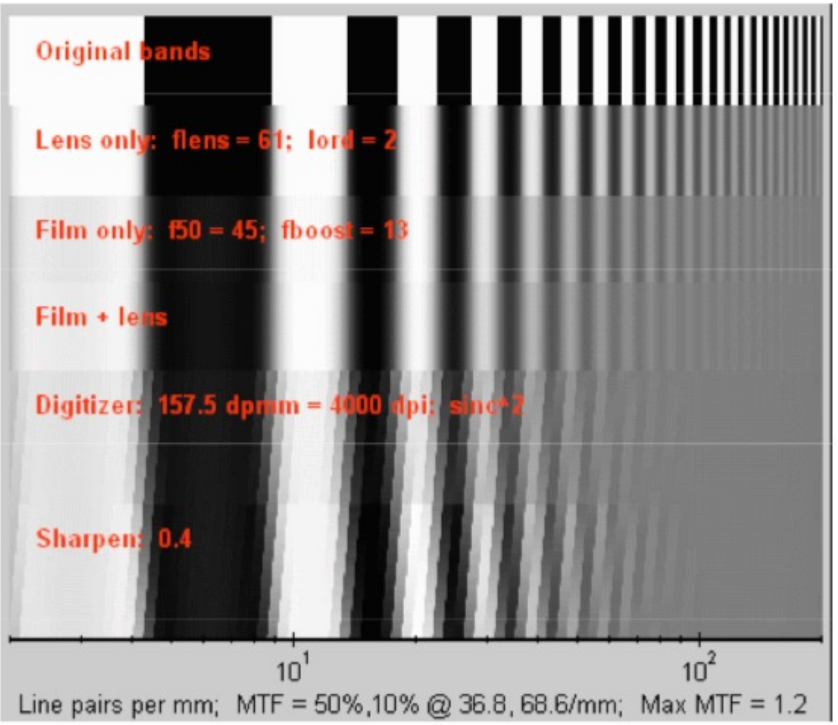
- however, this also increases patient dose
- so there is a trade-off
- doubling SNR increases dose by a factor of 4

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 %





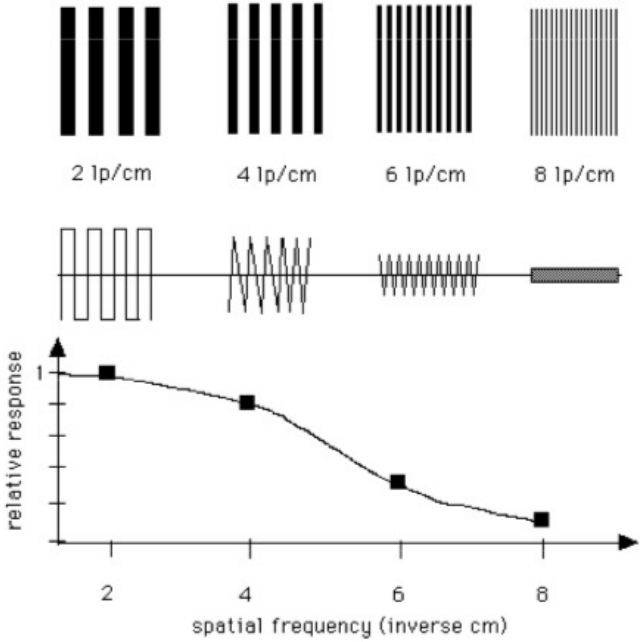
DQE: Detective Quantum Efficiency

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:

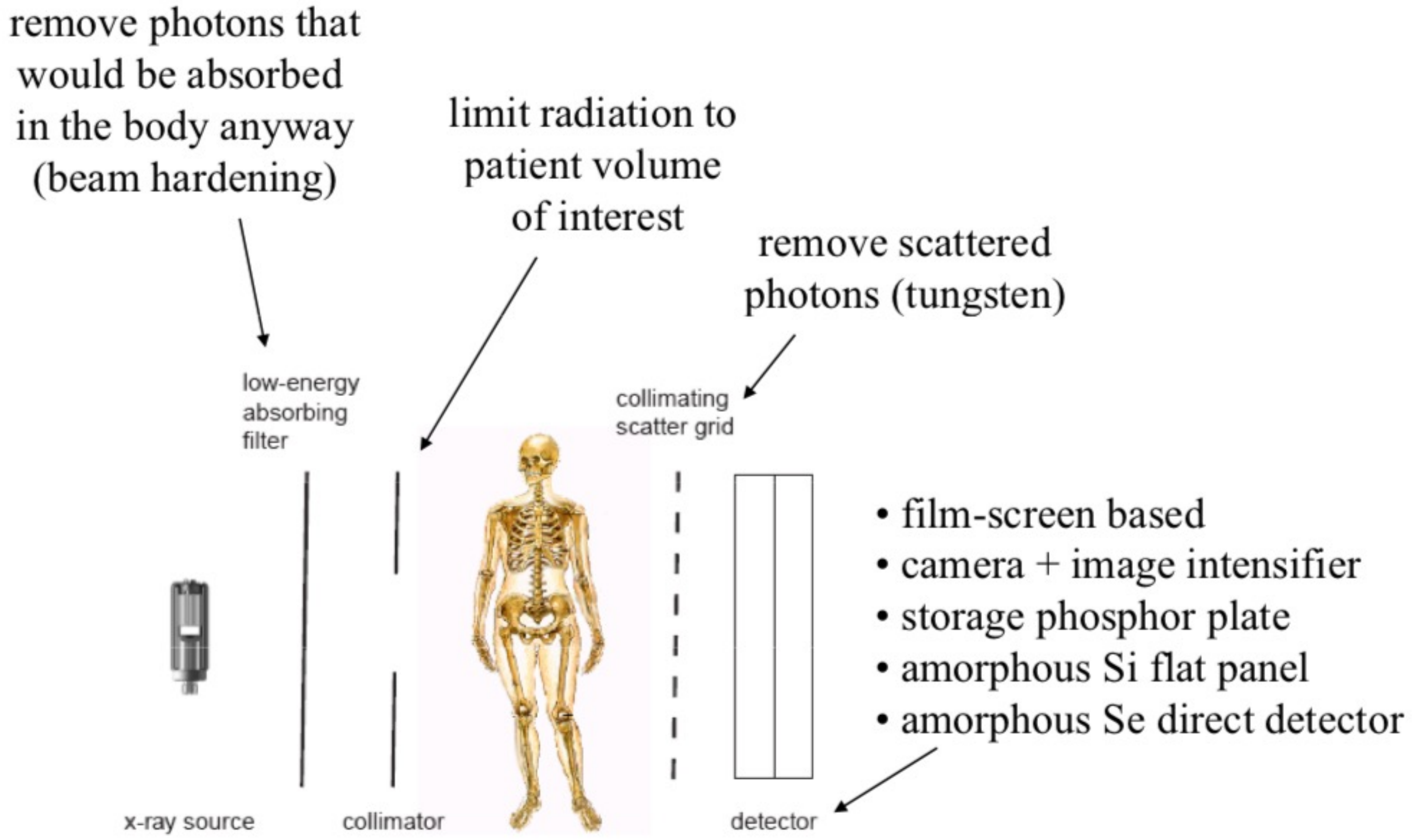
- quality of the anode tip (finer tips give better focus)
- patient thickness (thicker patients cause more scattering, which deteriorates resolution)
- light scattering properties of the phosphor (for phosphor-based systems)
- film resolution (for film-based systems)
- sampling procedure (for systems 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 ≥ 2000^2 pixels

Available Technology: Summary



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)

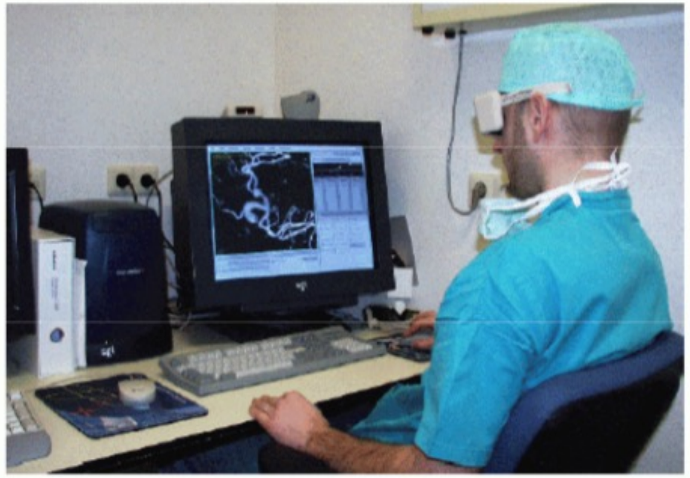
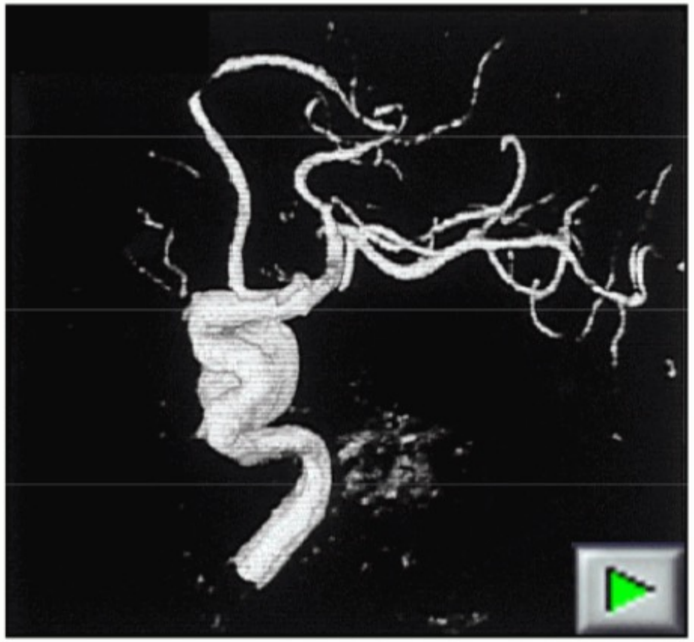
Case Studies (1)

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.

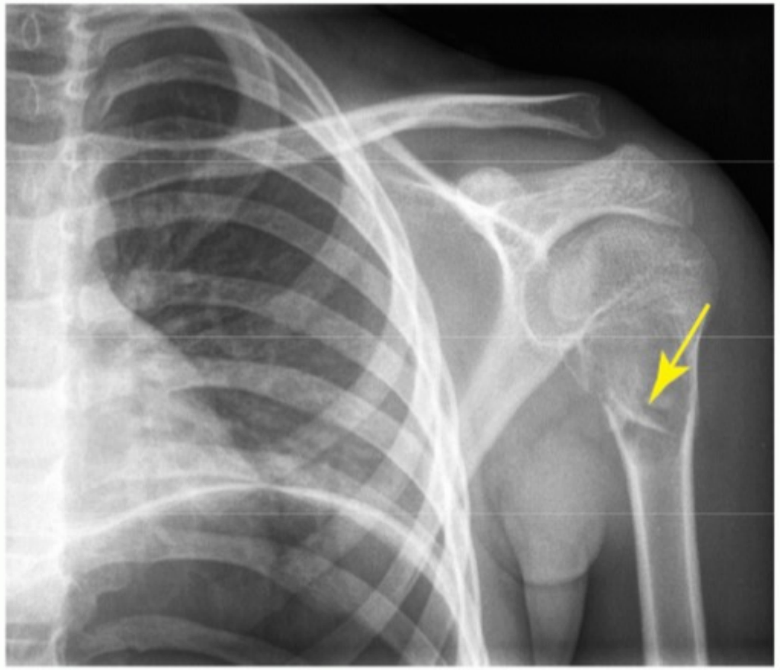
Case Studies (3)

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

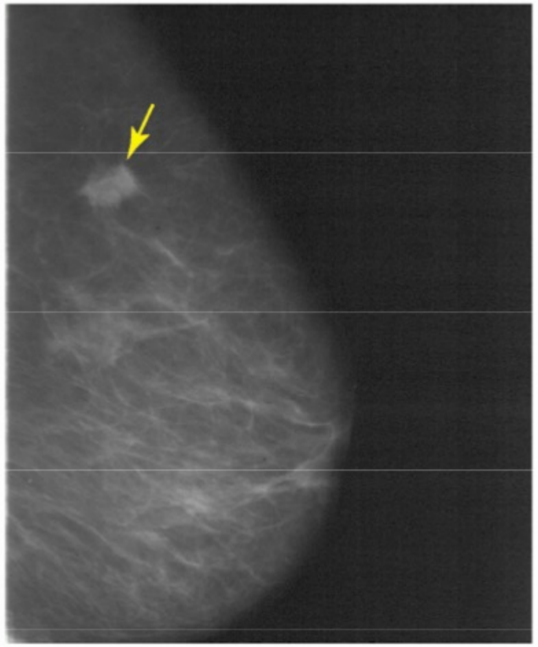
Case Studies (4)

Double mandibular fracture with strong displacement to the left.



Solitary humeral bone cyst known as ”fallen leaf sign”

Case Studies (5)

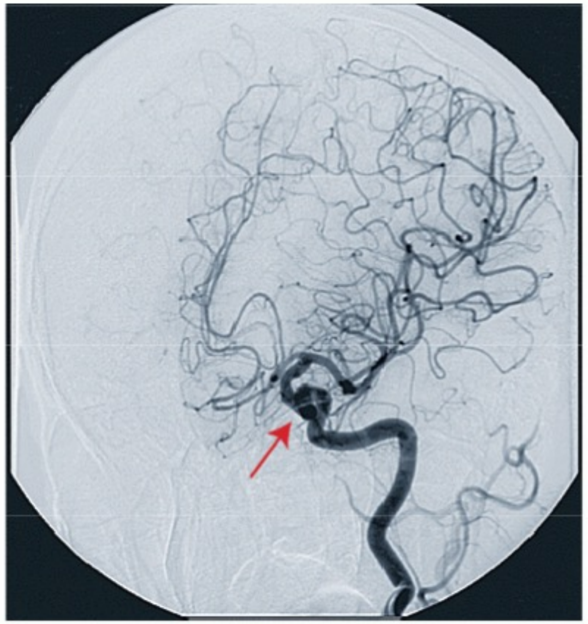


Dense opacity with spicular borders in the left breast, which suggests a malignant lesion

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

Case Studies (6)

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 (7)

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).

Case Studies (8)

Upper GI Series:

Typical Application of fluoroscopy: live (and continuous) X-ray imaging to monitor dynamic phenomena (also for instrument tracking in surgeries)