# Introduction to Medical Imaging

## Lecture 7: Radiography

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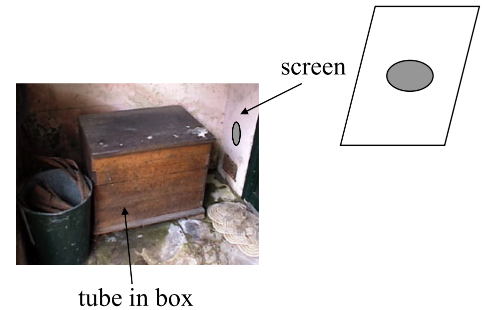
## X-Ray Discovery

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

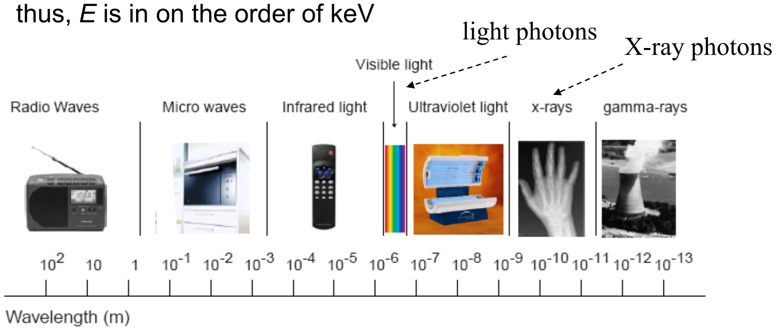




## X-Ray Physics

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)
$\lambda$: wavelength (on the order of $10^{-10}$ m)
thus, $E$ is in on the order of keV



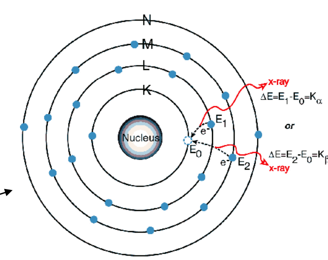

## X-Ray Generation

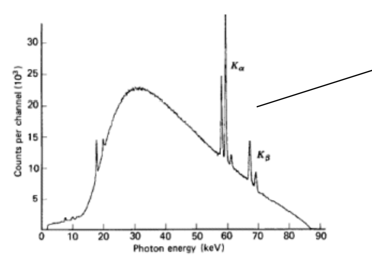
Electrons hitting anode release their energy via *Bremsstrahlung*

- 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

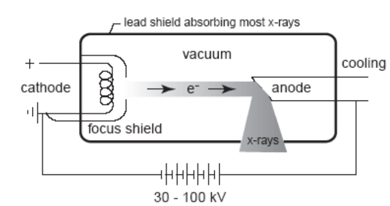
Important X-ray tube parameters:

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



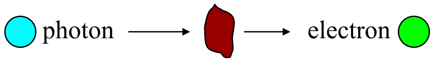

The spectrum from a tungsten target x-ray tube operated at 87 kVp.



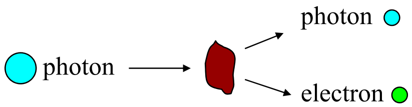
## X Ray Interaction With Matter

Three types of interaction with matter:

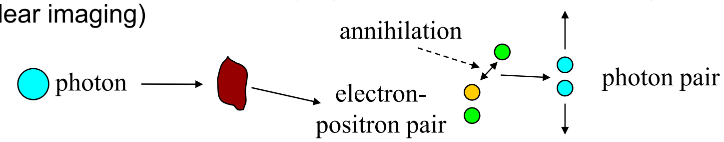
- *photo-electric absorption*: absorption of a photon by an atom and release of an electron along the same direction (*ionizes*)




- *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

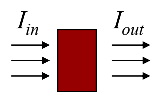
- desirable in function imaging (see later)

## 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:

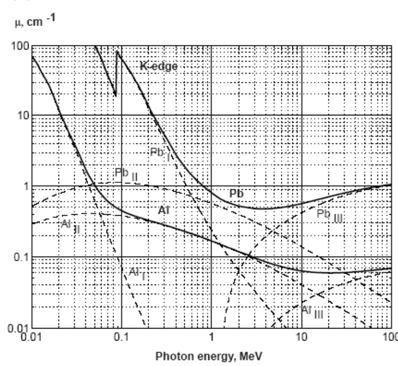
$$I_{in} = \int_0^\infty \sigma(E)dE \qquad I_{out} = \int_0^\infty \sigma(E) e^{-\int_{x_{in}}^{x_{out}} \mu(E,x)dx} dE$$

- 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

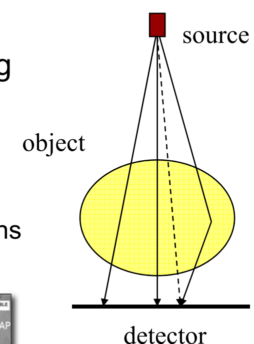
- dominates effects at energies >26keV (at 26keV photo-el = Compton)
- dense materials (such as bone) threshold higher

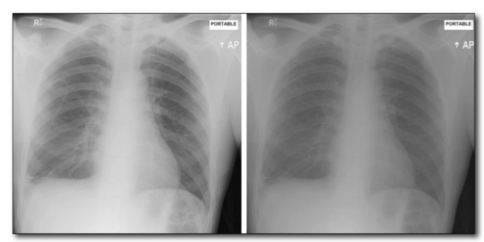
Scattered photons are detrimental to imaging

- 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

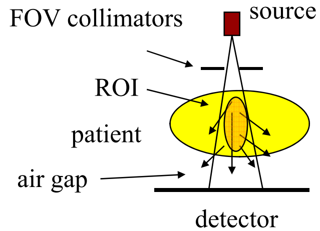
with scatter

## Scattered Radiation

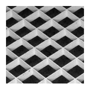
Depends on

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

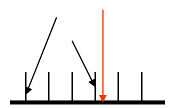
But scattered radiation has less energy than direct radiation




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):

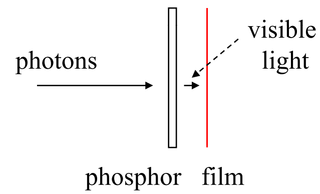
$$QE = \frac{\text{detected photons}}{\text{incoming photons}} 100 \quad \text{percent}$$

Photographic film: very inefficient (QE=2%)

- would require huge patient doses

Phosphor-based: Place film between two intensifying fluorescent screens

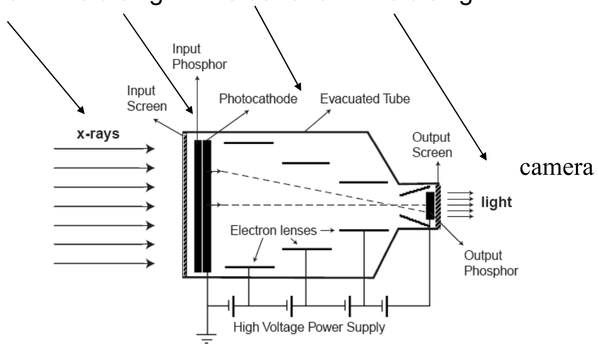
- made out of rare earth phosphors (gadolinium oxysulfide $Gd_2O_2S$)
- phosphor converts X-rays to scattered visible light
- light directed toward film is recorded (QE=25%)



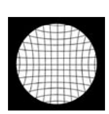

## X-Ray Detectors: Image Intensifier

Image 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: 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)

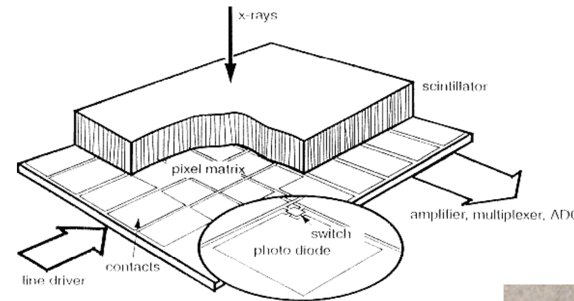
- can be manufactured in flat, large sheets
- can be coupled directly with the phosphor plate
- but still need to convert photons to visible light, affecting resolution

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)

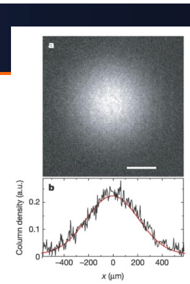
## Flat Panel Amorphous Silicon Detectors

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,

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

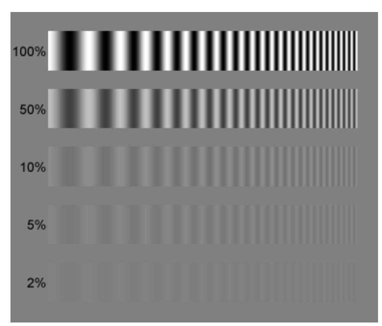
SNR improves as the number of photons N increases

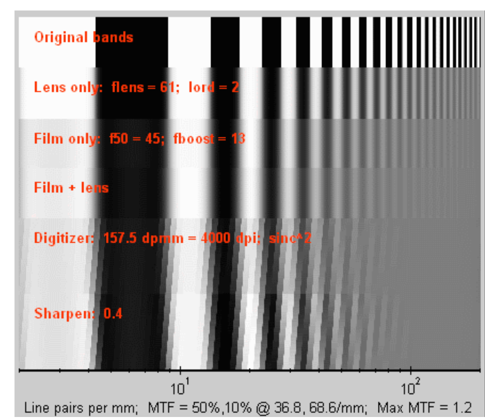
- however, this also increases patient dose
- so there is a trade-off
- doubling SNR increases dose by a factor of 4

## 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 %



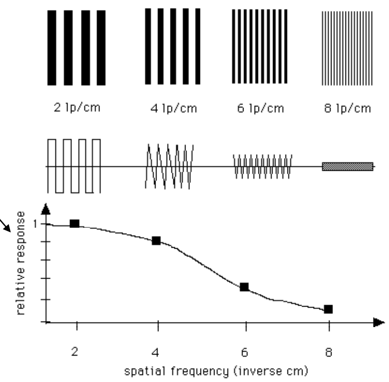
## DQE: Detective Quantum Efficiency

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:

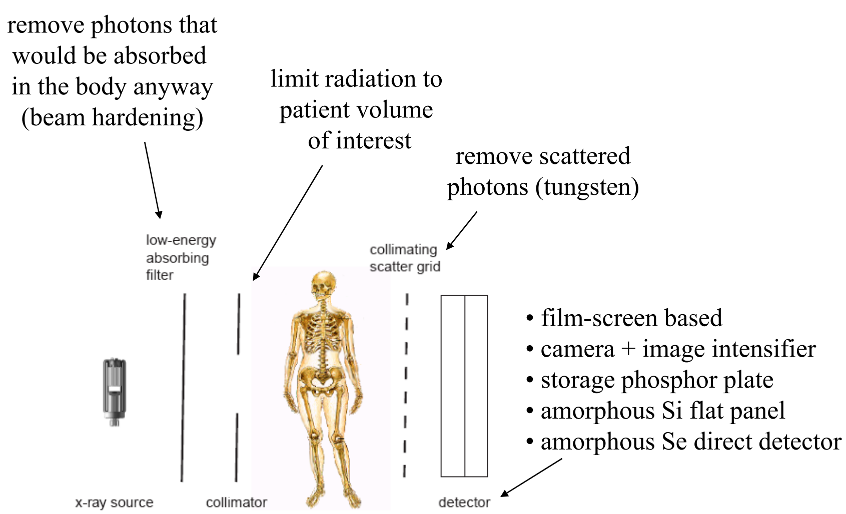
- quality of the anode tip (finer tips give better focus)
- patient thickness (thicker patients cause more scattering, which deteriorates resolution)
- light scattering properties of the phosphor (for phosphor-based systems)
- film resolution (for film-based systems)
- sampling procedure (for systems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

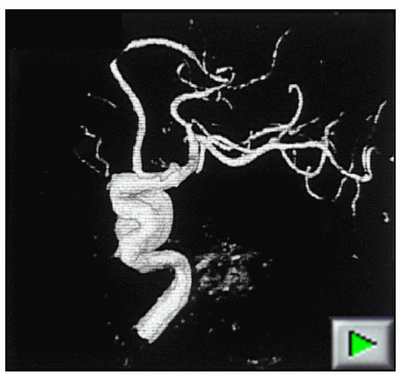
## Case Studies (1)

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.

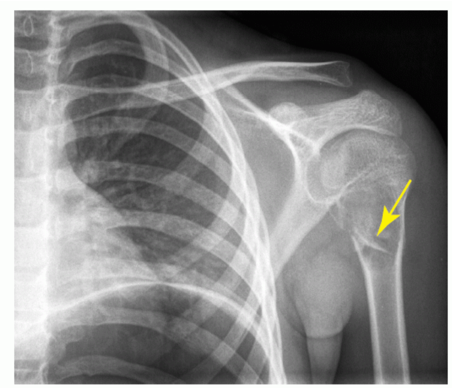
## Case Studies (3)



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

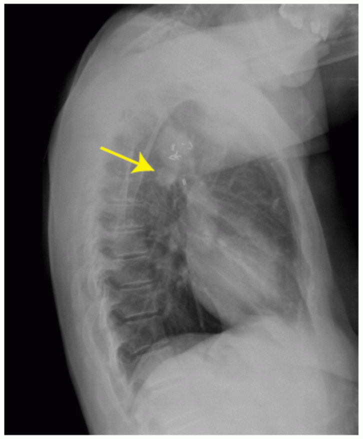
## Case Studies (4)

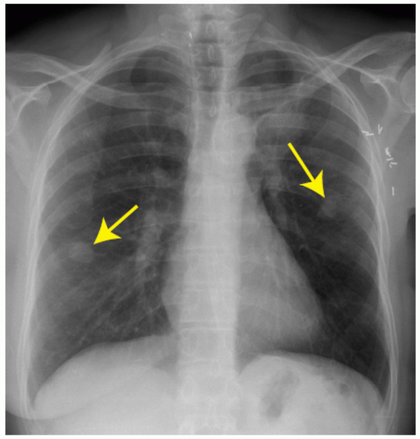
Double mandibular fracture with strong displacement to the left.



Solitary humeral bone cyst known as ”fallen leaf sign”

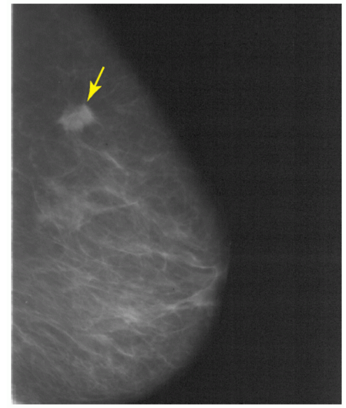
## Case Studies (5)



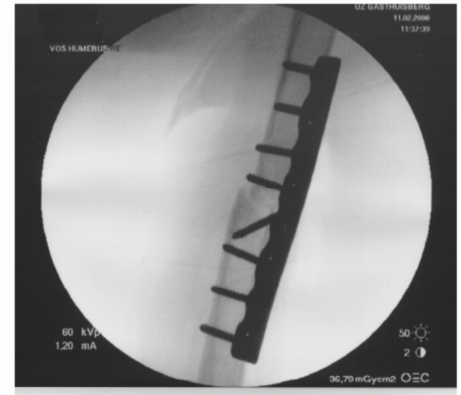


Radiographic chest image showing multiple lung metastases

## Case Studies (6)



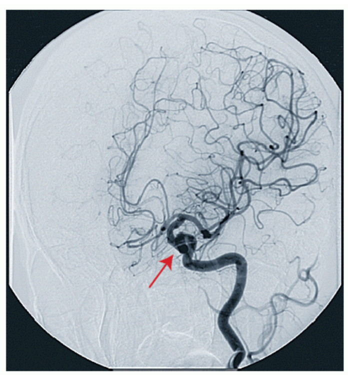
Dense opacity with spicular borders in the left breast, which suggests a malignant lesion



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

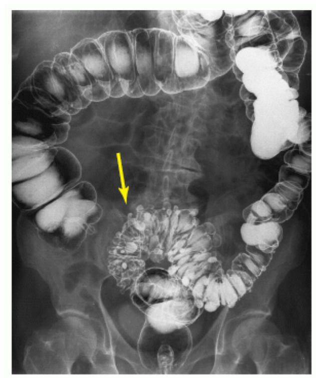
## Case Studies (7)

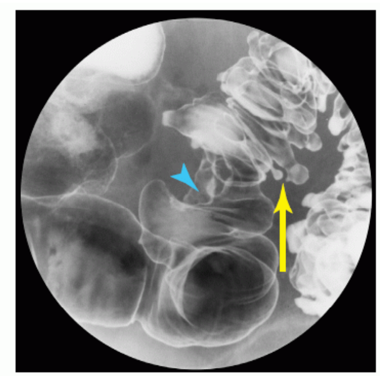
Cerebral angiogram obtained by injecting a iodine containing fluid into the arteries. The contrast dye subsequently fills the cerebral arteries, capillaries and veins.



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).

## Case Studies (9)

Upper GI Series:

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

## 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 ($\alpha$-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 doses

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 while 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 fatal cancer resulting from radiation:

- 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