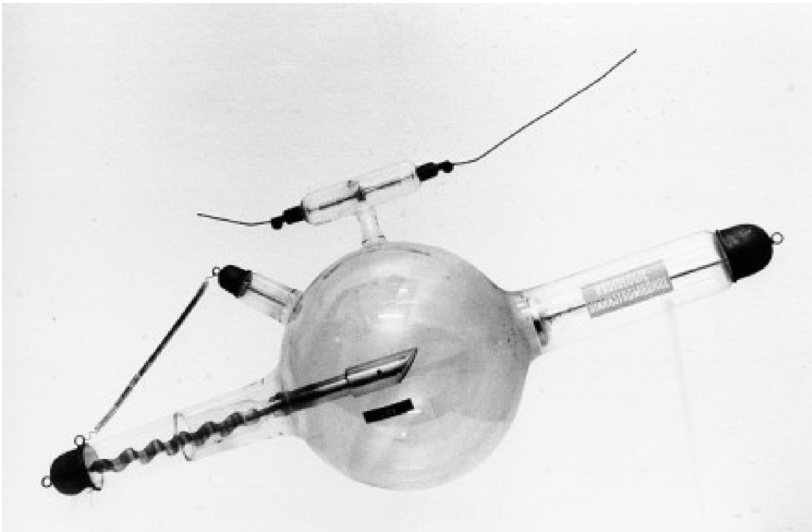
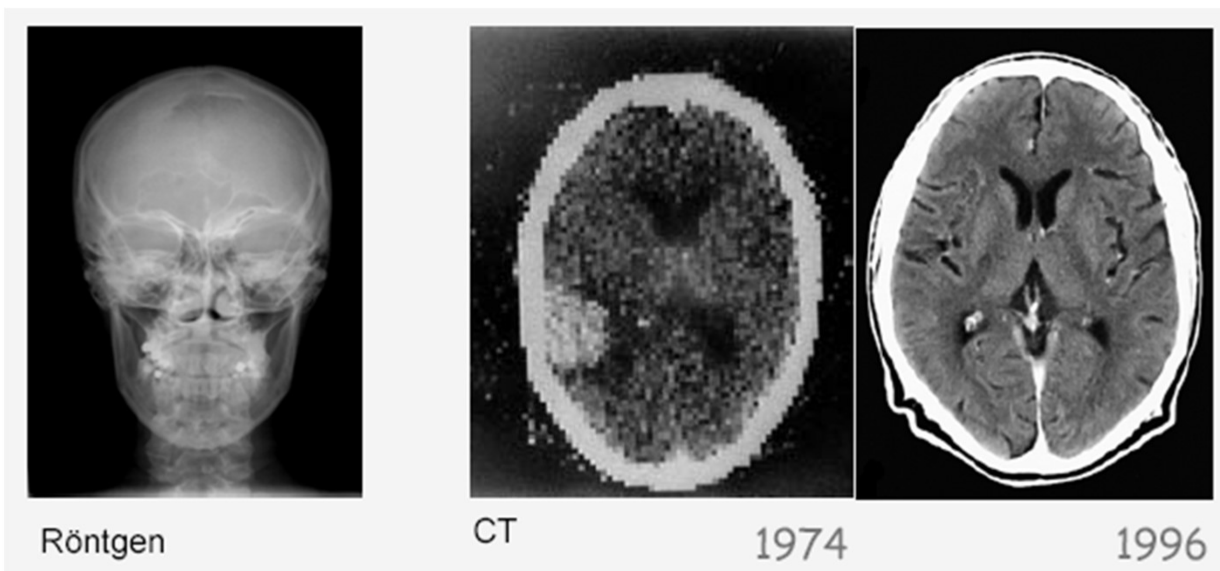
Imaging with x-rays





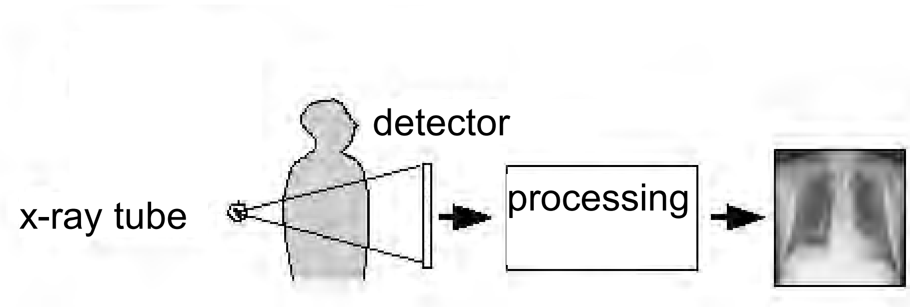
Contents:

- historical overview
- physics of x-rays
- generating x-rays
- interactions with matter
- detectors
- imaging with x-rays
    - projection radiography
    - computed tomography

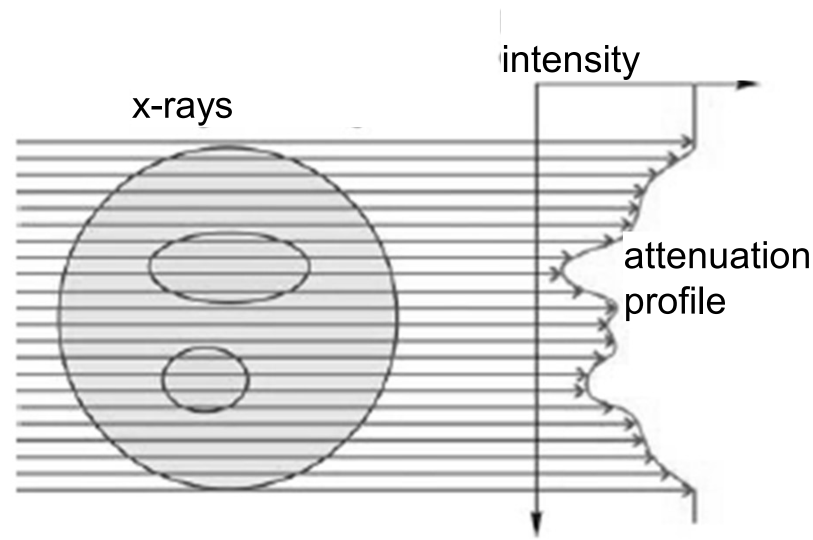
(pictures credits: Dössel, 2000; Morneburg, 1995; Kalender, 2000; Siemens, Philips, Internet)

principle

- ***active*** imaging through exposure of energy



- **attenuation** of x-rays by different tissues



Imaging with x-rays

history

1895 Wilhelm Conrad Röntgen (27.3.1845 - 10.2.1923)

discovery of x-rays on 8. November 1895
imaging of Mrs. Röntgen’s hand on 22. December 1895

1901 Nobel Physics prize awarded to Röntgen

1912 verification as e.m. wave with scattering experiments in crystals (Friedrich, Knipping, von Laue)

1917 Johann Radon: Radon-transform as mathematical principle for tomography
(*Über die Bestimmung von Funktionen durch ihre Integralwerte längs gewisser Mannigfaltigkeiten*. Ber. vor Sächs. Akad. Wiss., 69, 262)

Coolidge: x-ray tube with high vacuum

Imaging with x-rays

„Und läßt man der Phantasie weiter die Zügel schießen, stellt man sich vor, dass es gelingen würde, die neue Methode des photographischen Prozesses mit Hilfe von Strahlen aus den Crookeschen Röhren so zu vervollkommnen, dass nur eine Partie der Weichteile des menschlichen Körpers durchsichtig bleibt, eine tiefer liegende Schicht aber auf der Platte fixiert werden kann, so wäre ein unschätzbarer Behelf für die Diagnose zahlloser anderer Krankheitsgruppen als die Knochen gewonnen.“

Anonym, Frankfurter Zeitung, 7. Januar 1896

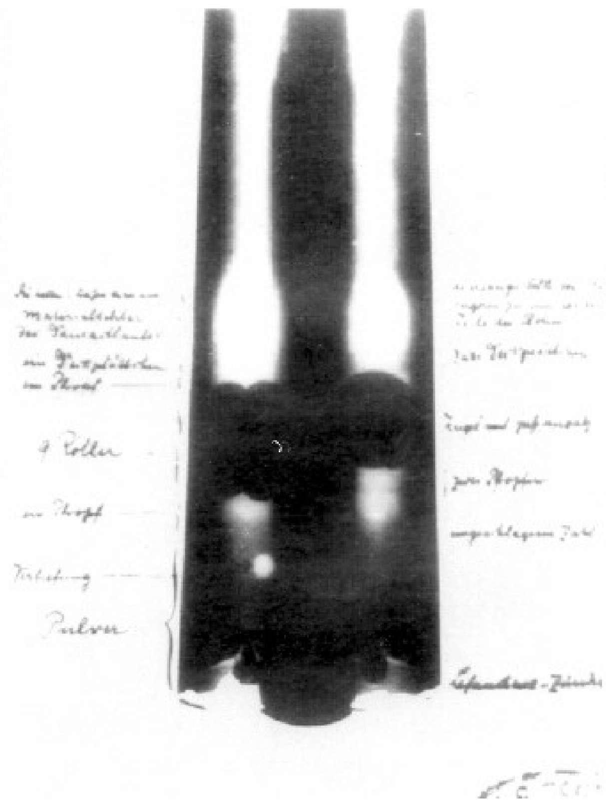
(cited in: W. Kalender Computertomographie, Publicis MCD Verlag, 2000)

Imaging with x-rays

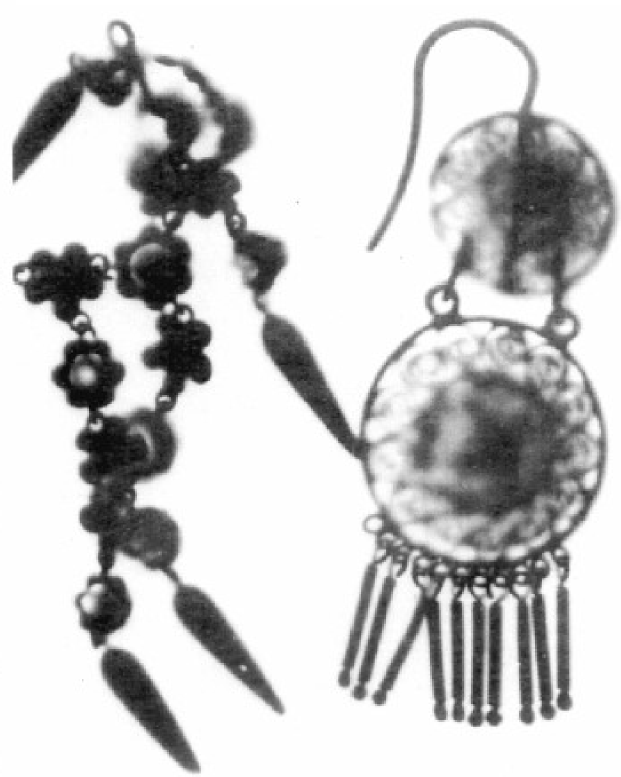
early applications of x-rays (as of 1896)

x-raying of non-transparent objects

hand with ring

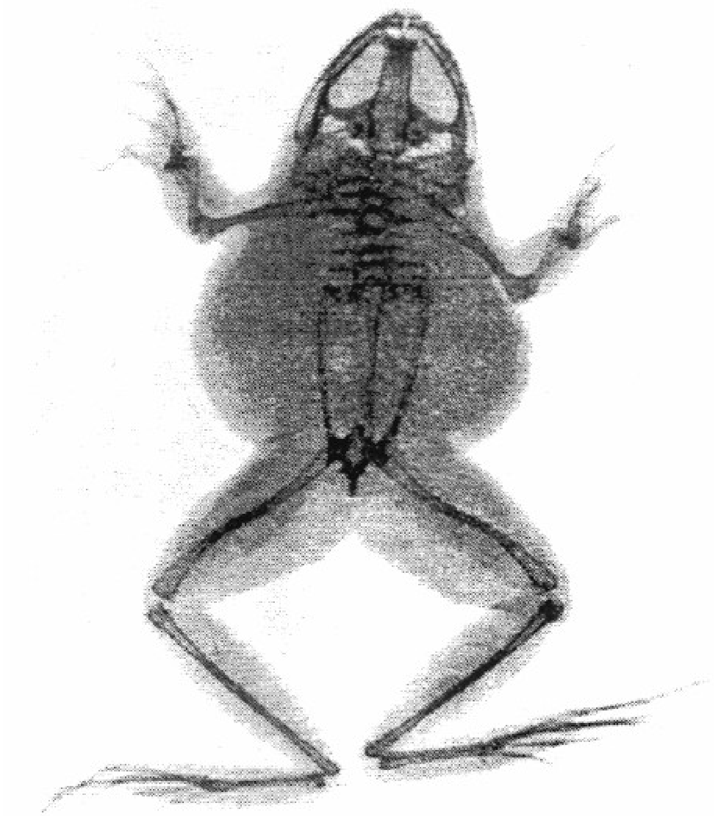


shotgun with bullets

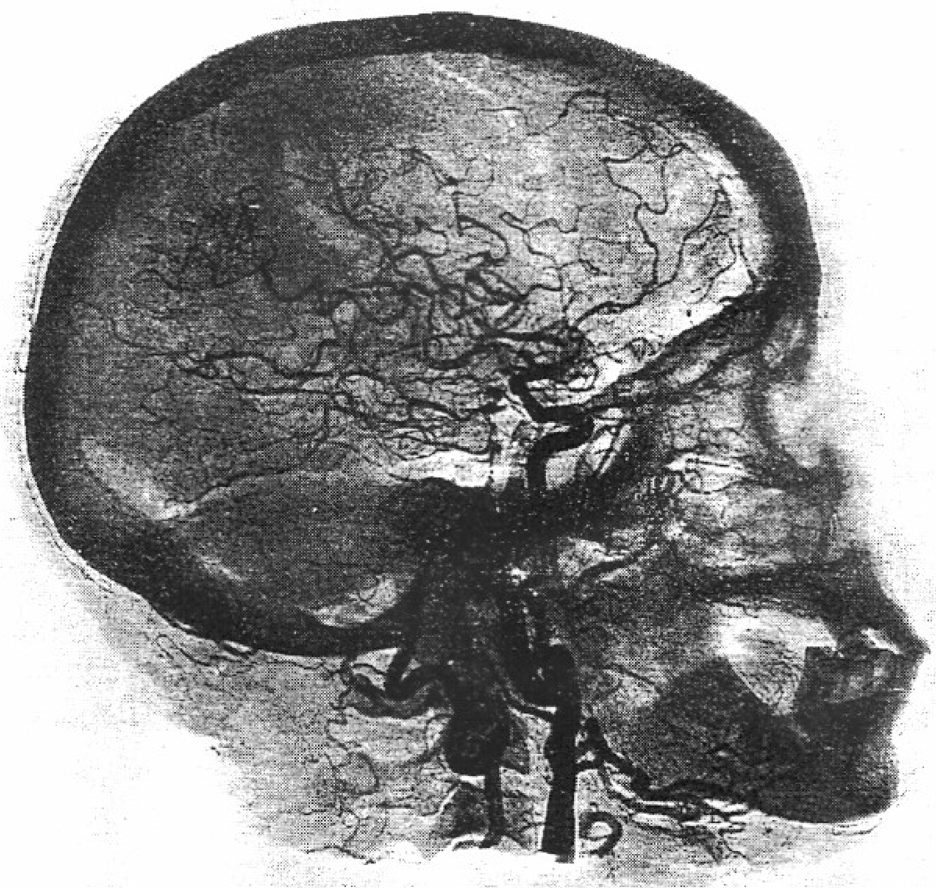


authenticity of jewelry

early applications of x-rays (as of 1896)



radiogram of a frog
(San Francisco 1896)

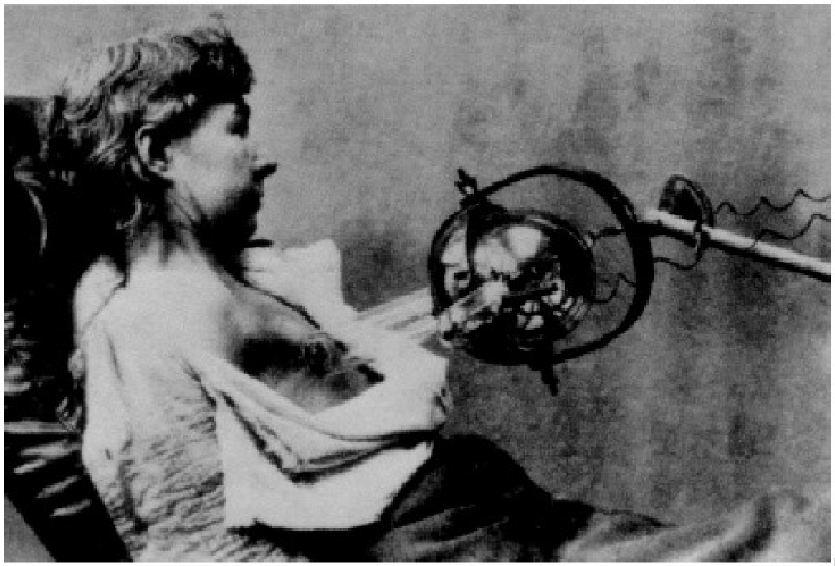


arteriogram (1904)

Mauld, R.F., The Early History of X-Rays in Medicine, Fig. 9 and Fig. 17 aus Michette, A.; Pfauntsch, S., X-Rays: The First Hundred Years; (Wiley 1996)

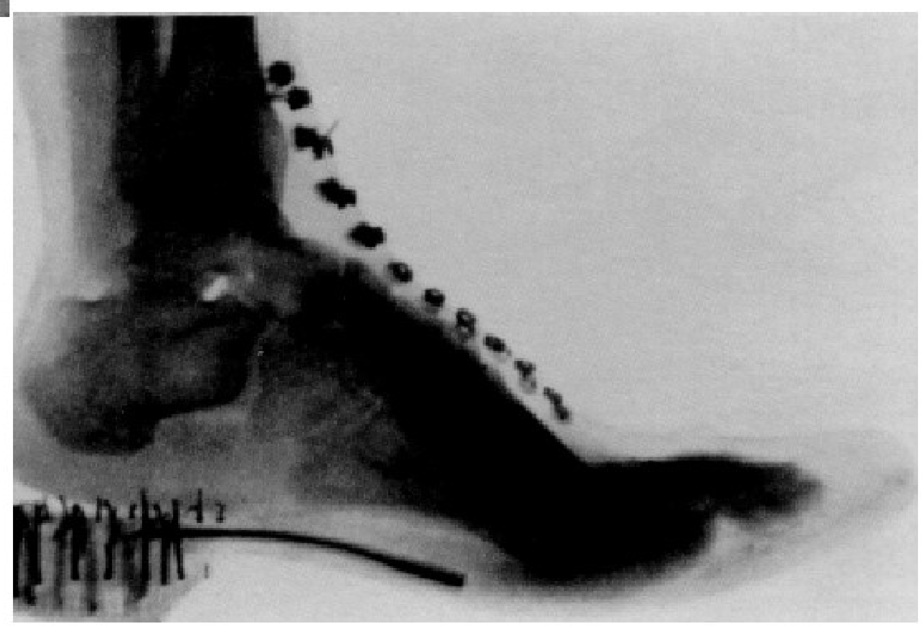
Imaging with x-rays

early applications of x-rays (as of 1896)

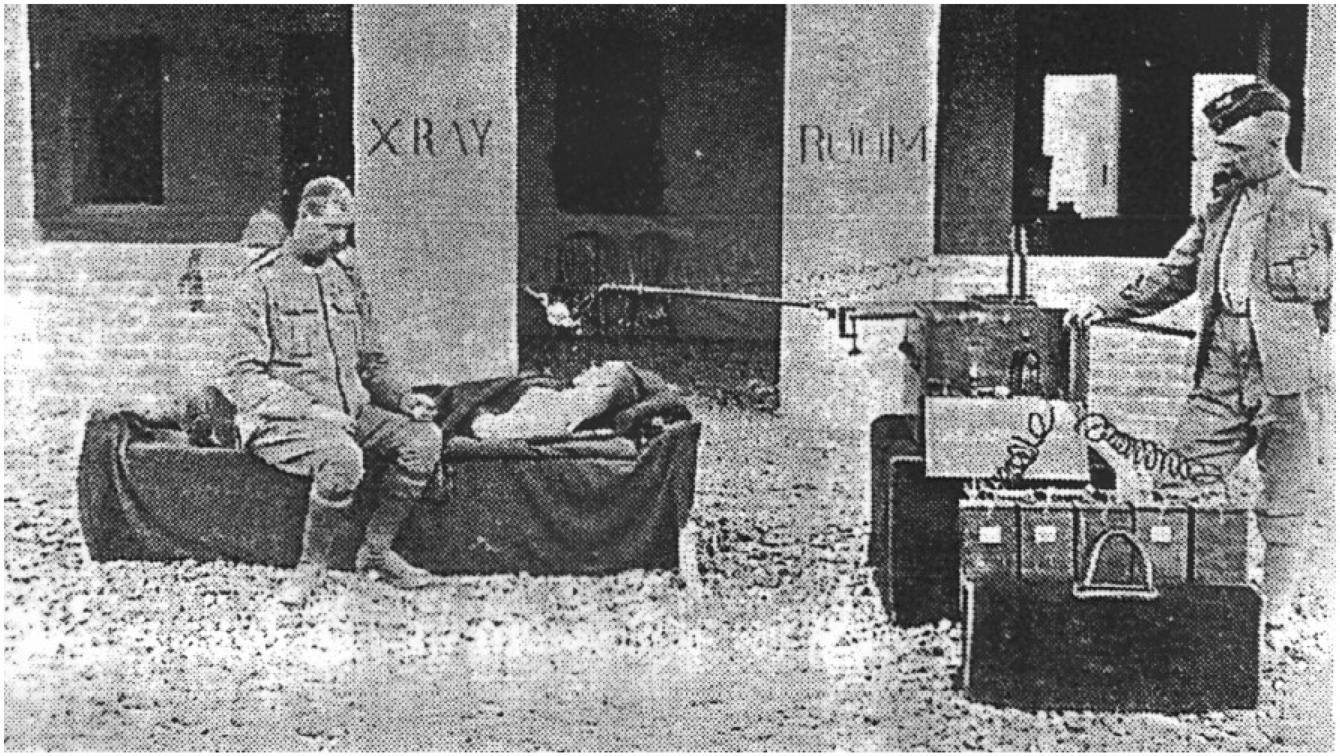


treatment of breast cancer (1905)

adjustment of shoes controlled with x-rays



early applications of x-rays

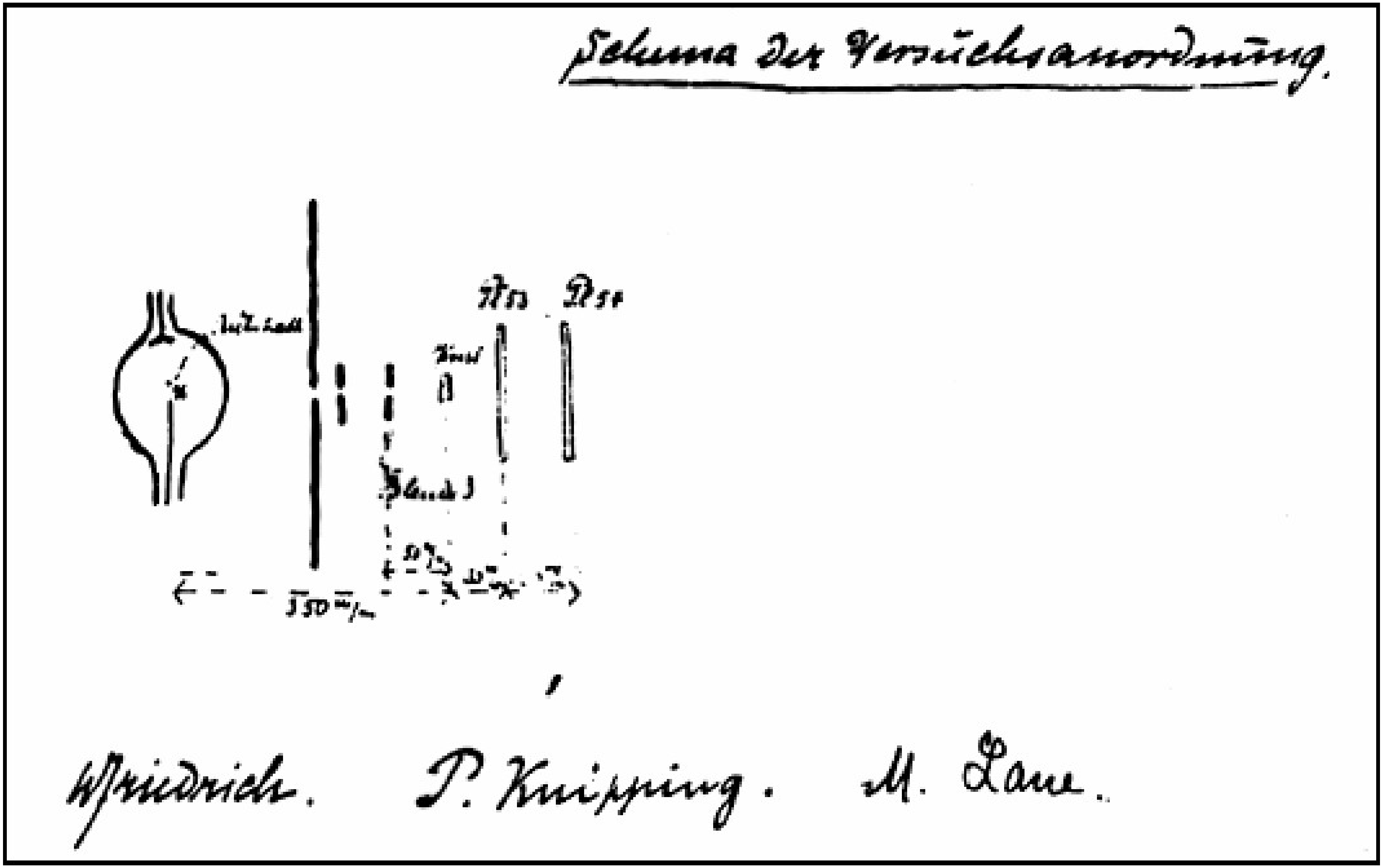


Radiography in the Sudan campaign in the desert some 1200 miles from Kairo (1898). The X-ray tube was described as being suspended by means of an ingenious holder. The use of the inverse square law is to the advantage of the operator on the right but not of the soldier near the patient. It was this installation that the casualties from the battle of Omdurman were radiographed.

Mauld, R.F., The Early History of X-Rays in Medicine, Fig. 11 aus Michette, A.; Pfauntsch, S., X-Rays: The First Hundred Years; (Wiley 1996)

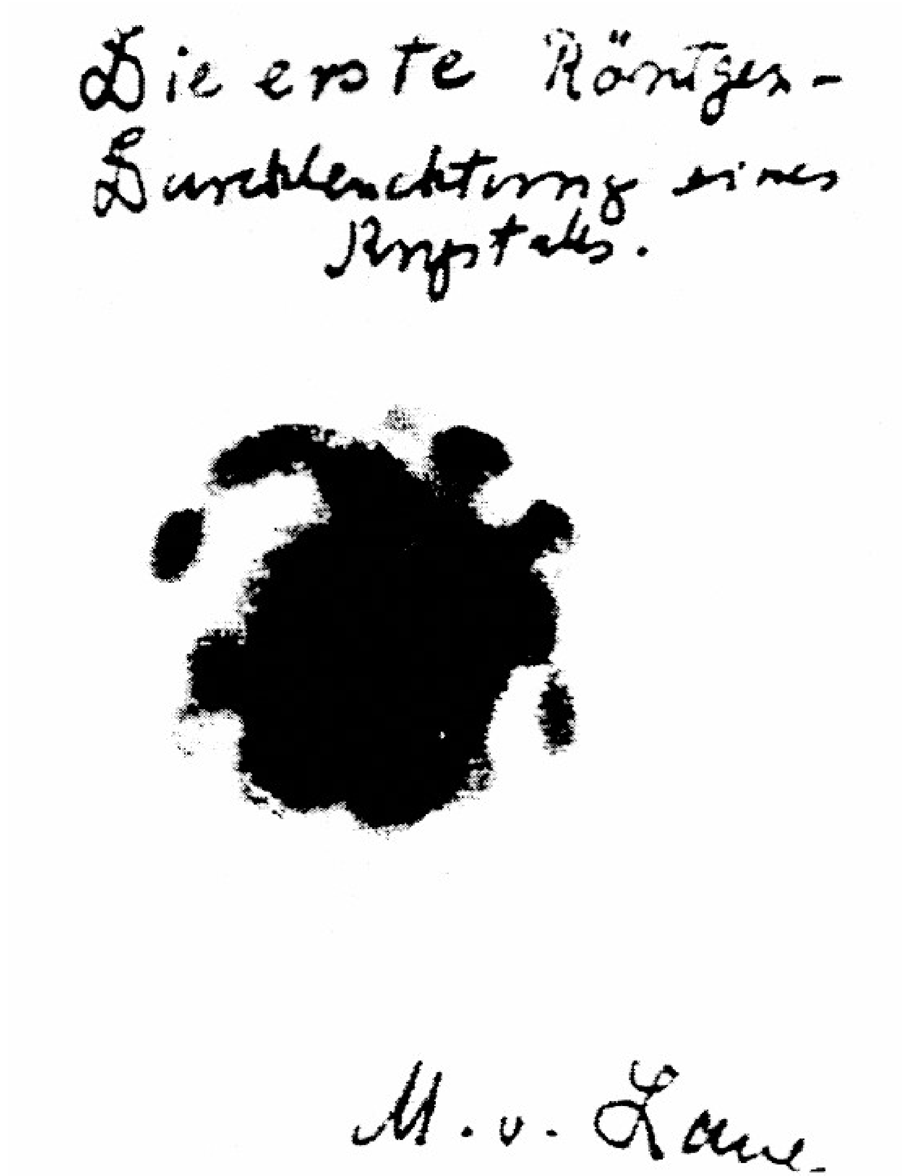
Imaging with x-rays

historical experiment by M. von Laue



Imaging with x-rays

historical experiment by M. von Laue



Imaging with x-rays

1938 Gabriel Frank: a method to generate cross-sectional images of the body with x-rays
(German patent specification 1940)

1957/58 S.I. Tetel‘Baum, B.I. Korenblyum
construction of one of the first CT-scanner ar Politechnical Institute, Kiev, Russia

1961 William D. Oldendorf
first x-ray CT images of a head phantom
(idea: stationary detector; rotating probe)

1963/64 A.M. Cormack
first description of an x-ray-based tomographic method (~CT)
(*Representation of a function by its line integrals, with some radiological applications*. J Appl Phys, 34, 2722, 1963)

Imaging with x-rays

1967 Godfrey N. Hounsfield (engineer)

EMI Lab., England; beginning of computed tomography

M.M. Ter-Pergossian: physical aspects of diagnostic radiology

1971/72 G.N. Hounsfield, J. Ambrose, J. Perry

first CT for clinical applications

1973 transversal/axial CT

1979 A.M. Cormack u. G.N. Hounsfield:

Nobel Physiology and Medicine prize

1980 digital radiography

2000 ca. 30.000 CT-installations worldwide

basics of electromagnetic waves – x-rays:

light:

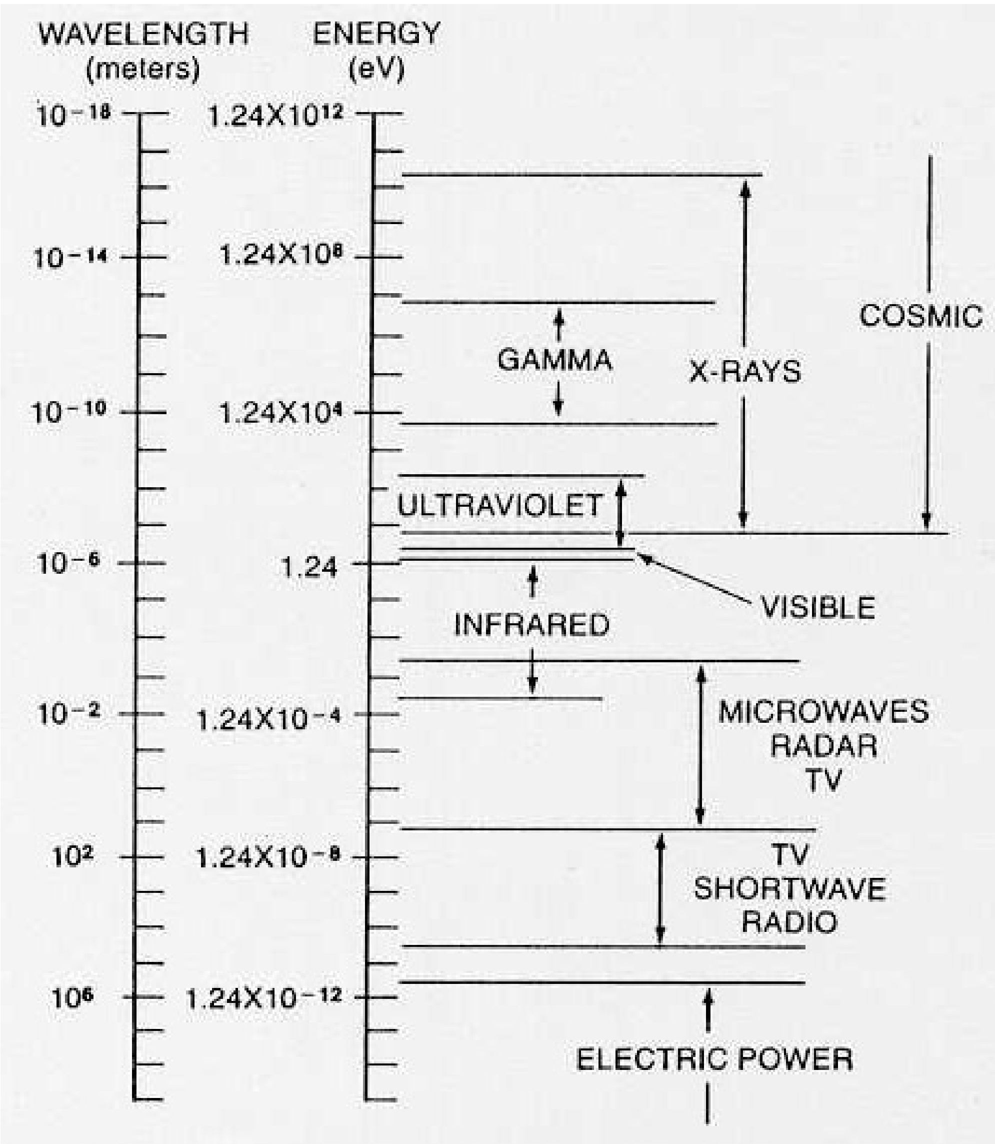
particle properties
(photons, light quanta)

$E = h f$ [E] = 1eV
$h = 4.136 \cdot 10^{-21}$ MeV s
$c = 2.997 \cdot 10^{-8}$ m s^{-1}

wave properties
e.m. wave with (mean) frequency f resp. wavelength λ

$$c = \lambda f$$

basics of electromagnetic waves – x-rays:



range of wavelengths for x-rays

$\lambda = 3 \cdot 10^{-8}$ m - $3 \cdot 10^{-14}$ m

$\Leftrightarrow$

$f = 10^{16}$ Hz - 10^{22} Hz

$= 10^{5}$ GHz - 10^{13} GHz

range of photon energies for x-rays:

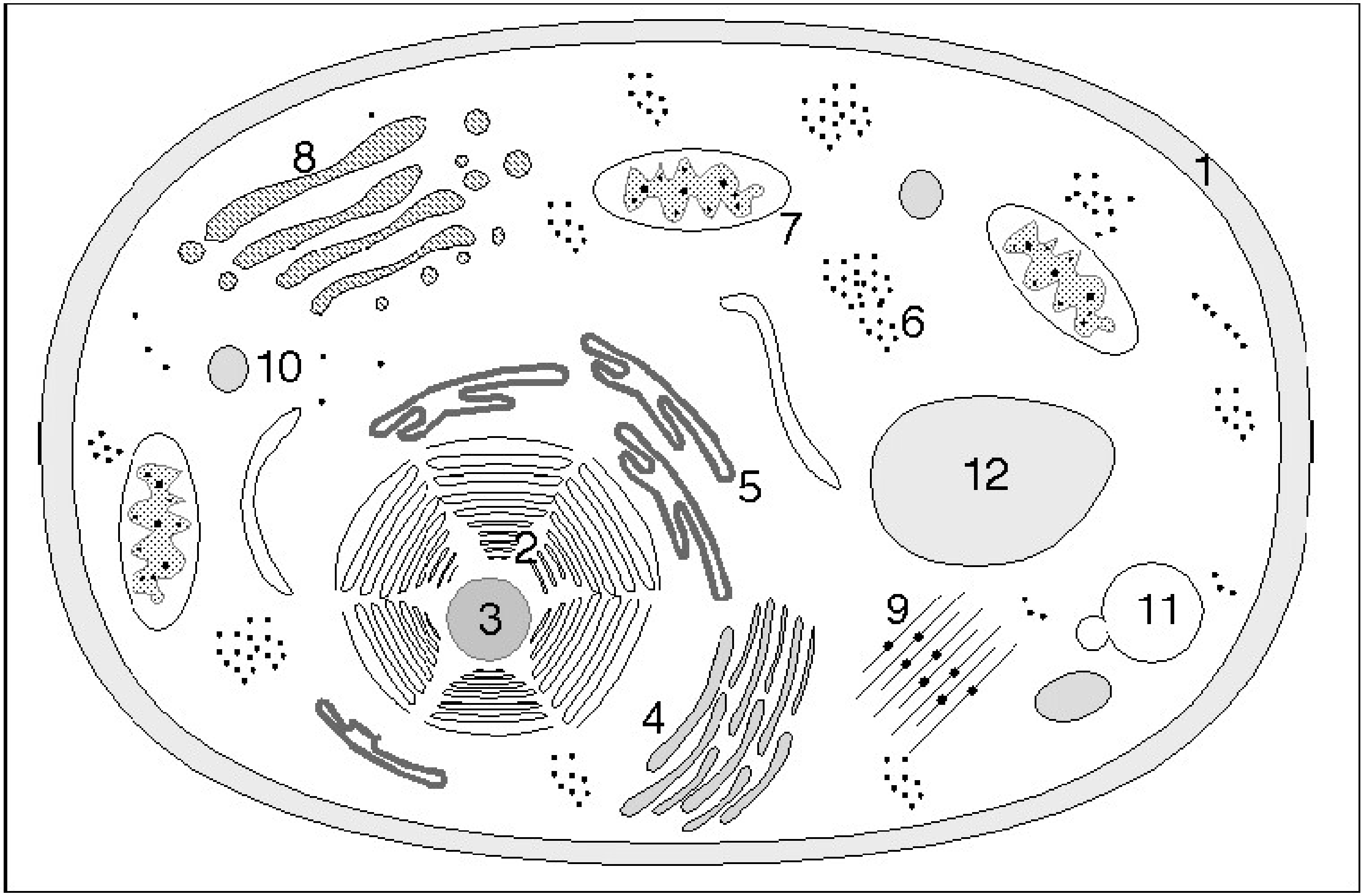
$E \approx 42$ eV - 41.36 MeV

Imaging with x-rays

basics of electromagnetic waves – x-rays:

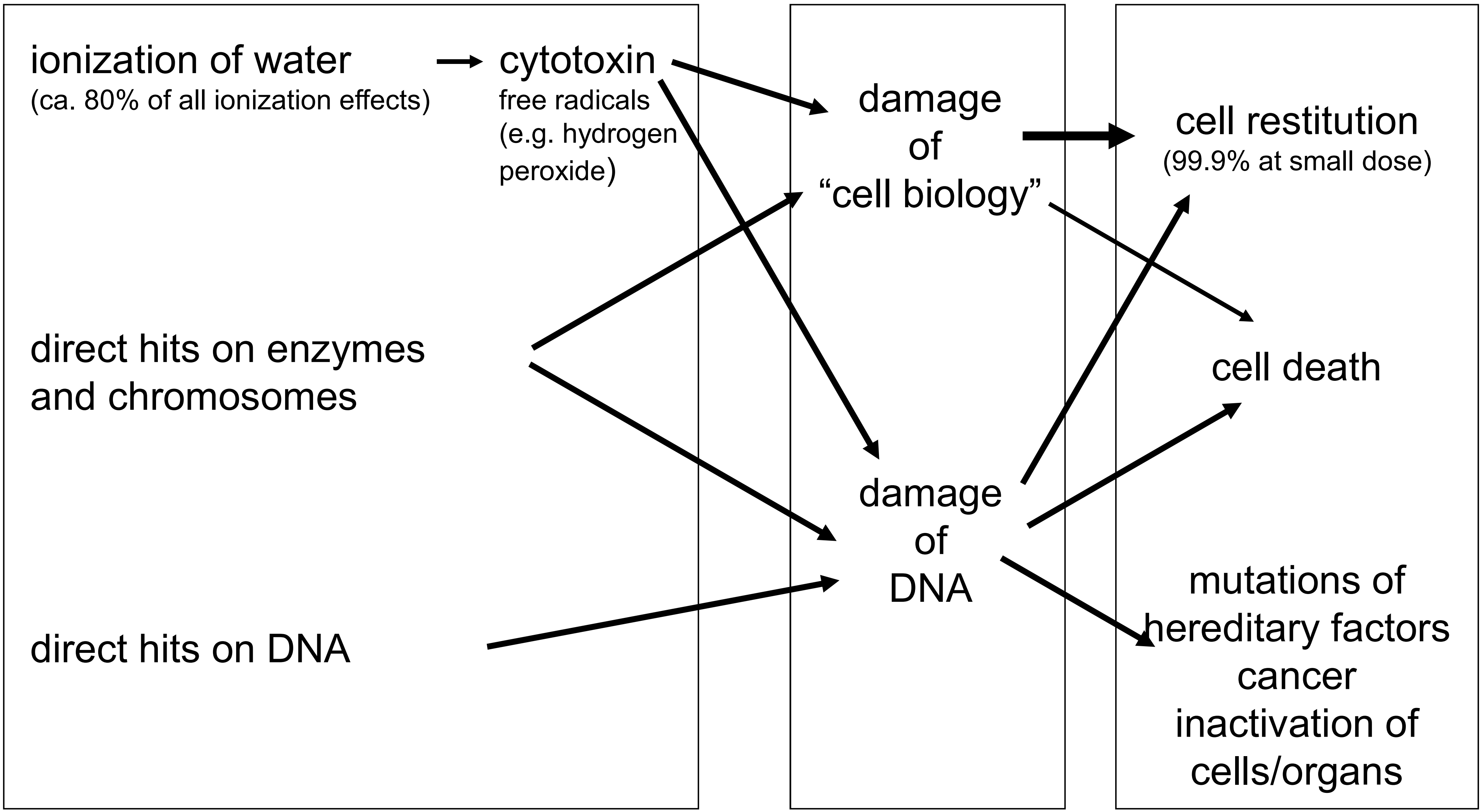
Frequenz			Wellenlänge	
	$3 \cdot 10^{14}$		10^{-6}	
	$3 \cdot 10^{13}$	Höhenstrahlen ($<10^{-5}$ nm)	10^{-5}	
	$3 \cdot 10^{12}$		10^{-4}	
	$3 \cdot 10^{11}$		10^{-3}	
GHz	$3 \cdot 10^{10}$	Röntgen-, Gammastrahlen (10^{-5} - 10 nm)	10^{-2}	nm
	$3 \cdot 10^{9}$		10^{-1}	
	$3 \cdot 10^{8}$		1	
	$3 \cdot 10^{7}$		10	
	$3 \cdot 10^{6}$	UV-Strahlung (10 - 400 nm)	100	
	$3 \cdot 10^{5}$	sichtbares Licht (400-780 nm)	1	
	$3 \cdot 10^{4}$		10	µm
GHz	$3 \cdot 10^{3}$	Wärmestrahlen (Infrarot) (780 nm - 1 mm)	100	
	300		1	
	30		10	mm
	3	Mikrowellen (10 - 100 mm)	100	
	300	Fernsehen (2 m)	1	
MHz	30	Ultrakurzwellen (1-10m)	10	m
	3	Kurzwellen (10-80 m)	100	
	300	Mittelwellen (200-600 m)	10^{3}	
kHz	30	Langwellen (> 600 m)	10^{4}	
	3		10^{5}	
	300		10^{6}	
Hz	30		10^{7}	
	3		10^{8}	

biological impact of ionizing radiation:



1) Zellmembran	5) endoplasmat. Retikulum, rauh	9) Zentrosom
2) Zellkern	6) freie Ribosomen	10) Lipidtröpfchen
3) Nukleolus	7) Mitochondrien	11) Lysosomen
4) endoplasmat. Retikulum, glatt	8) Golgi-Vesikel	12) Vakuolen

biological impact of ionizing radiation:



biological impact of ionizing radiation:

cancer mortality per 10 mSv (1 rem) and 1.000.000 persons

leukemia	20
breast cancer	25
lung cancer	20
bone cancer	5
thyroid cancer	5
other	50
total	125

“natural” cancer mortality ~ 200.000 per 1.000.000 persons
natural radiation exposure ~ 2.2 mSv per year

biological impact of ionizing radiation:

natural radiation exposure:

cosmic radiation		*0.3 mSv/a*
terrestrial radiation		*0.5 mSv/a*
outdoors	0.43 mSv/a	
indoors	0.57 mSv/a	
incorporated radioactive substances		*0.3 mSv/a*
inhalation of Radon reaction products		*1.3 mSv/a*

biological impact of ionizing radiation:

artifical radiation exposure:

medical applications		*1.4 mSv/a*
x-ray diagnosis	1.30 mSv/a	
nuclear medicine	0.07 mSv/a	
radiotherapy	0.03 mSv/a	
fallout atomic bomb tests		*0.01 mSv/a*
consumer goods, research		*0.03 mSv/a*
technical sources	0.01 mSv/a	
industrial products	0.01 mSv/a	
stray radiation emitters (TV)	0.01 mSv/a	
job-related exposition		*0.01 mSv/a*
non-military use of nuclear power		*0.01 mSv/a*
total dose (natural + artifical)		***1 - 4 mSv/a***

biological impact of ionizing radiation:

radiation damage

deterministic: above some threshold, severity of damage increases with dose

stochastic: damage probability increases with dose, no threshold

somatic damage: affects the whole body (e.g. malfunction of organs)

genetic damage: recessive mutations affecting succeeding generations (fusion of mutated genes and accumulation in population)

base items and units in dosimetry:

energy dose	$D = \frac{\text{absorbed energy}}{\text{mass}} = \frac{dW}{dm}$ unit: Gy (Gray) 1Gy = 1J/kg Formerly: rd (Rad) 1 rd = 0.01 Gy
absorbed dose rate	$\dot{D} = \frac{\text{energy dose}}{\text{time}} = \frac{dD}{dt}$ unit: Gy/sec (or /min, /h, /d, /a) formerly: rd/sec

base items and units in dosimetry:

ion dose	$J = \frac{\text{accumulated charge quantity}}{\text{mass}} = \frac{dQ}{dm}$ unit: C/kg = As/kg formerly: R (Röntgen) 1 R = $2.58 \cdot 10^{-4}$ C/kg
ion dose rate	$\dot{J} = \frac{dJ}{dt}$ unit: A/kg

base items and units in dosimetry:

equivalent dose

$$H = q \cdot D$$

$$q = \text{radiation - type - dependent factor}$$

unit: Sv (Sievert) 1 Sv = 1 J/kg

formerly: rem 1 rem = 10 mSv

type of radiation	q
x-rays and gamma rays	1
beta rays	1
alpha rays	20
neutron rays	10

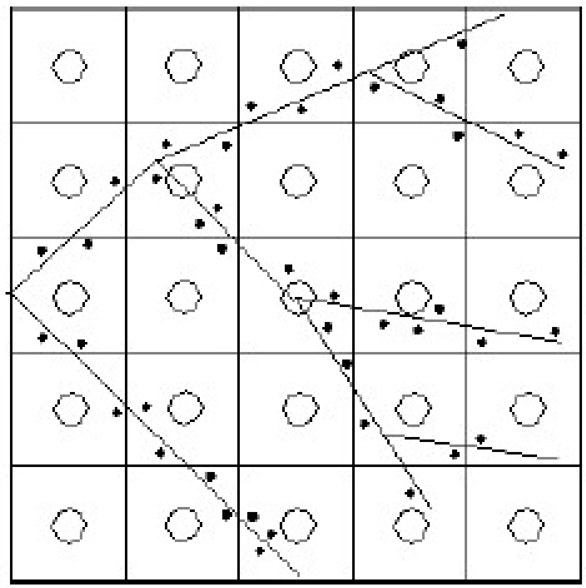
biological dose rate

$$\dot{H} = \frac{dH}{dt}$$

unit: Sv/sec (or /min, /h /d /a)

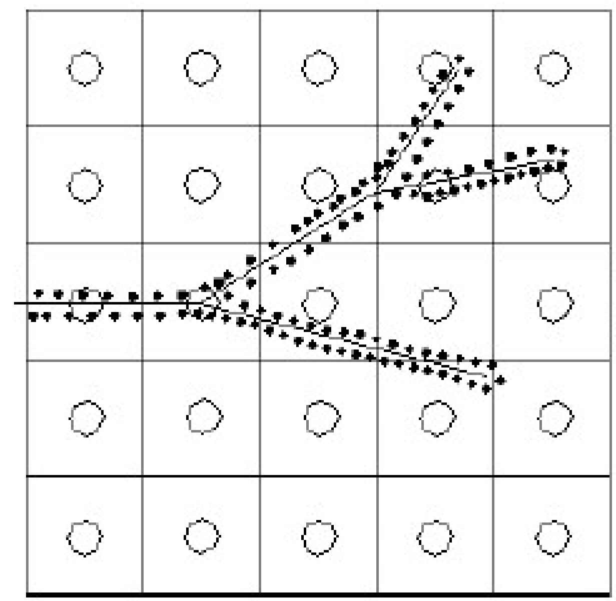
microscopic distribution of deposited energy of various radiation types:

weakly ionizing particle (e.g. e^-)



tissue cells

strongly ionizing particle (e.g. e^+)



tissue cells

radiation protection:

in general:

- avoid unnecessary exposition to ionizing radiation

- if not avoidable:
  ALARA (**A**s **L**ow **A**s **R**easonably **A**chievable)

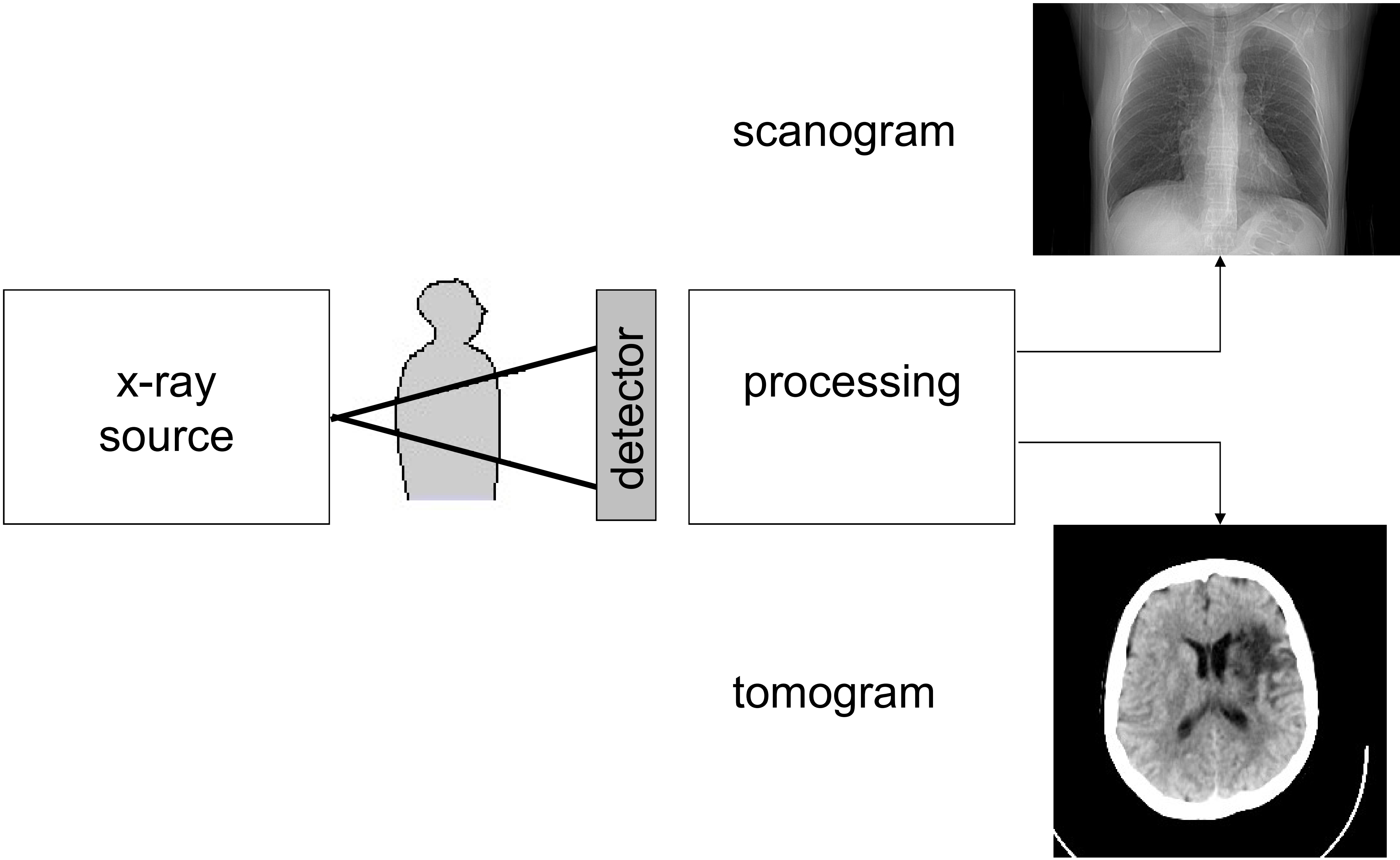
- obligation to inform patients

dose D decreases with squared distance A $\qquad D \sim 1/A^2$

dose D increases with exposition time t_{exp} $\qquad D \sim t_{exp}$

generation
of
x-rays

Imaging with x-rays



generation of x-rays:

basic principle: ***photoelectric effect***

- heating of cathode (U_H) → free electrons

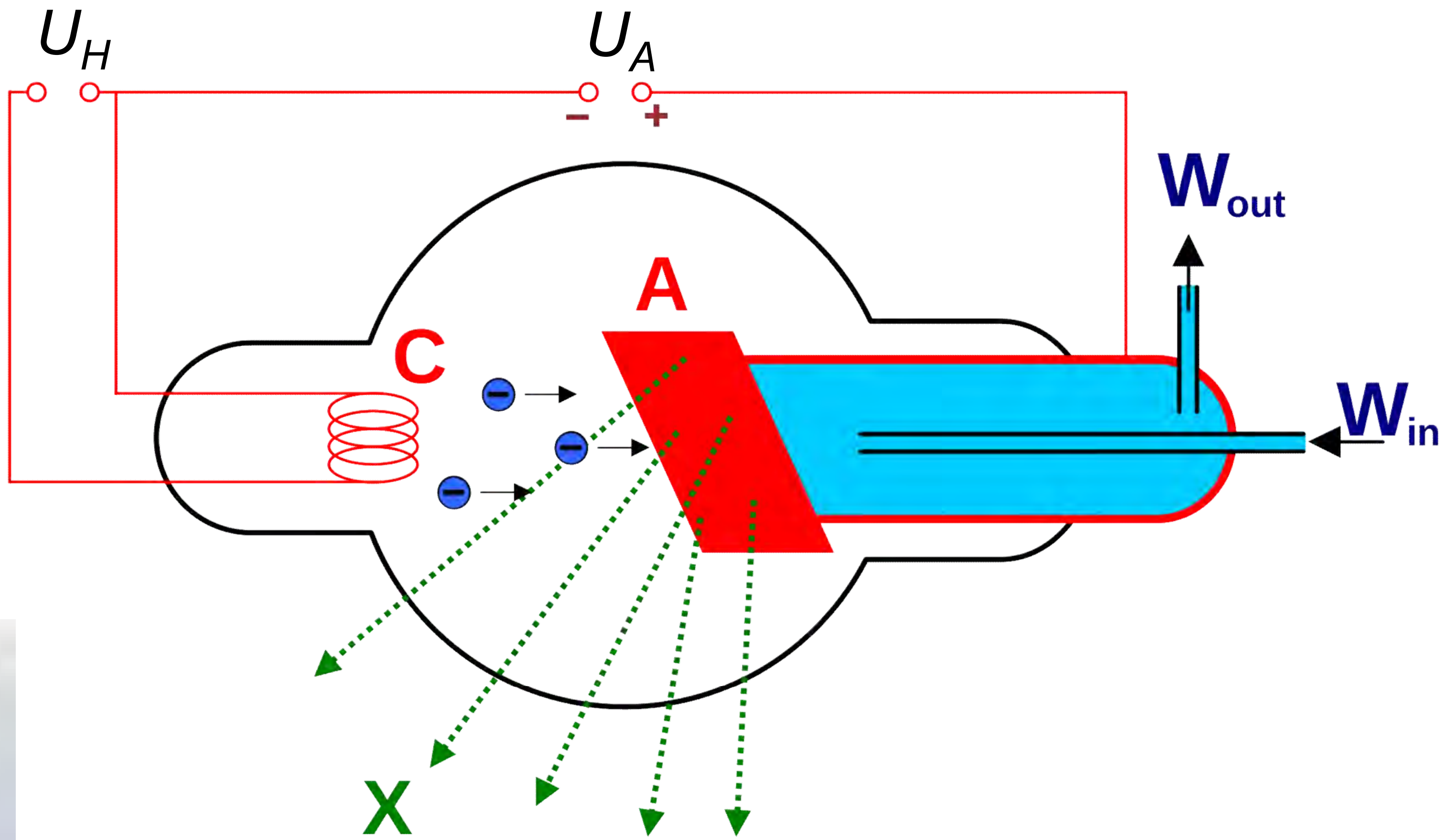
- acceleration of electrons in electrical field (U_A: 100 – 150 kV)

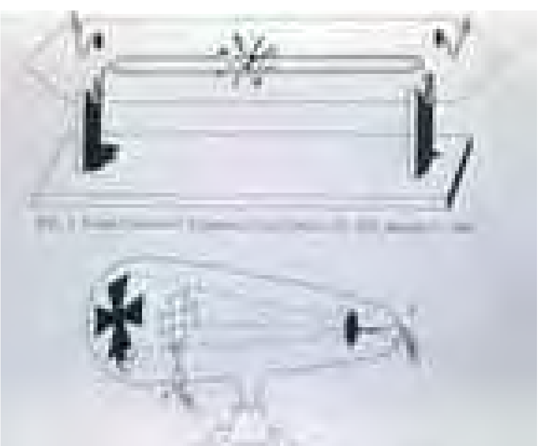
- deceleration of electrons in anode
(conversion: 99 % E_{kin} → heat, 1 % → x-rays)

- Bremsstrahlung, characteristic radiation

- vacuum (< 10^{-5} mbar); avoid interactions with molecules in air

generation of x-rays:




W. Crookes 1904

C: cathode (-)
A: anode (+)
W_{in} and W_{out}: water inlet and outlet of the cooling device

x-ray frequency:

energy of accelerated electrons:

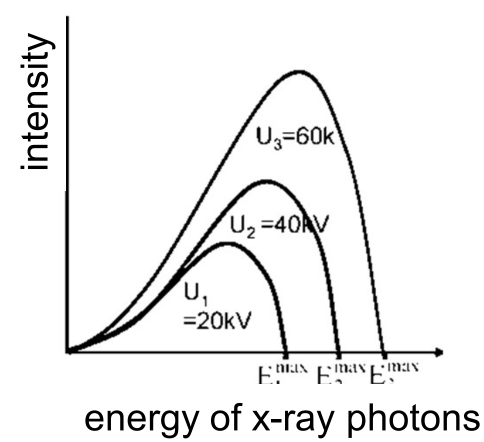
$$E_{kin} = e \cdot U_A$$

with $E_{Photo} = h \cdot \nu$

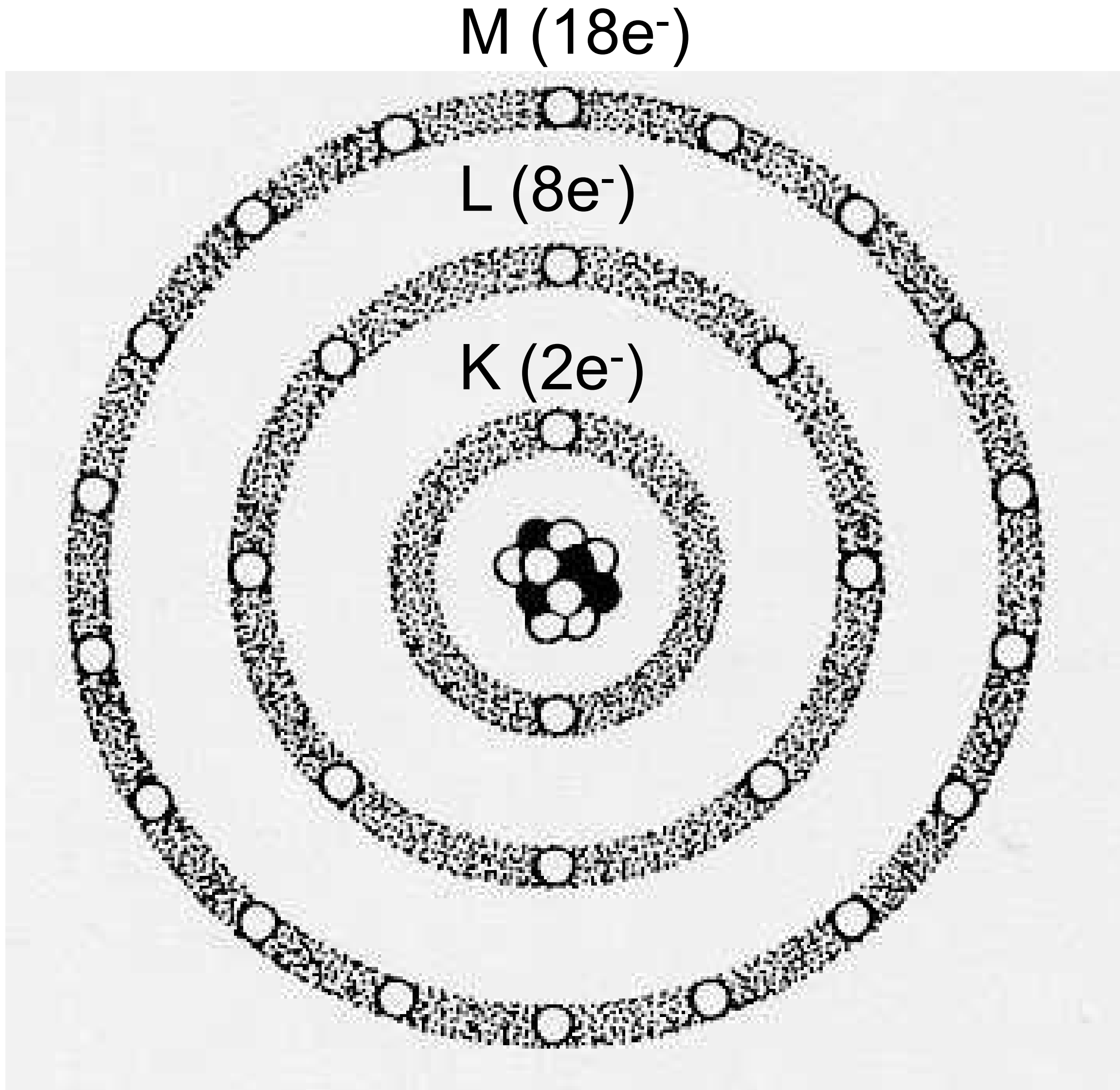
$$\Rightarrow \nu = \frac{e}{h} \cdot U_A$$

x-ray frequency depends linearly of acceleration voltage U_A

U_A	λ=1/υ	radiation strength
1 kV	1.242 nm	weak
10 kV	0.124 nm	medium
100 kV	0.012 nm	hard



Bohr model

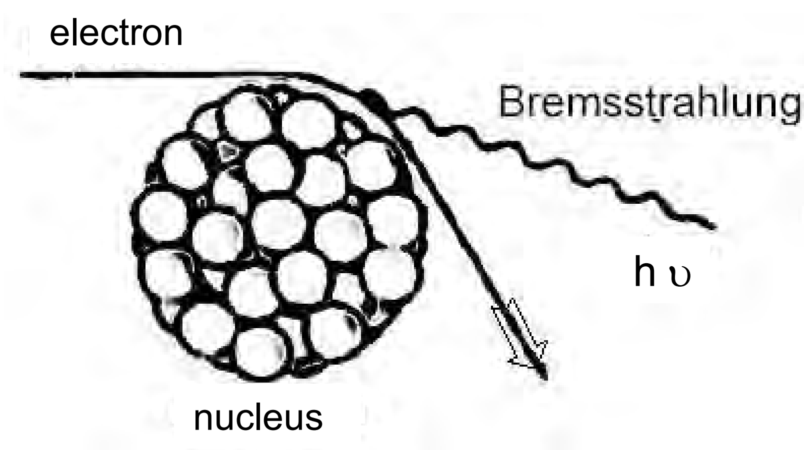


x-ray energy

1. ***Bremsstrahlung***:

accelerated electrons approach nucleus (between nucleus and K-shell)

- deflection (due to Coulomb potential of nucleus and shell electrons)
- deceleration (E_{kin} converted into e.m. energy)
- emission of energy in form of “Bremsstrahlung“



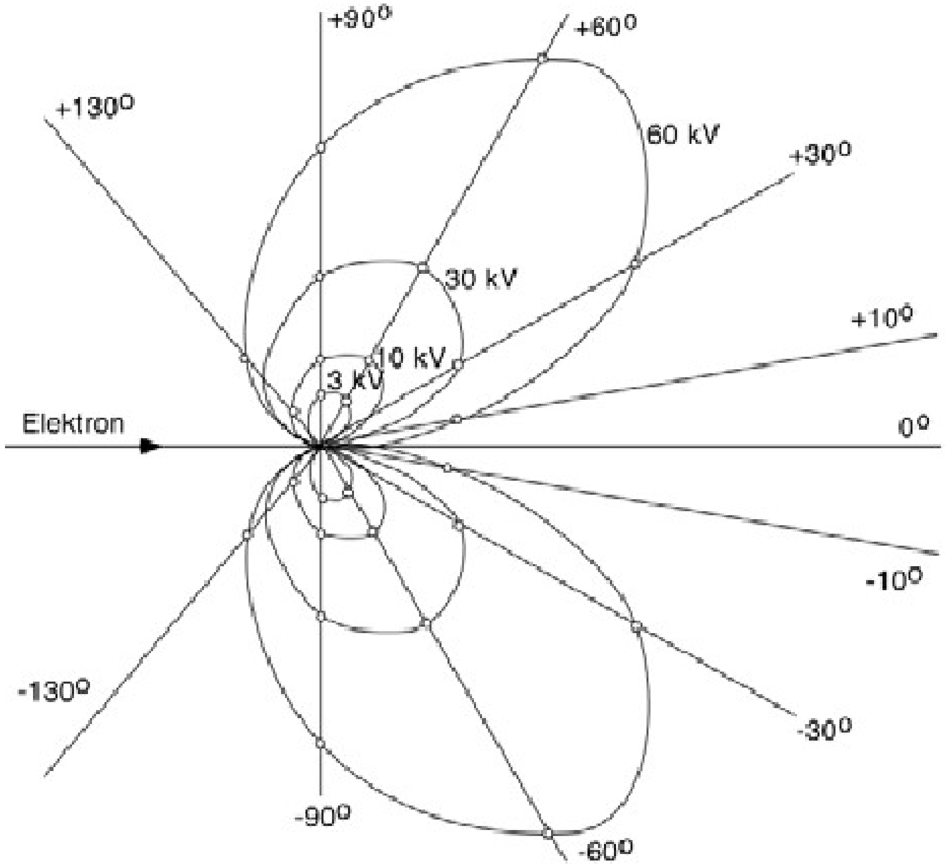
energy of Bremsstrahlung depends on trajectory of electrons

⇒ broad energy spectrum !

x-ray energy

1. Bremsstrahlung:

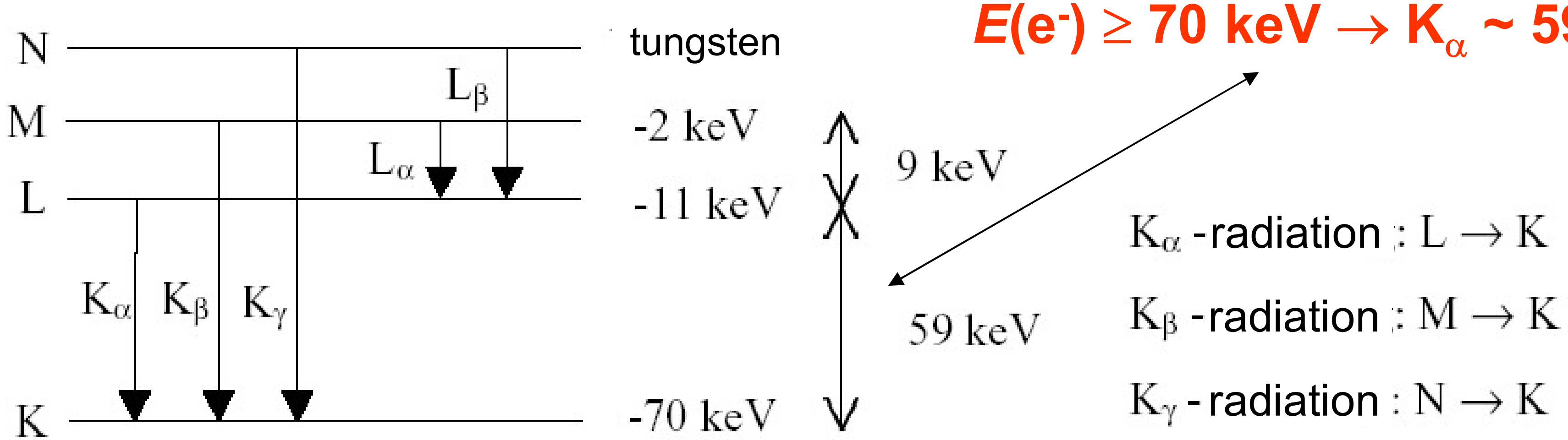
spatial distribution of intensity of Bremsstrahlung (“radiation lobes”)



x-ray energy

2. ***characteristic radiation***:
   accelerated electron strikes bound electron from K- (or L-)shell
   ⇒ ionization
   vacant energy level (core hole) taken by electron from outer shell
   emission of energy difference ($h\upsilon=E_m-E_n$) as quantized photon with characteristic frequency υ

$E(e^-) \geq$ 70 keV → K_α ~ 59 keV



energy of characteristic radiation solely material-dependent !

Energie der Röntgenstrahlung:

*2. **characteristic radiation***:

energy of K_{α}-radiation (Moseley's law):

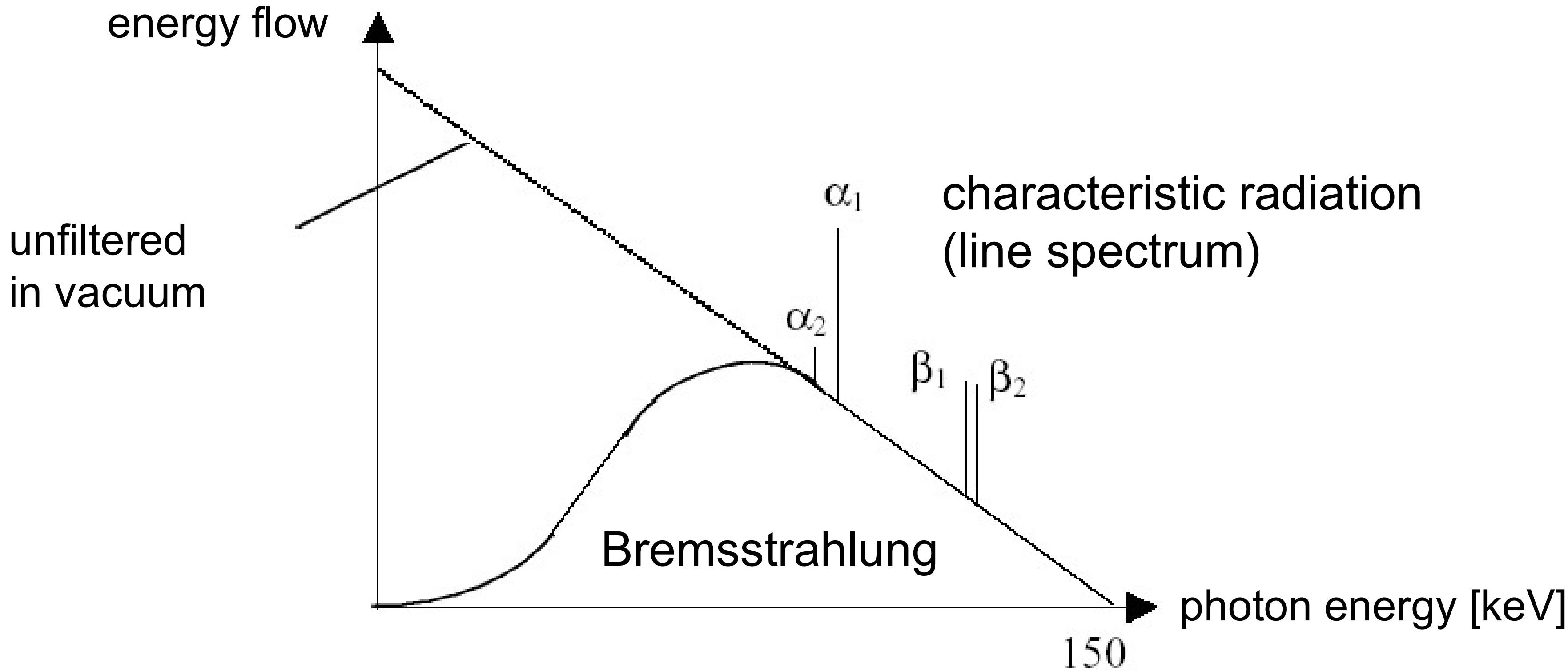
$$E_{K\alpha} = \frac{3}{4} \cdot R_{\infty}(Z-1)^2$$

$$\text{with } R_{\infty} = \text{Rydberg constant } (3.29 \cdot 10^{15} s^{-1})$$

$$Z = \text{atomic number}$$

Energie der Röntgenstrahlung:

3. full energy spectrum:



generation of x-rays:

- frequency depends on acceleration voltage
- energy depends on material properties

⇒ requirements for anode material:

- high atomic number Z (yield increases with Z)
- high melting point T_{max}
- high heat conductivity κ

- measure for quality = $Z \cdot T_{max} \cdot \kappa$

mostly used: tungsten or tungsten-rhenium

generation of x-rays:

quality criteria for x-ray sources in medical imaging

- high power $\Rightarrow$ short exposition times
- small focus $\Rightarrow$ acuity
- adjustable energy of quanta $\Rightarrow$ contrast
- low production costs
- low maintenance, long lifetime

generation of x-rays:

quality criteria for x-ray sources in medical imaging

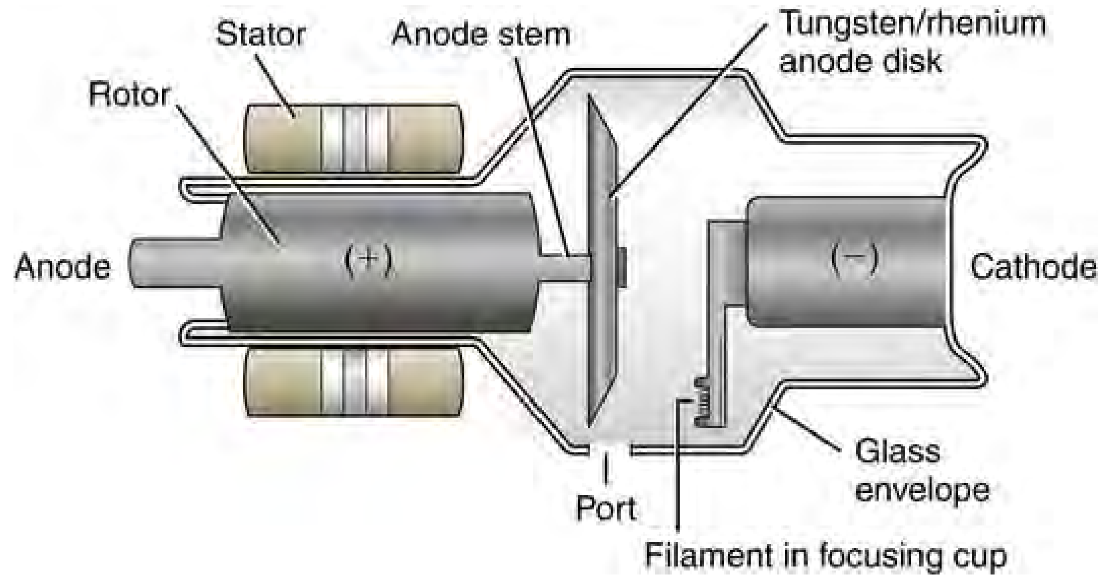
a high power and small focus

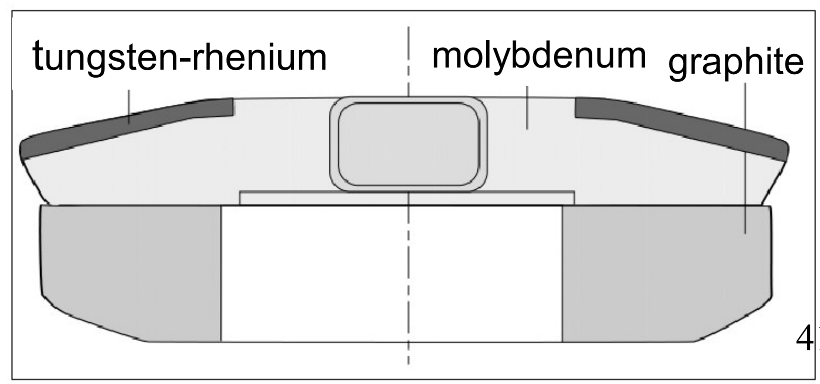
can be achieved with a

tilted anode

and with a

rotating anode (heat dissipation)



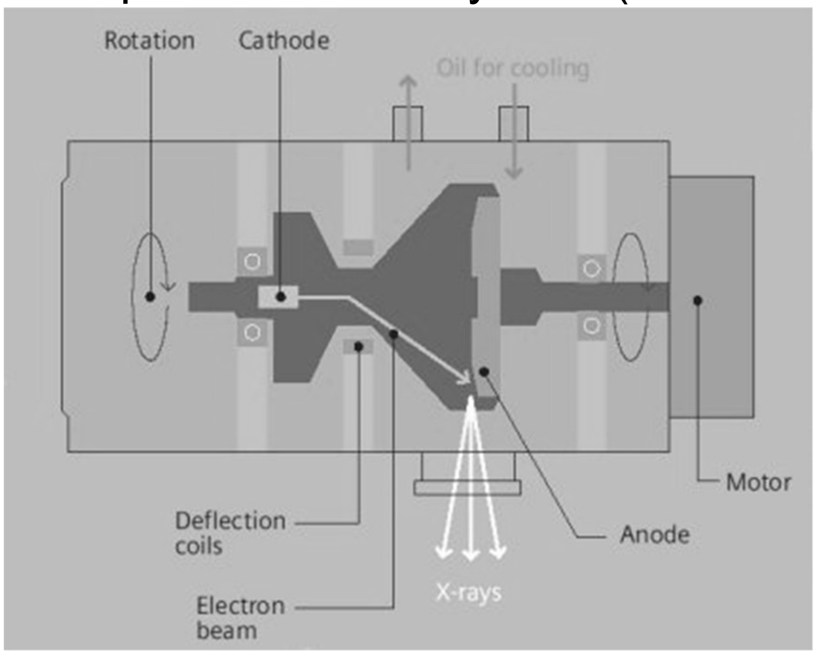


anode material:

element	atomic number Z	max. temperature T_{max} [°C] @ $1.33 10^{-2}$ Pa	heat conductance κ [W cm^{-1} K^{-1}]	stationary anodes		rotating anodes		
				$Z T_{max} \kappa$	order	$\sqrt{\kappa \rho c}$	$Z T_{max} \sqrt{\kappa \rho c}$	order
Cu	29	1032	3,98	119113	8	3,68	110135	10
Mo	42	2167	1,38	125599	7	1,88	171106	8
Ag	47	832	4,18	163450	4	3,18	124350	9
Ta	73	2587	0,55	103868	9	1,13	213402	6
W	74	2757	1,3	265223	1	1,81	369273	1
Re	75	2557	0,71	136160	6	1,38	264650	4
Os	76	2280	0,87	150754	5	1,77	306706	3
Ir	77	2220	1,46	249572	3	2,06	352136	2
Pt	78	1742	0,71	96472	10	1,41	191585	7
Au	79	(1063)	3,14	263687	2	2,81	235975	5
U	92	(1132)	0,25	26036	11	0,75	78108	11

generation of x-rays:

example: Straton x-ray tube (Siemens, 2003)



direct cooling of anode
mechanics outside vacuum
rotation time: 0.37 sec
sub-mm volumen scans @ 500 mAs for 20 sec (64 mm/sec)
dose reduction
indep. of patient height and anatomy

efficiency η and radiation power *D*:

$$\eta \equiv \frac{\text{radiation power}}{\text{electrical power}} = k \cdot Z \cdot U_A \quad [\%]$$

where

k = $1.1 \cdot 10^{-9}$ [V^{-1}]

Z = atomic number of anode material

U_A = acceleration voltage

example: tungsten anode, Z=74, U_A=125 kV $\Rightarrow \eta = 1.02$ %

(in praxis: < 1% due to filtering and suppression; reminder: heat)

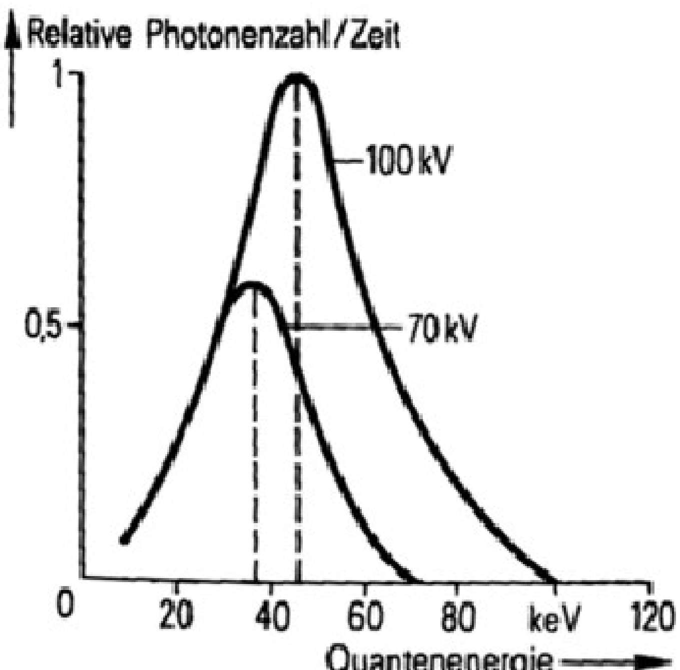
$D \equiv Z \cdot I \cdot U_A^2$

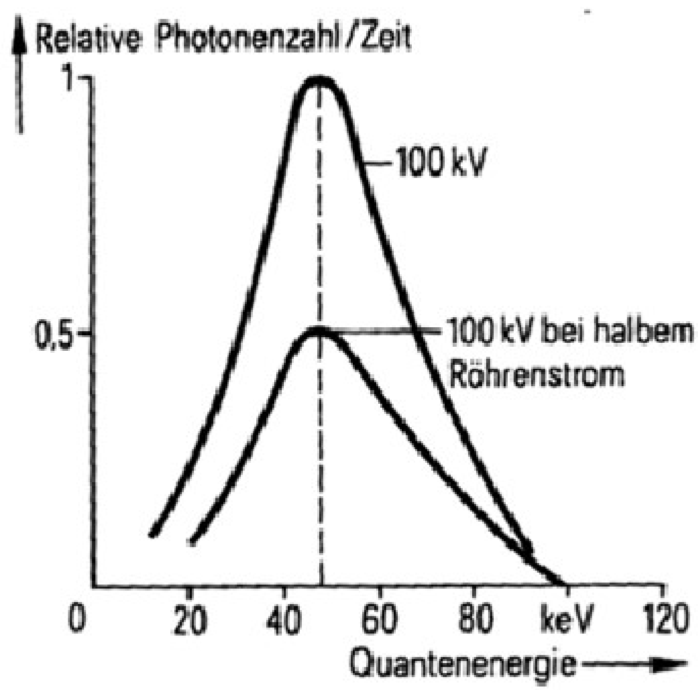
with

I = current in tube (mostly fixed!)

Impact of acceleration voltage and tube current:

- flux density of x-ray radiation: $\psi \sim Z \cdot I \cdot U_a^n$
- acc. voltage determines “strength“ of radiation
  potential impact (unfiltered: *n*=2; with filter up to *n*=5)
- tube current determines number of photons/sec
  linear impact





x-rays:

classification

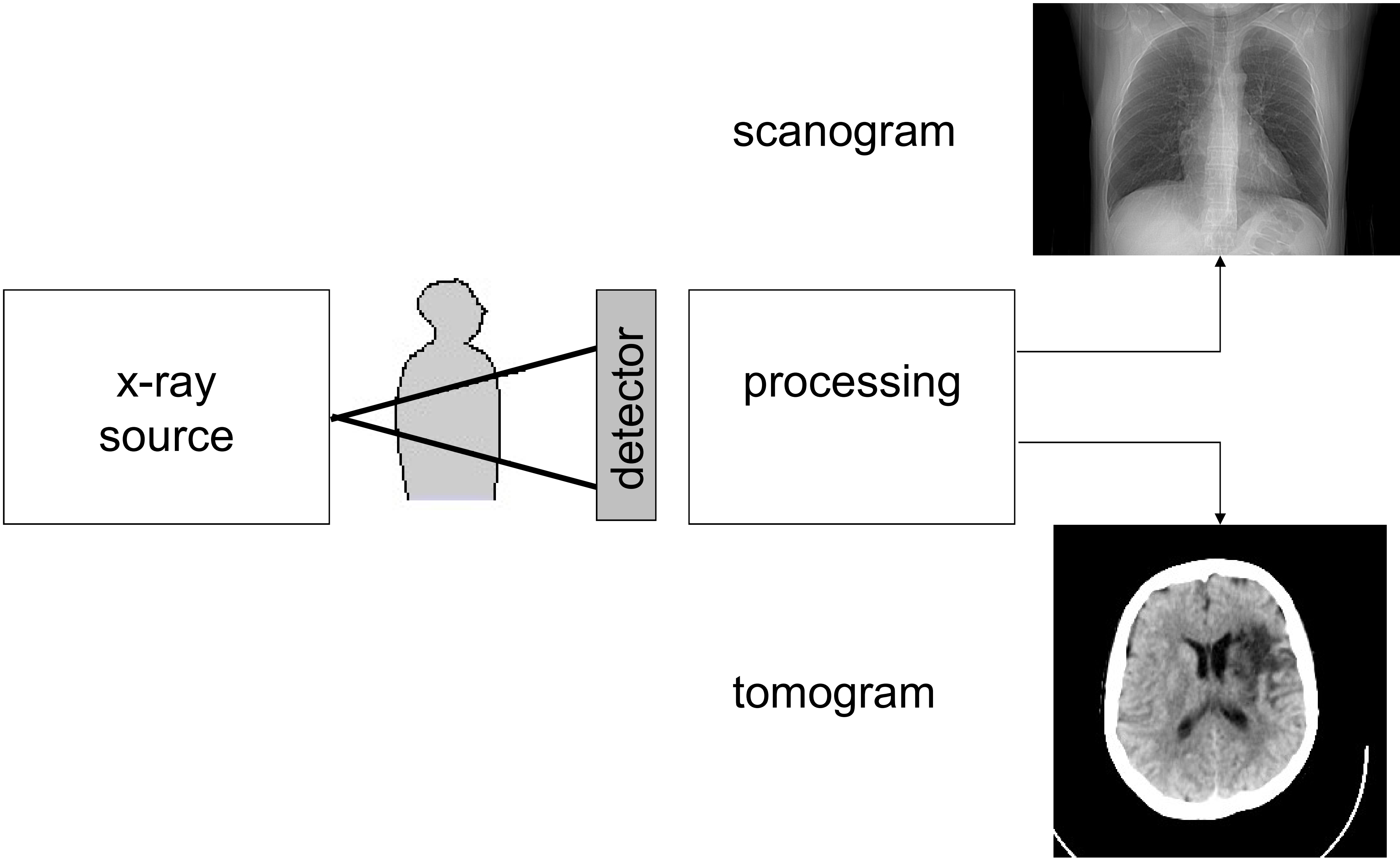
radiation strength	λ_{min} - λ_{max} [nm]	f_{min} - f_{max} [GHz]	E [keV]
extra weak	0.25 - 0.06	$1.2\ 10^9$ - $3.3\ 10^9$	5.0 - 13.6
weak	0.06 - 0.02	$3.3\ 10^9$ - $1.5\ 10^{10}$	13.6 - 62
medium	0.02 - 0.01	$1.5\ 10^{10}$ - $3.0\ 10^{10}$	62 - 124
hard	0.01 - 0.005	$3.0\ 10^{10}$ - $6.0\ 10^{10}$	124 - 248
extra hard	< 0.005	> $6.0\ 10^{10}$	> 248

typical CT x-ray tube:
acceleration voltage 120 kV
tungsten anode: ~ 20 - 120keV

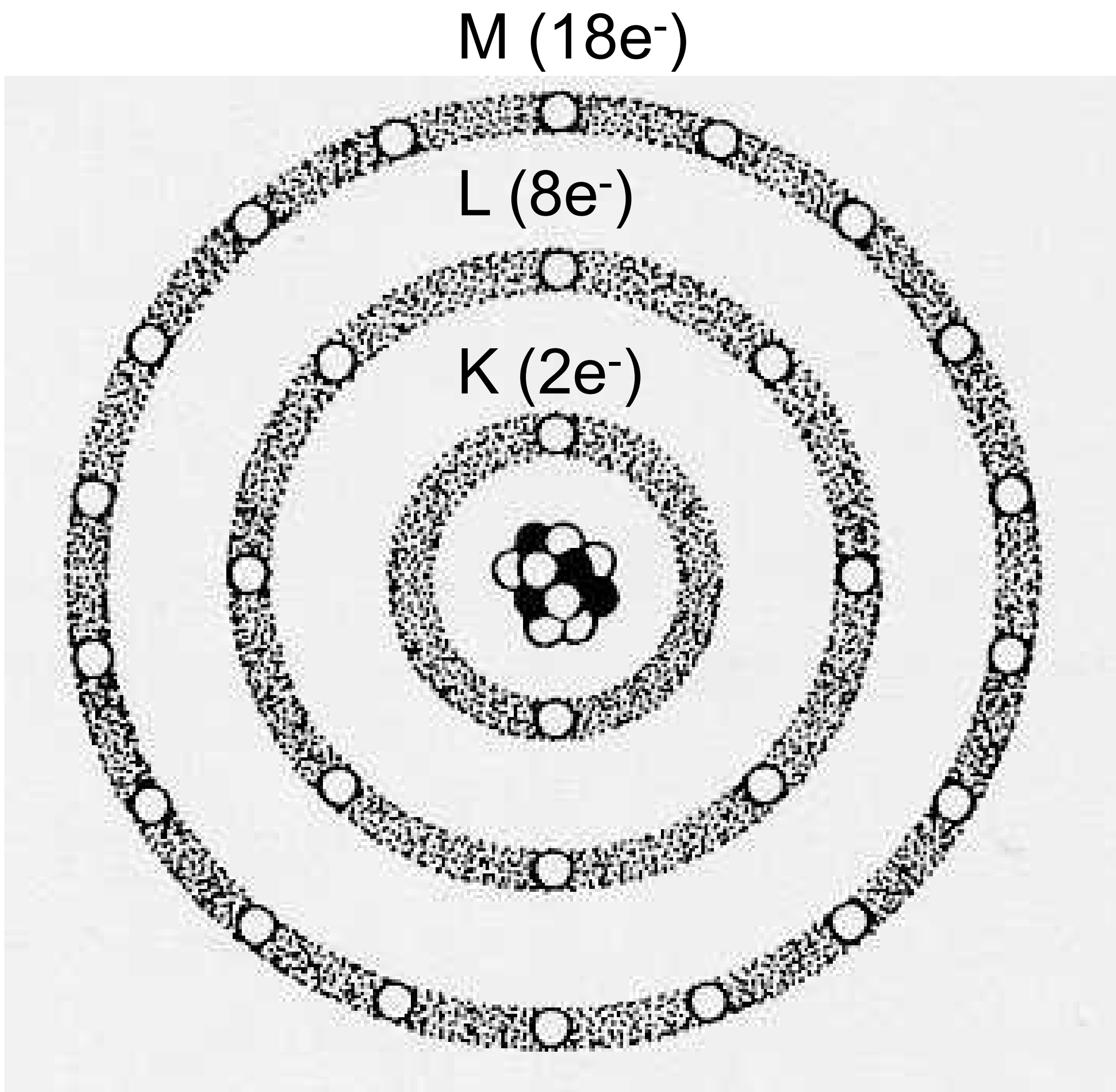
onset of ionization of living tissue at 15 eV !!

interactions with matter

Imaging with x-rays

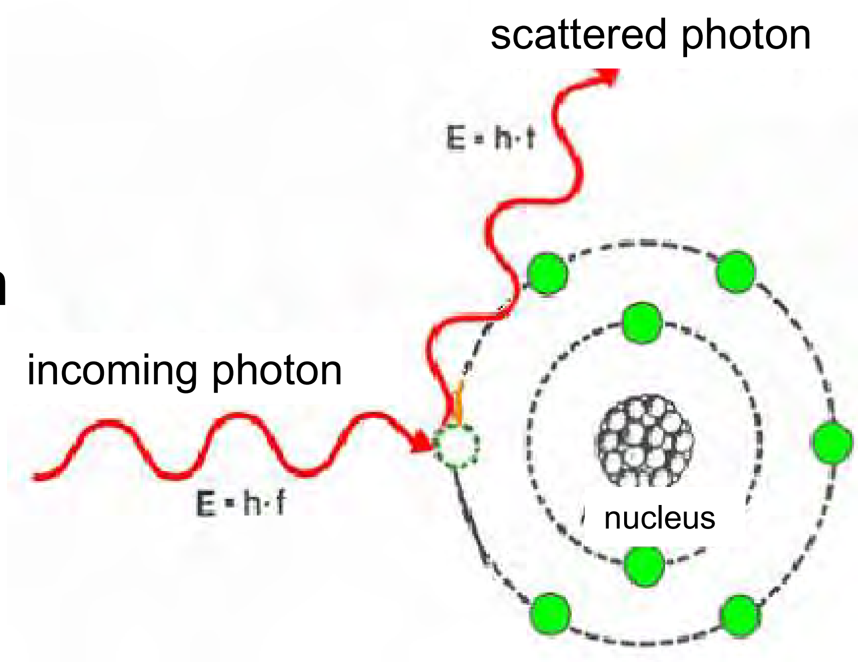


Bohr model



interaction with matter ***coherent scattering***

- incoming photon interacts with object and changes its trajectory, but
  - no absorption
  - no change of energy of photon



- size of scattering body << wave length

- probability of occurrence:
  ~ 5 % of applied x-rays

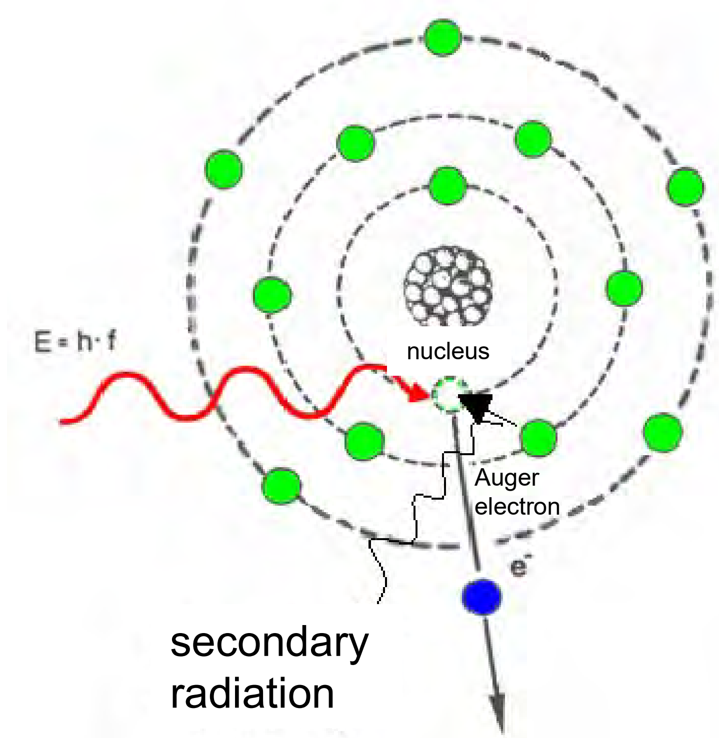
- disadvantageous for imaging: background noise (“*film fog*“)

interaction with matter ***photo effect***

- photon transfers its full energy to shell electron (*K*- and *L*-shell)
- effect depends on photon energy
- effect size ~ $1/E^3$ (at high energies)

- energy balance: $h \cdot f = 1/2\, m_e v^2 + E_a$

- secondary radiation when core hole is filled up with electron from outer shell (Auger electron)



probability of occurrence ~ Z^3

⇒ amplifies absorption differences of different tissues !

⇒ important for diagnostic radiology !

interaction with matter ***Compton effect***

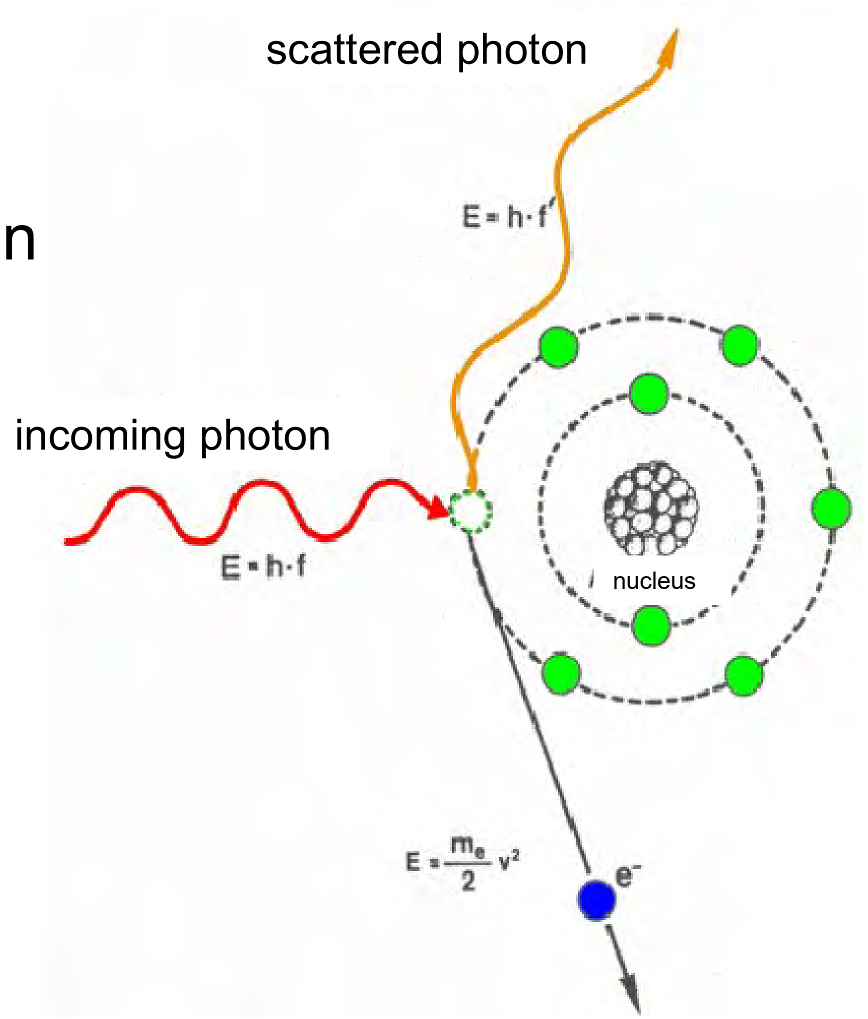
- $E_\gamma < 1.022$ MeV:

- amount of energy transferred to electron depends on scattering angle φ

- higher probability for electrons in outer shells (binding energy irrelevant)

- energy balance:

$$E_\gamma + E_{0e} = E_{\gamma'} + E_e$$



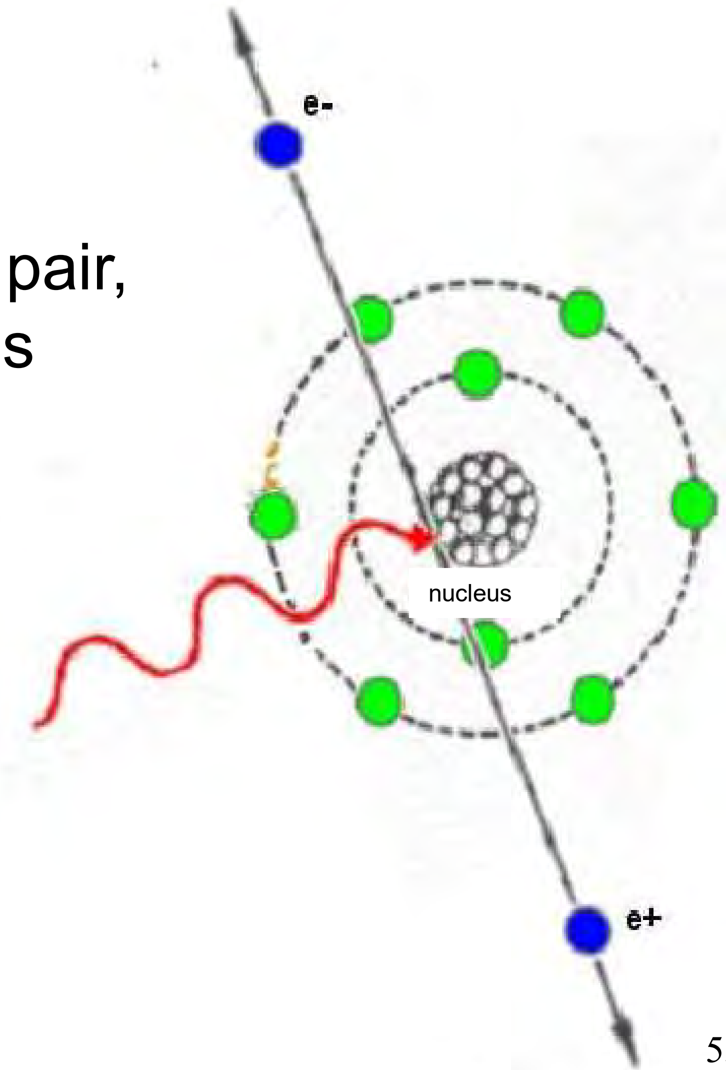
interaction with matter ***pair production***

- $E_\gamma \geq 1.022$ MeV:

- production of an electron (e^-) / positron (e^+) pair, if energy of incoming photon equals/exceeds twice the rest energy of an electron

- energy balance:

$$E_\gamma = E_e + E_p + 2m_e c^2$$



summary of types of interaction

1. photo effect: γ-quant transmits its full energy to a shell-electron
$\Rightarrow$ **absorption**

2. Compton effect: scattering on electron; scattered radiation has lower energy and different trajectory
$\Rightarrow$ **scattering**

3. pair production: radiation ($E \geq 1.022$ MeV and if near nucleus) is being transformed into electron and positron
$\Rightarrow$ **transformation radiation $\rightarrow$ matter**

$\Rightarrow$ **attenuation = absorption + scattering**

quantitative assessment of attenuation **absorption law**

particle rate: $$N = \frac{\text{particles}}{\text{time}} = \frac{\Delta n}{\Delta t}$$

intensity: $$I = \frac{\text{energy}}{\text{area} \cdot \text{time}} = \frac{E}{\Delta A \cdot \Delta t}$$

with mono-energetic radiation: $E = \Delta n \cdot E_\gamma$

$$\Rightarrow I = \frac{E_\gamma \cdot \Delta n}{\Delta A \cdot \Delta t} = \frac{E_\gamma}{\Delta A} \cdot N$$

$$\Rightarrow I \propto N$$

quantitative assessment of attenuation

absorption law (Lambert's law)

$$dN = -\mu \cdot N \cdot dx$$

μ = linear absorption coefficient

$$\Rightarrow N(x) = N_0 \cdot e^{-\mu \cdot x}$$

$$\Rightarrow I(x) = I_0 \cdot e^{-\mu \cdot x}$$

cf. Lambert-Beer law in optics

quantitative assessment of attenuation — absorption law

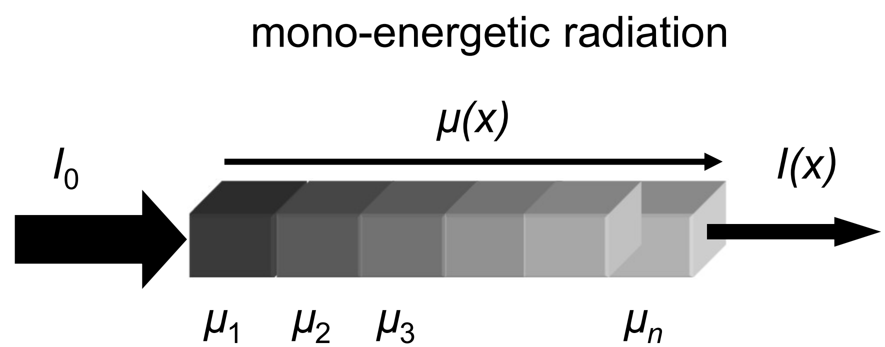
for **fixed** E_γ, we have in general:

$$dN = -\mu(x,y,E_\gamma,\rho,Z)\cdot N\cdot dx \Leftrightarrow \frac{dN}{N} = -\mu(x,y,E_\gamma,\rho,Z)\cdot dx$$

$$\Leftrightarrow \int_{N_0}^{N}\frac{1}{N}dN = -\int_0^x \mu(x,y,E_\gamma,\rho,Z)dx$$

$$\Leftrightarrow \ln\left(\frac{N}{N_0}\right) = -\int_0^x \mu dx \Leftrightarrow N = N_0\cdot\exp\left(-\int_0^x \mu dx\right)$$

$$\Rightarrow \boxed{I = I_0\cdot\exp\left(-\int_0^x \mu dx\right)}$$



attenuation coefficient μ **absorption law**

in general, we have:

$$\mu = \tau + \sigma + (\chi)$$

τ → photo effect σ → Compton effect (χ) → pair production

$$\mu = \frac{\rho}{A} \cdot N_A \cdot \mu' = \frac{\rho}{A} \cdot N_A \cdot (\tau' + \sigma')$$

μ' = atomic cross-section

where $\tau' = \tau'(E_\gamma, Z) = Z^5 \cdot C(Z) \cdot \tau_0'(E_\gamma)$

and $\sigma' = \sigma'(E_\gamma, Z) = Z \cdot \sigma_0'(E_\gamma)$

attenuation coefficient μ **absorption law**

in general: $$\mu_{ges} = \mu_{photo} + \mu_{compt} + \mu_{pair}$$ [cm^{-1}]

$$\mu_{photo} = \frac{Z^{3.8}}{E_\gamma^3}$$

$$\mu_{compt} \approx \frac{Z}{E_\gamma}$$

$$\mu_{pair} \approx Z^2 \ln E_\gamma$$

(alternatively: mass absorption coefficient μ/ρ [cm^2/g])

attenuation coefficient μ

Z dependence

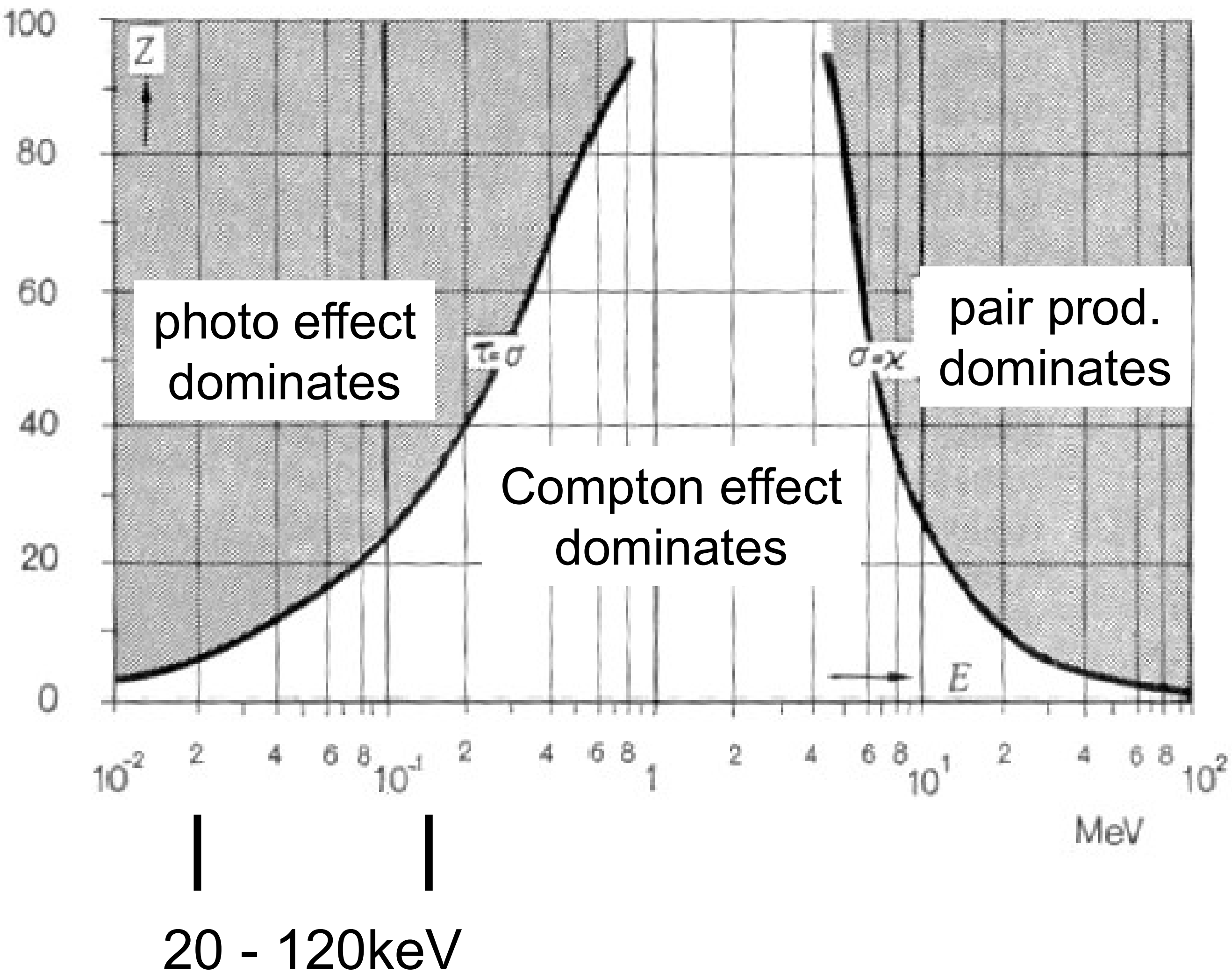
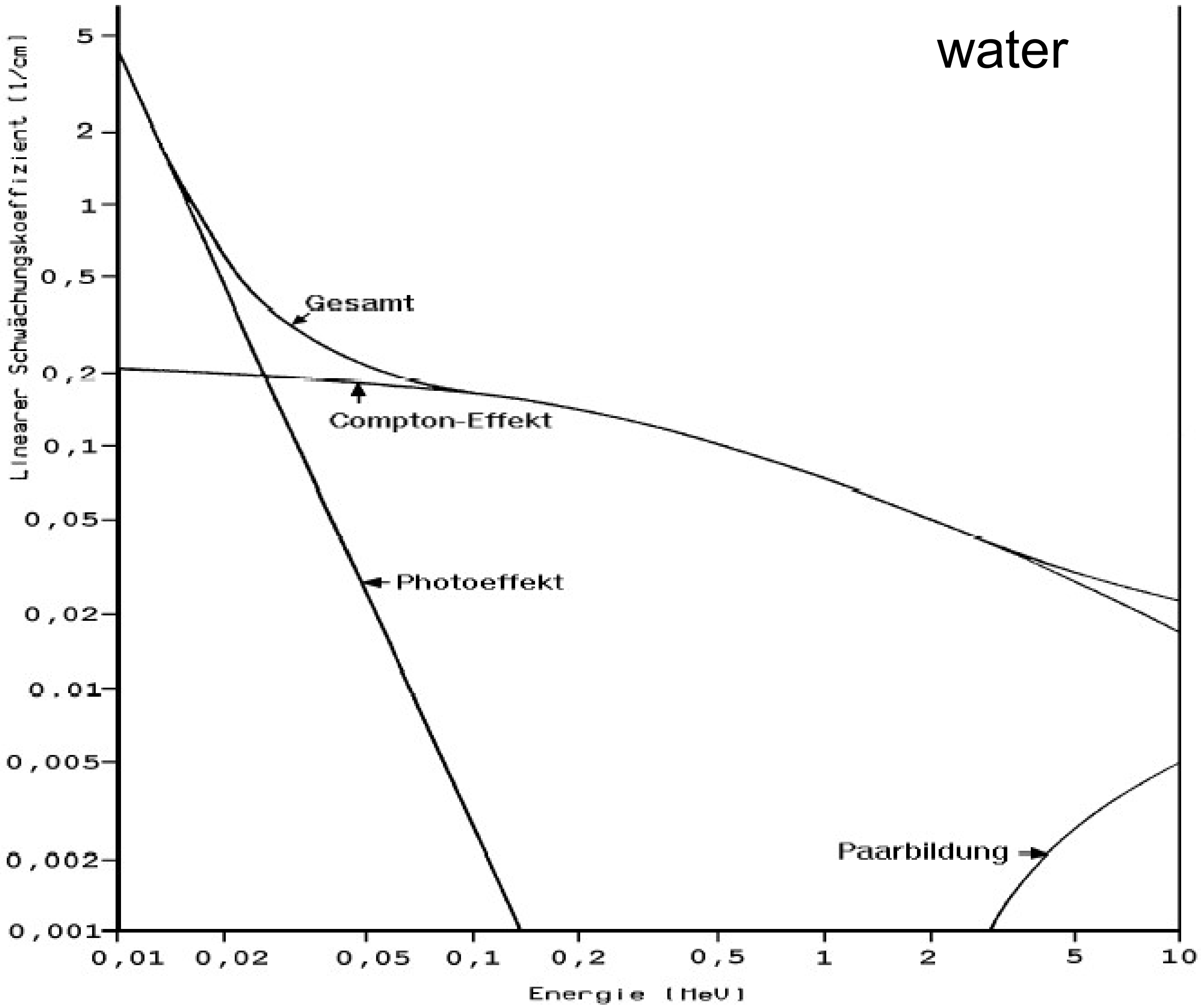
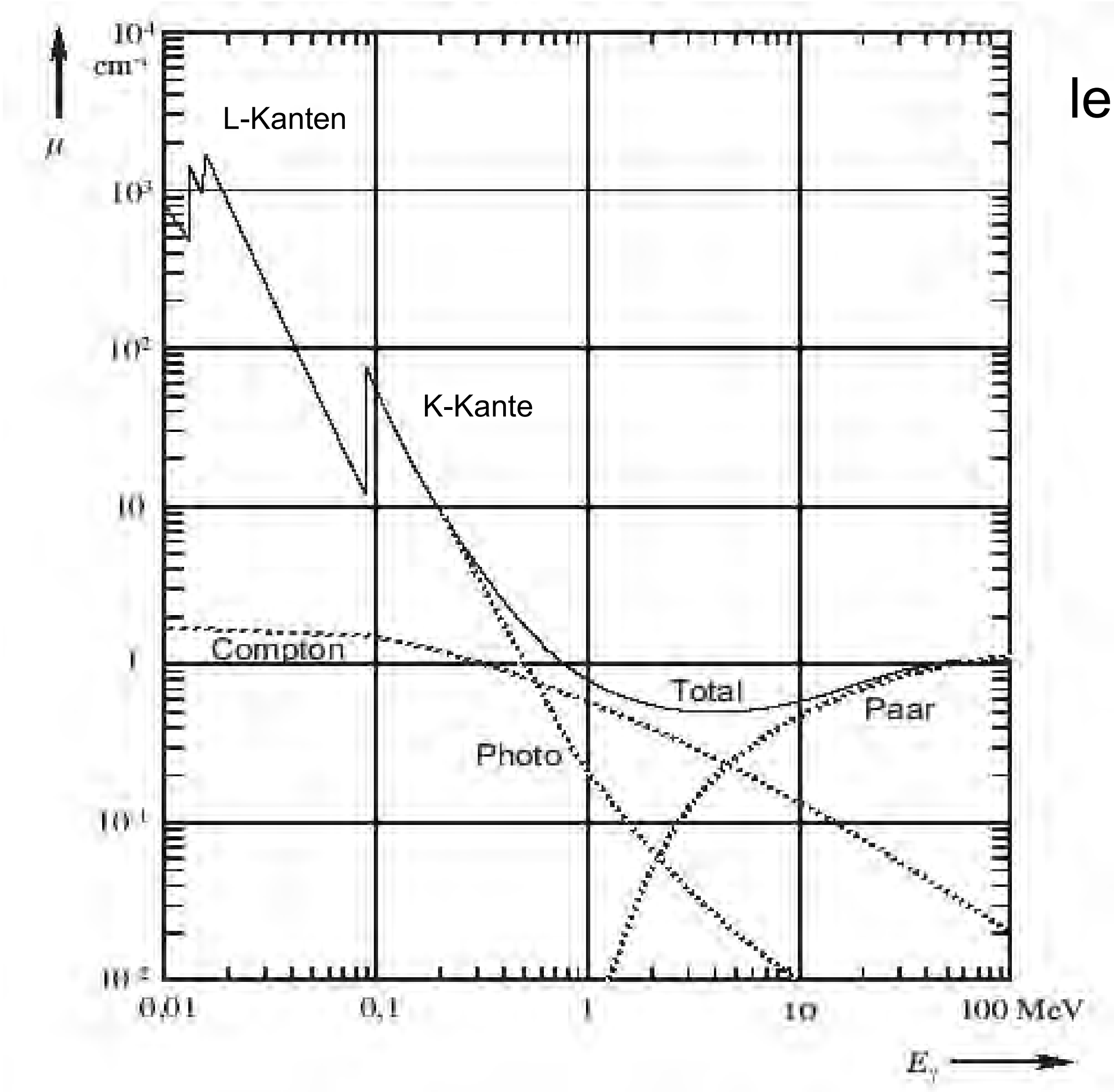


Abb. 6–15. Die Bereiche mit überwiegendem Photoeffekt, Compton-Effekt und Paarbildungseffekt werden durch die Kurven begrenzt, längs deren der Photoabsorptionskoeffizient τ gleich dem Streukoeffizienten σ bzw. der Streukoeffizient σ gleich dem Paarbildungskoeffizienten $\varkappa$ als Funktion der Ordnungszahl Z und der Photonenenergie E ist

attenuation coefficient μ **Z dependence**



attenuation coefficient μ **Z dependence**



lead

mass absorption coefficient **attenuation law**

number of absorbed resp. scattered particles is proportional to the density of the absorber

mass absorption coefficient μ´= μ/ρ [cm^2/g] equals absorption coefficient, if absorber has density ρ=1

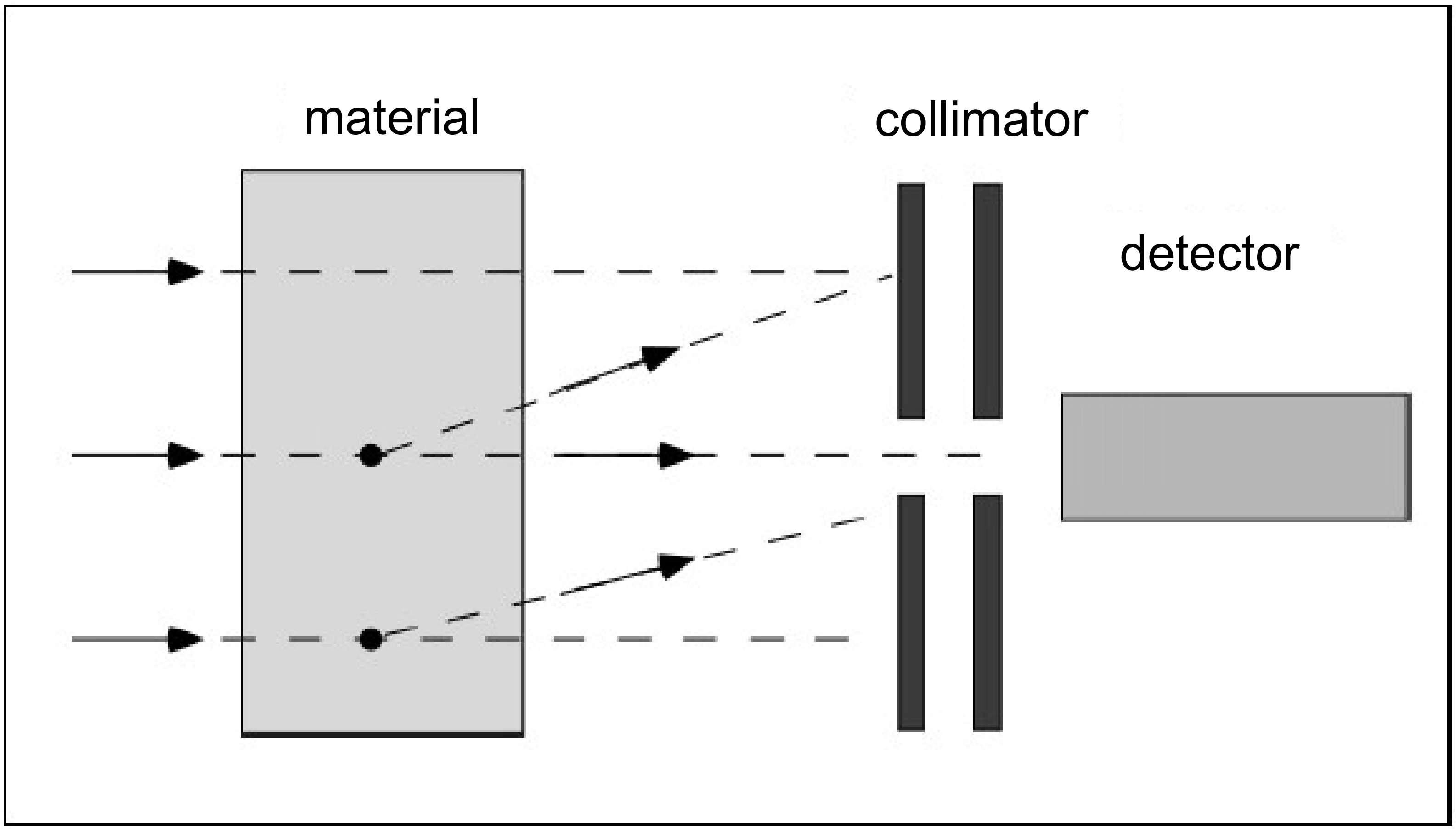
for mixed elements, we have:

$$\mu' = \sum_i \left(\frac{\mu}{\rho}\right)_i \cdot p_i$$

p_i = mass contribution of *i*-th element

$$\sum_i p_i = 1$$

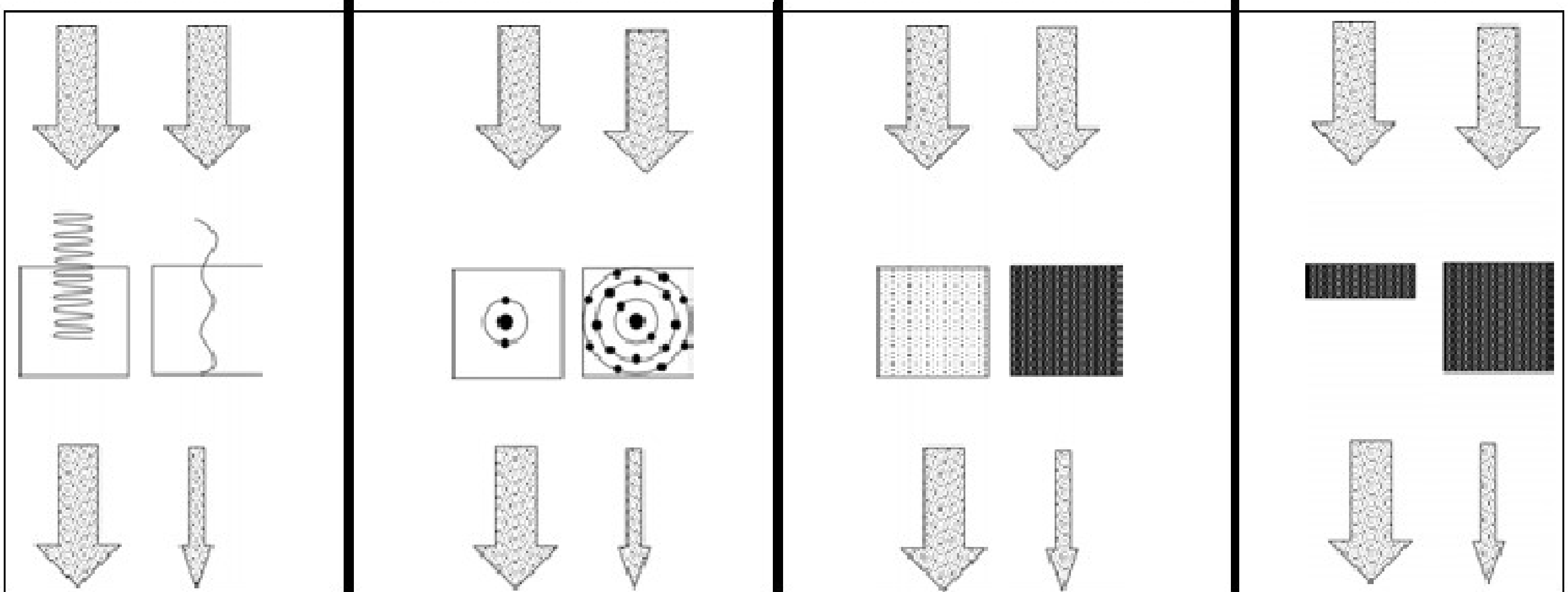
experimental set-up to measure attenuation coefficient μ



attenuation coefficient μ **absorption law**

schematic model of factors that contribute to attenuation

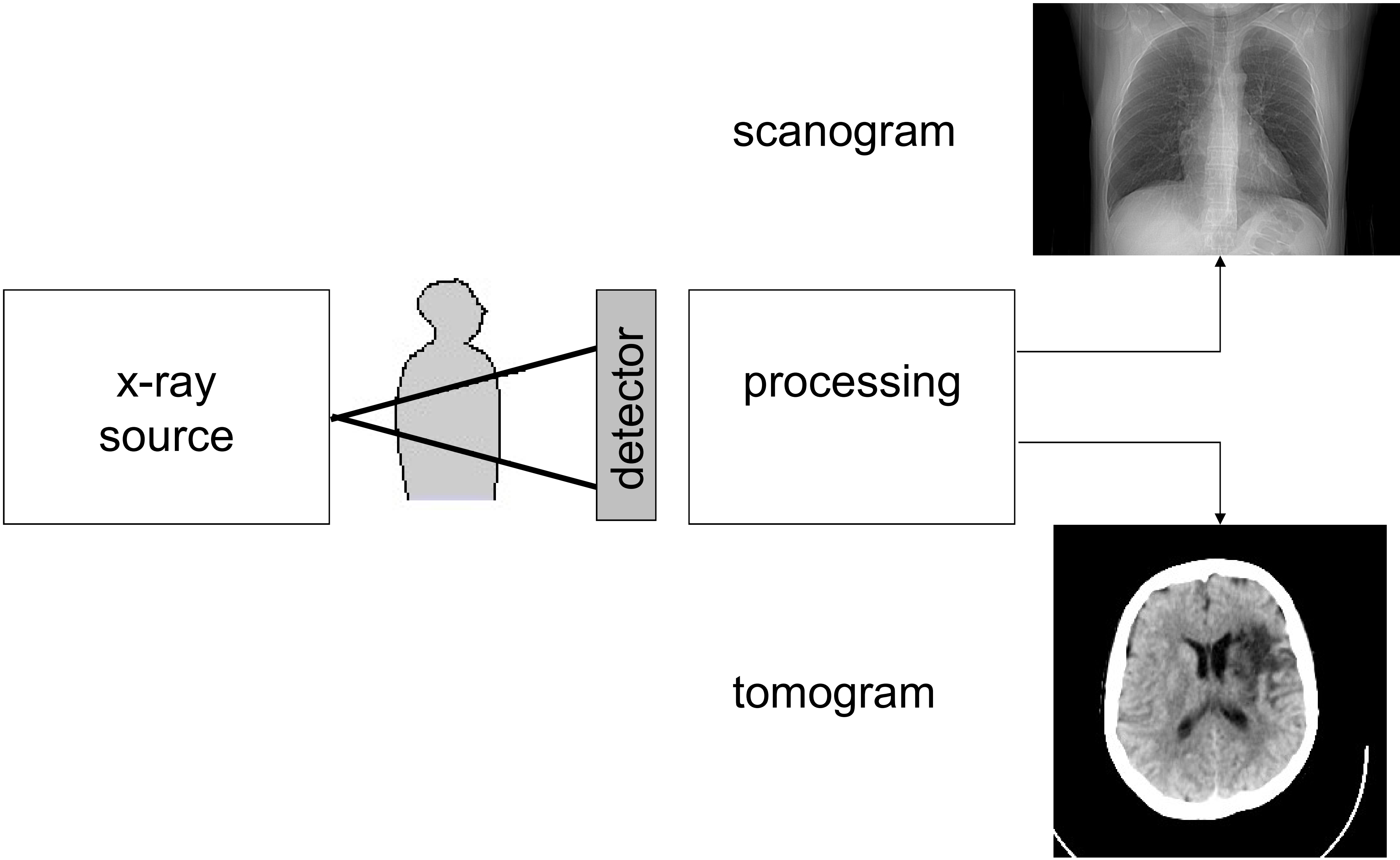
wave length λ	atomic number Z	density ρ (spec. wght)	thickness d



attenuation increases with λ^3	attenuation increases with Z^3	attenuation increases linearly with ρ	attenuation increases linearly with d

detectors

Imaging with x-rays



Imaging with x-rays

detectors

for projection radiography

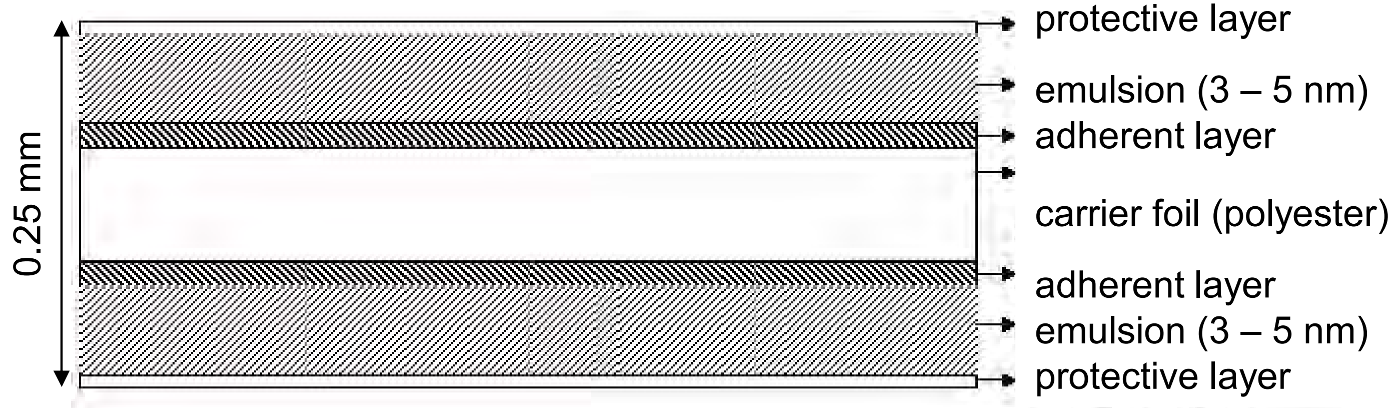
- x-ray film
- intensifying screen (amplifying foil)
- storage phosphor plate (digital luminescence radiography; DLR)
- selenium film (xeroradiography)
- CCD camera
- x-ray image amplifier

for computed tomography (CT)

- gas detectors
- solid-state detectors

x-ray film

double layer film



emulsion:
silver bromide crystals

$Br^- + h\upsilon \rightarrow Br + e^-$

released Ag^+ ions nucleate at exposed regions

processing: reduction of nucleated ions to silver ($Ag^+ + e^- \rightarrow Ag$)

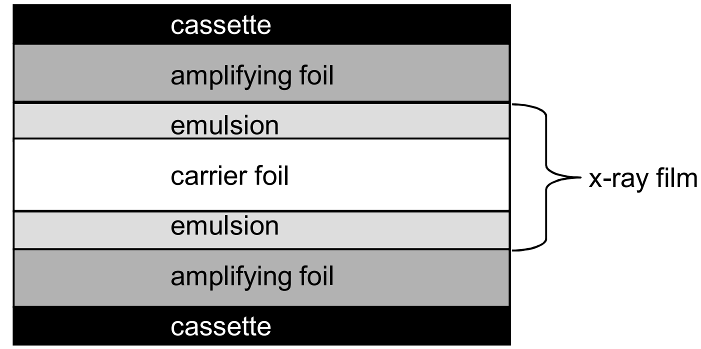
resolution: $\geq$ 0.025 mm

high μ - low blackening

blackening depends on attenuation coefficient μ, exposition time, dose

only 1% of quanta contribute to image !!

intensifying screen (amplifying foil)



improved usage of dose

convert x-rays into visible light

luminescence:
- γ-quant generates free e^-
- excited e^- relaxes to ground state by emission of light

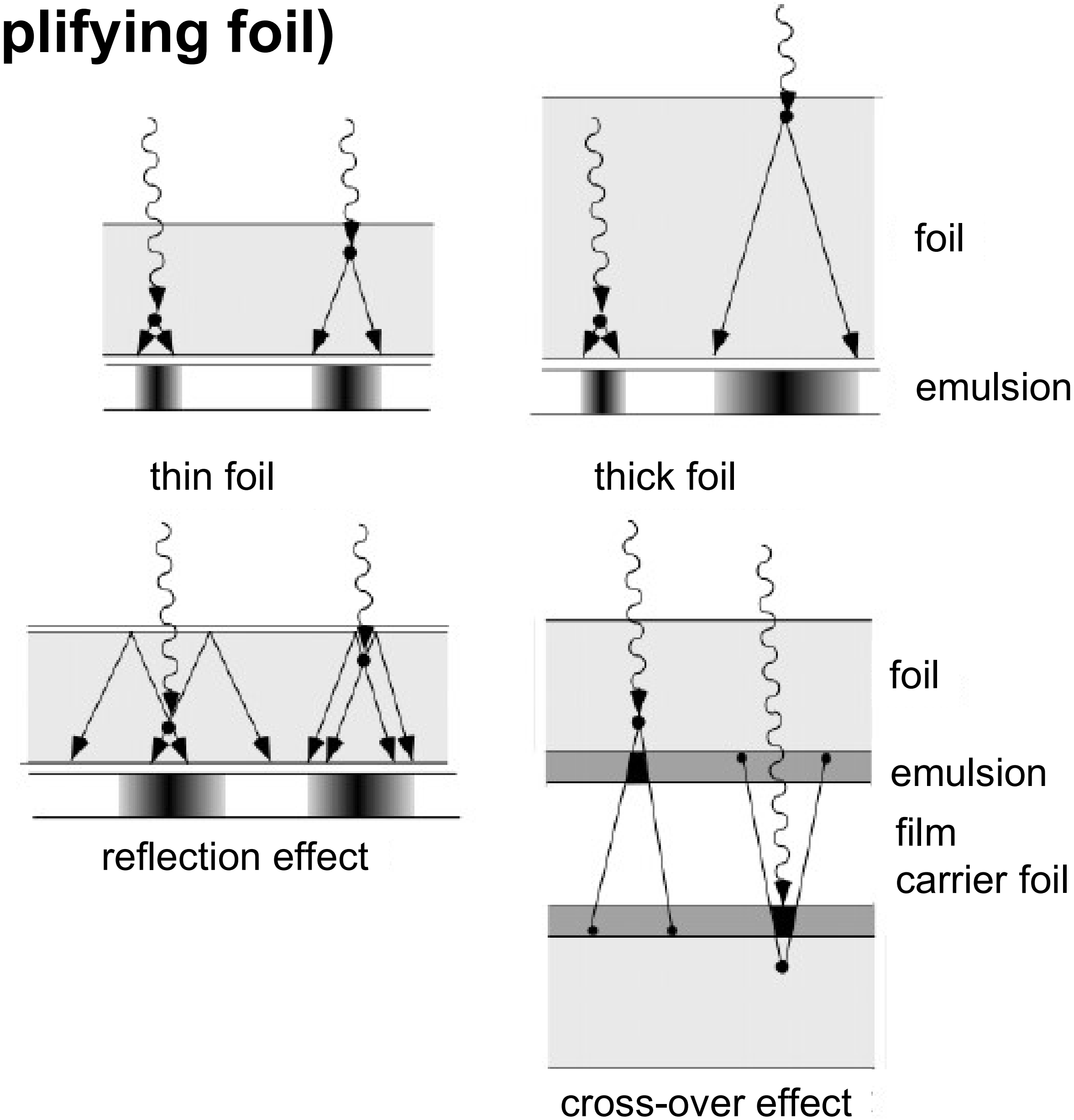
amplification factor V: dose without foil/dose with foil (typical: 10-20)

requirements: high absorption, high yield of quanta,
sufficient adjustment of spectrum to film sensitivity

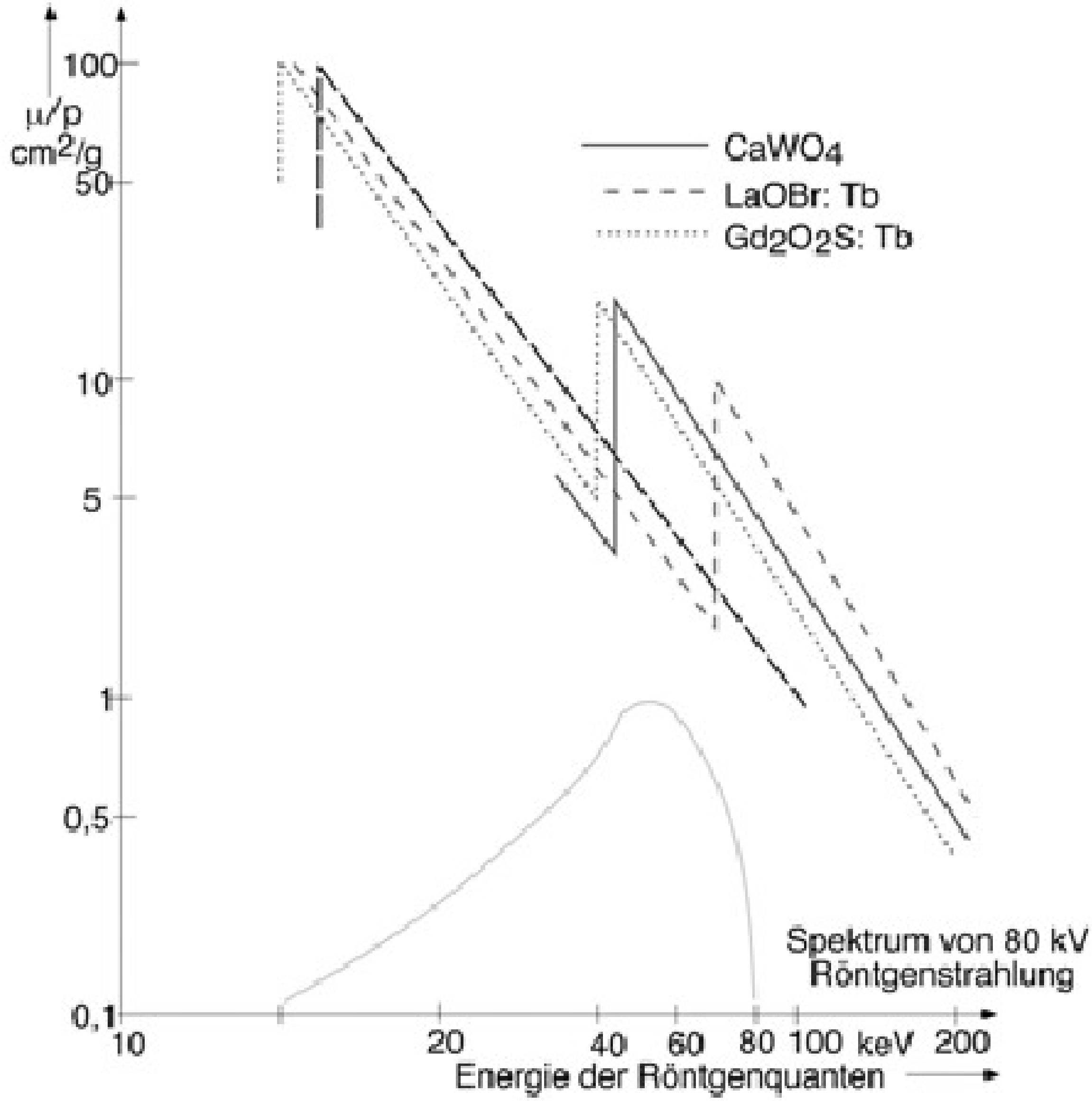
materials: calcium wolframite ($CaWO_4$) cross-section (cs): 4%
lanthanum oxybromide (doped w terbium) LaOBr:Tb cs 13 %
gadolinium sulfide (doped w terbium) Ga_2O_2S:Tb cs 19 %

intensifying screen (amplifying foil)

foil thickness
and
image fuzziness



mass absorption coefficient of different materials



storage phosphor plate (digital luminescence radiography)

same general principle as with amplifying foils (luminescence)

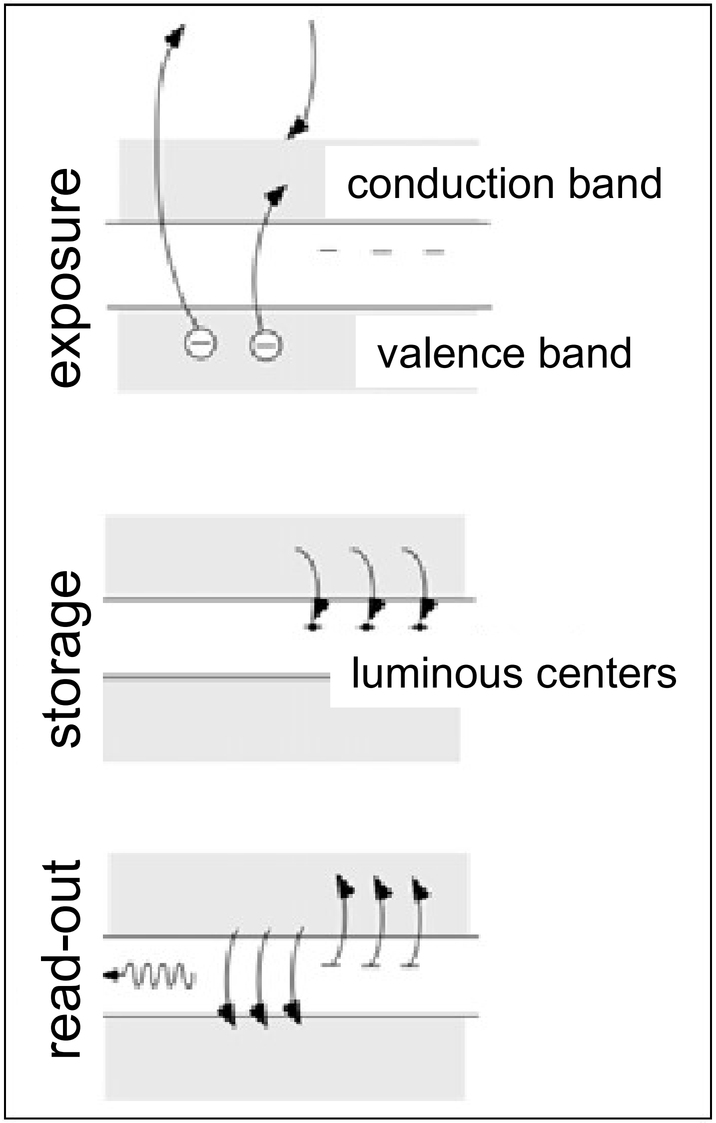
difference: excited e^- do not reach ground state (optically forbidden), generation of “traps“

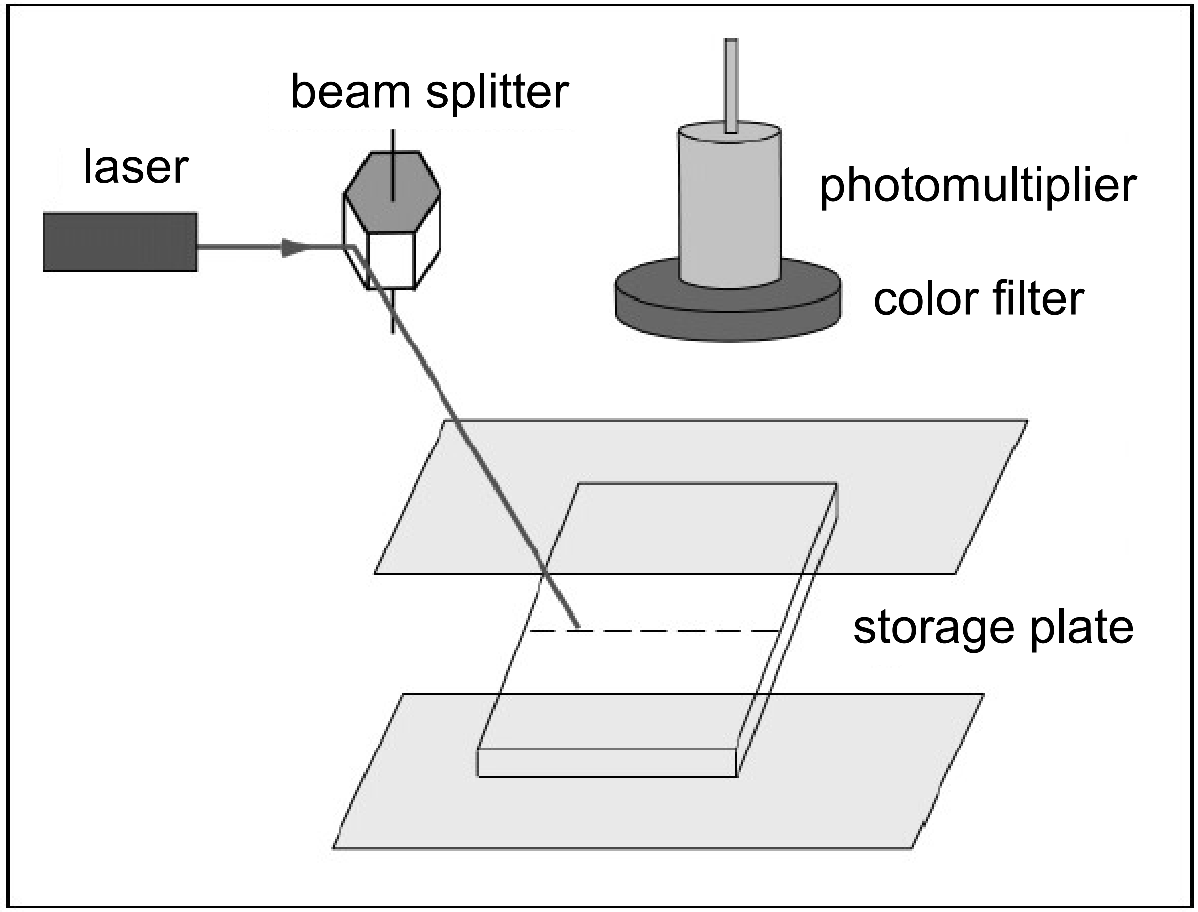
read-out: excitation of traps with laser, relaxation by emission of light (wavelength different to that of laser !!)

scanning of foil with laser scanner, color filter, photomultiplier, digitization

directly digitized image, high resolution, higher dynamic range compared to film

storage phosphor plate (digital luminescence radiography)





selenium film (xeroradiography)

basic principle: copy machine

selenium film mounted on a compact carrier is positively charged up (corona discharge)

γ-quanta release e^- from carrier, neutralization of charges in film

toner clings at positively charged areas only

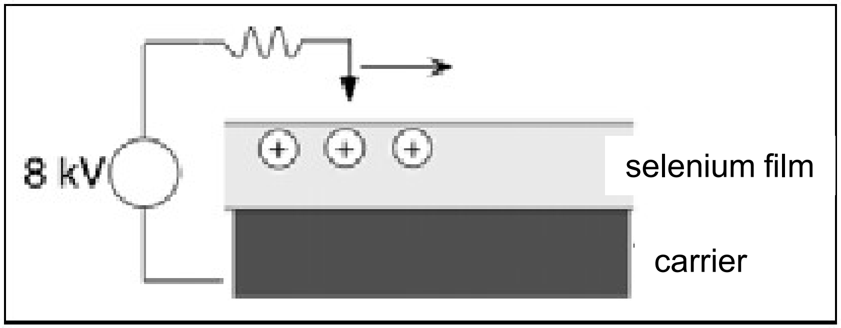
copy of pattern onto paper

for digital radiography: spatial sampling of “charge image” with comb-like capacitor and digitization

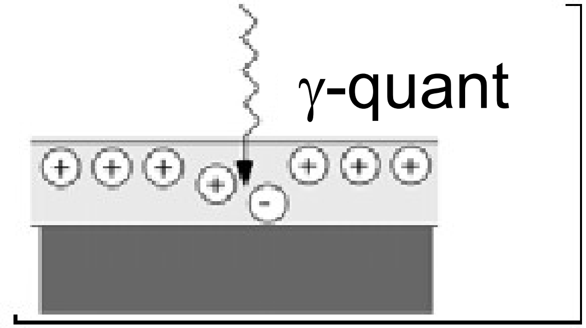
higher dynamic range compared to film

selenium film (xeroradiography)

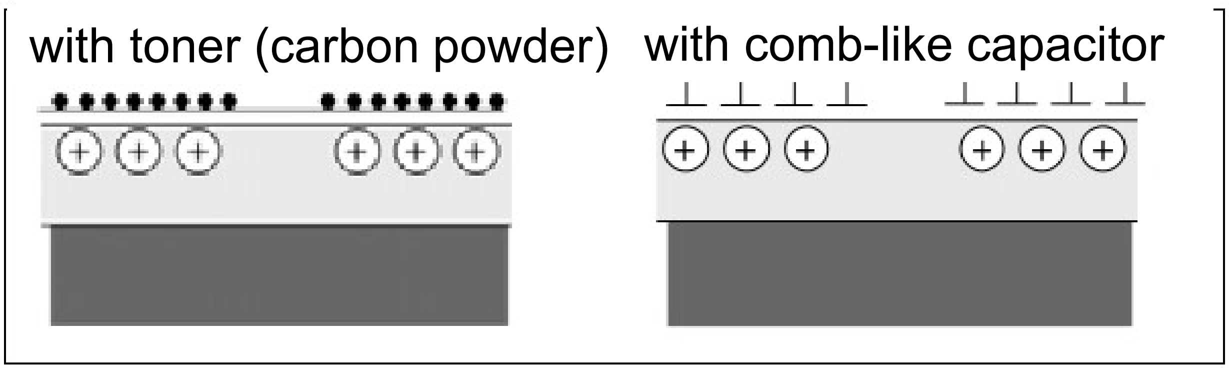
charge up



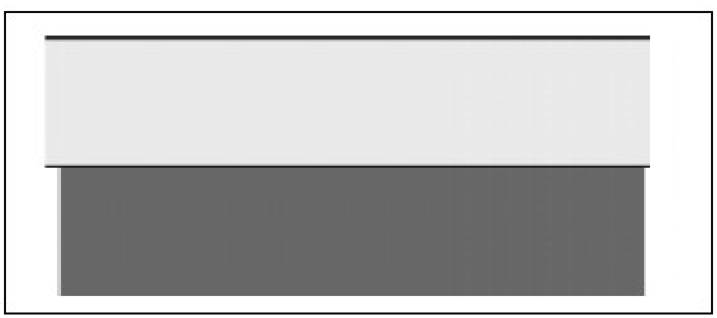
exposure



read-out



delete



CCD camera

most recent development for digital x-ray imaging

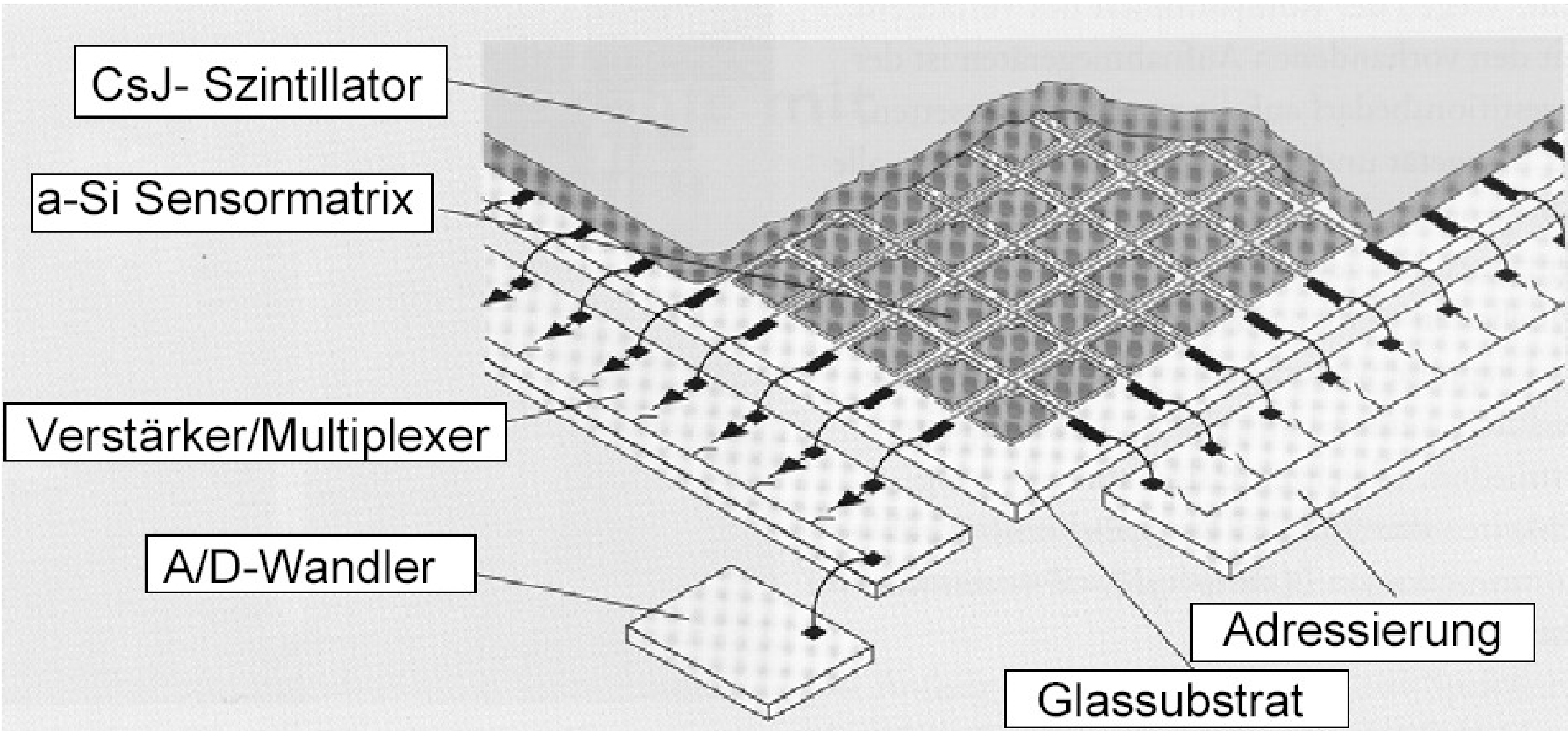
CCD-chips (charge-coupled devices) as camera for lines or planes

technological problems:
- decrease/increase of radiation with suitable optical elements
- requires large sensor areas

current systems: 1024 x 1024 pixel on an area of 20 cm x 20 cm (caesium iodide converter)

pros: flat, low weight, small
cons: expensive, requires higher dose (spatial sampling)

CCD camera

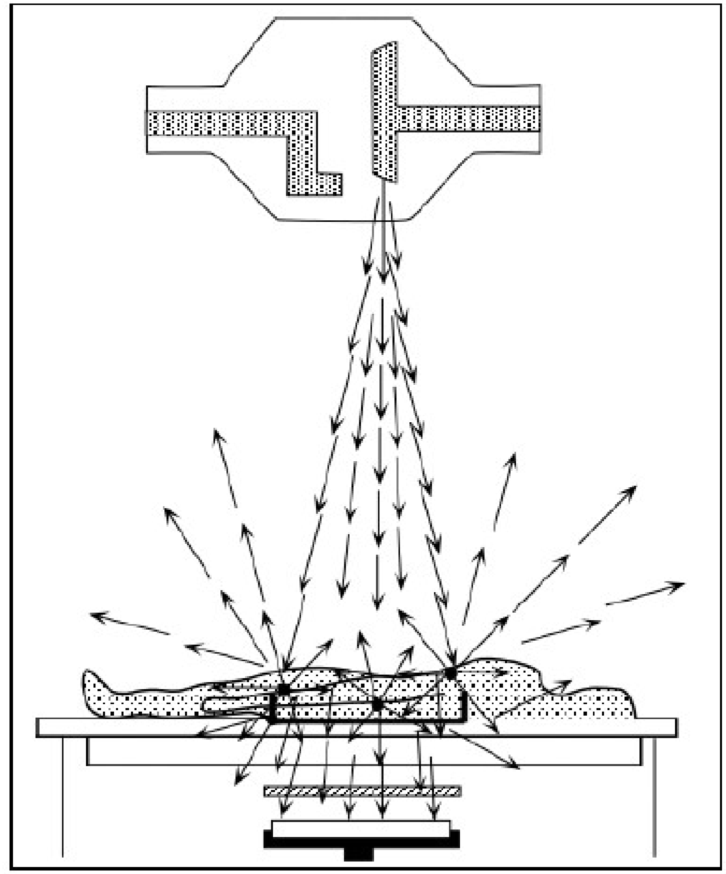

CsJ- Szintillator
a-Si Sensormatrix
Verstärker/Multiplexer
A/D-Wandler
Adressierung
Glassubstrat

raster (anti-scatter grid)

Compton effect → scattered x-rays

amount of scattered x-rays *P*:

$$P = \frac{J_s}{J_p + J_s}$$



where

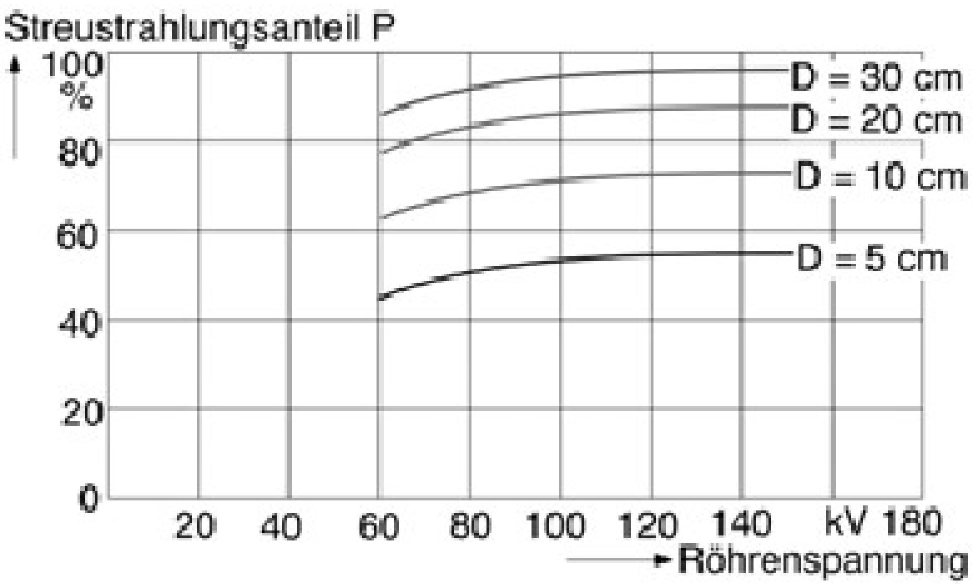
J_s = intensity of scatted x-rays in detector

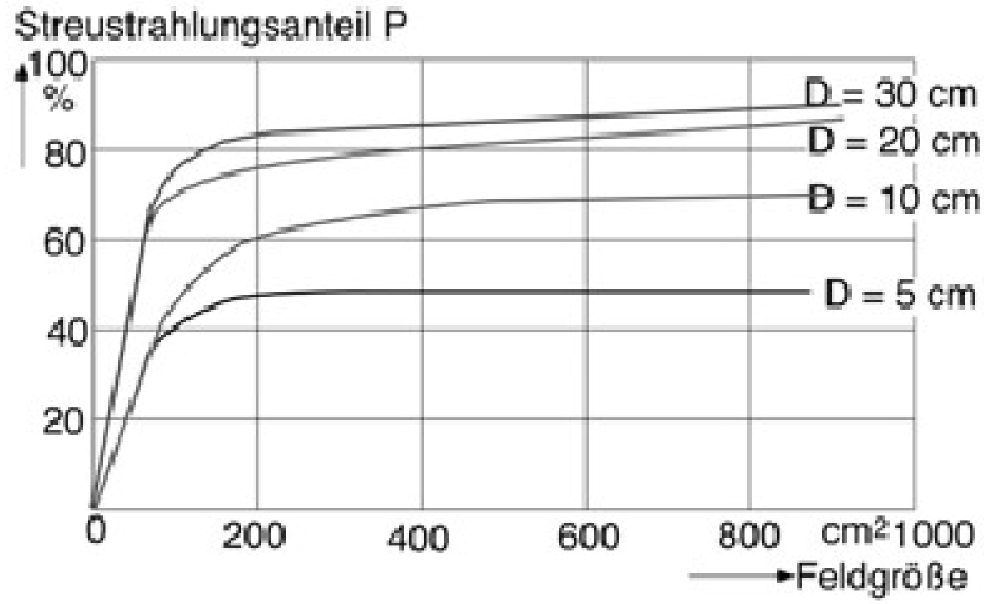
J_p = primary intensity: non-scattered x-rays in detector

scattered radiation leads to a diminished contrast !!

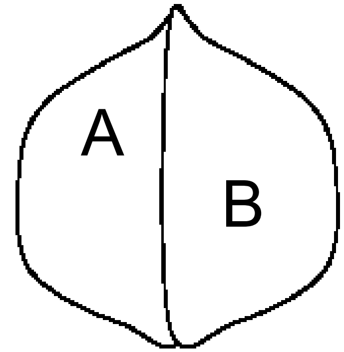
raster (anti-scatter grid)

amount of scattered x-rays depending on tube potential, object thickness D, and field of view





raster (anti-scatter grid)



contrast: $K = \frac{J_A - J_B}{J_A + J_B}$

J_A = x-ray intensity in area A

J_B = x-ray intensity in area B

for areas with small differences ΔJ in contrast, we have

$$K = \frac{\Delta J}{2J} \quad (J = \text{mean x-ray intensity in A and B})$$

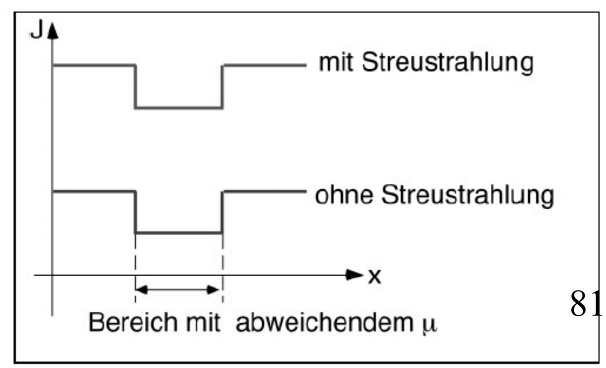
contrast without scattered radiation

$$K_o = \frac{\Delta J_p}{2J_p}$$

contrast with scattered radiation

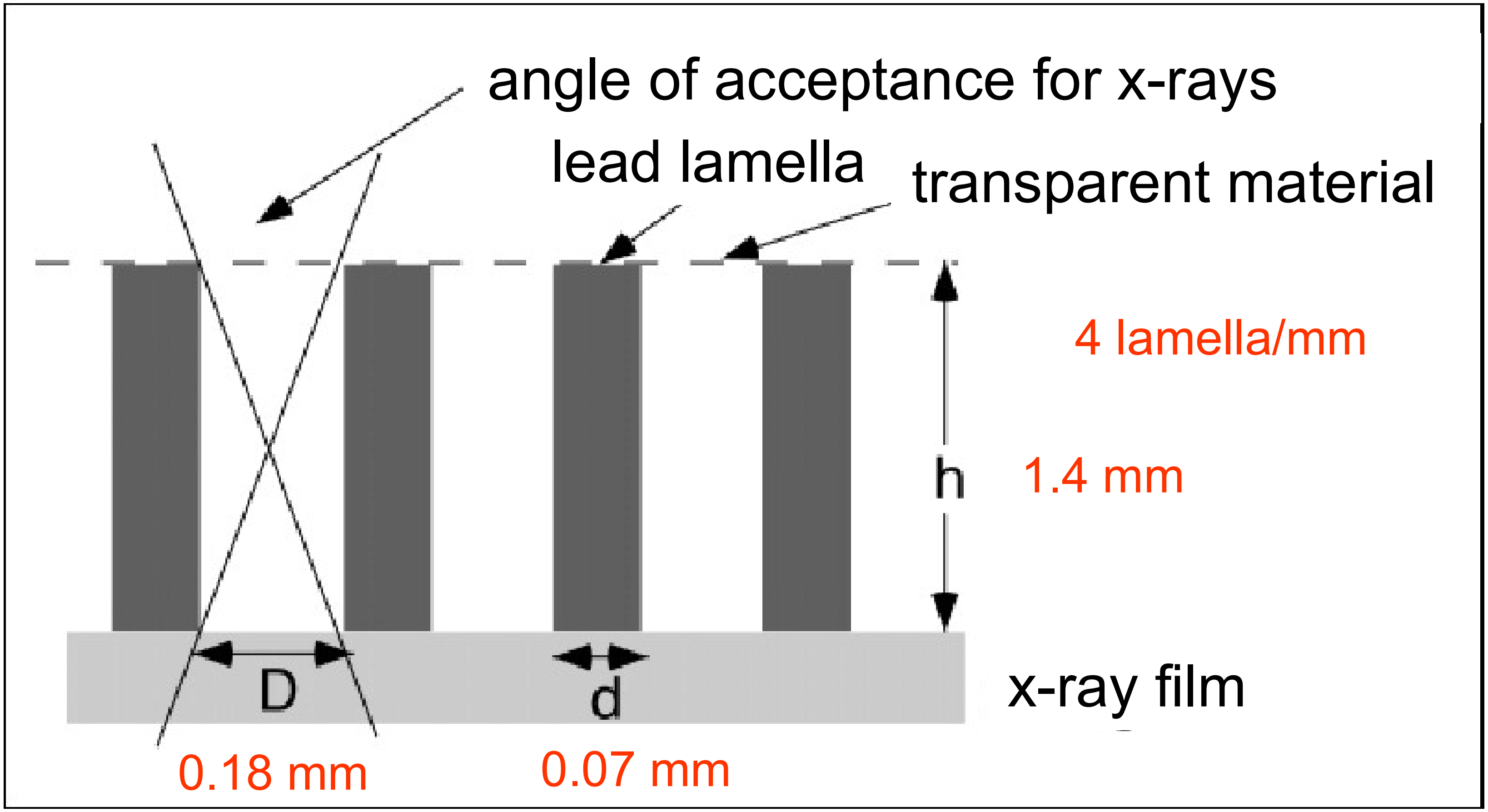
$$K_s = \frac{\Delta J_p}{2(J_p + J_S)} = K_o \cdot \frac{J_p}{J_p + J_S} = K_o \cdot \frac{1}{1 + \frac{J_S}{J_p}}$$

scattered radiation increases total intensity, but does not lead to smearing !!!

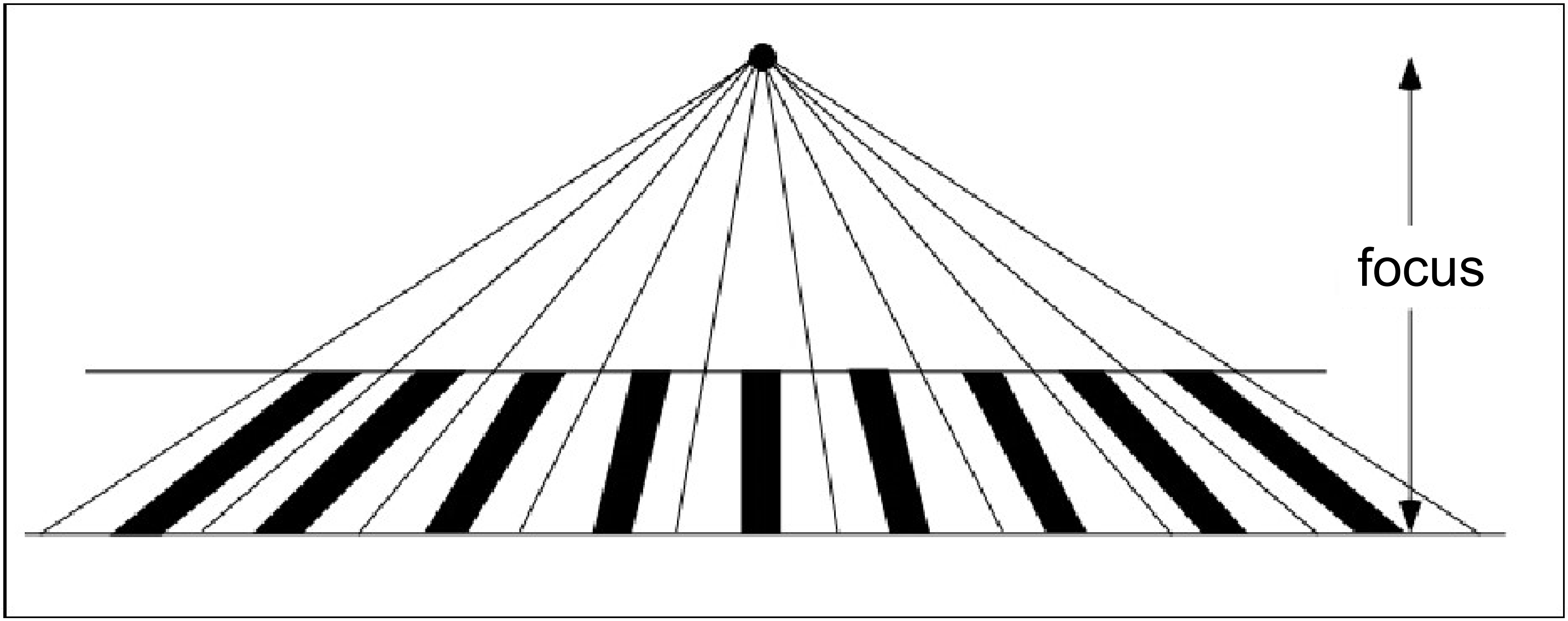


raster (anti-scatter grid)

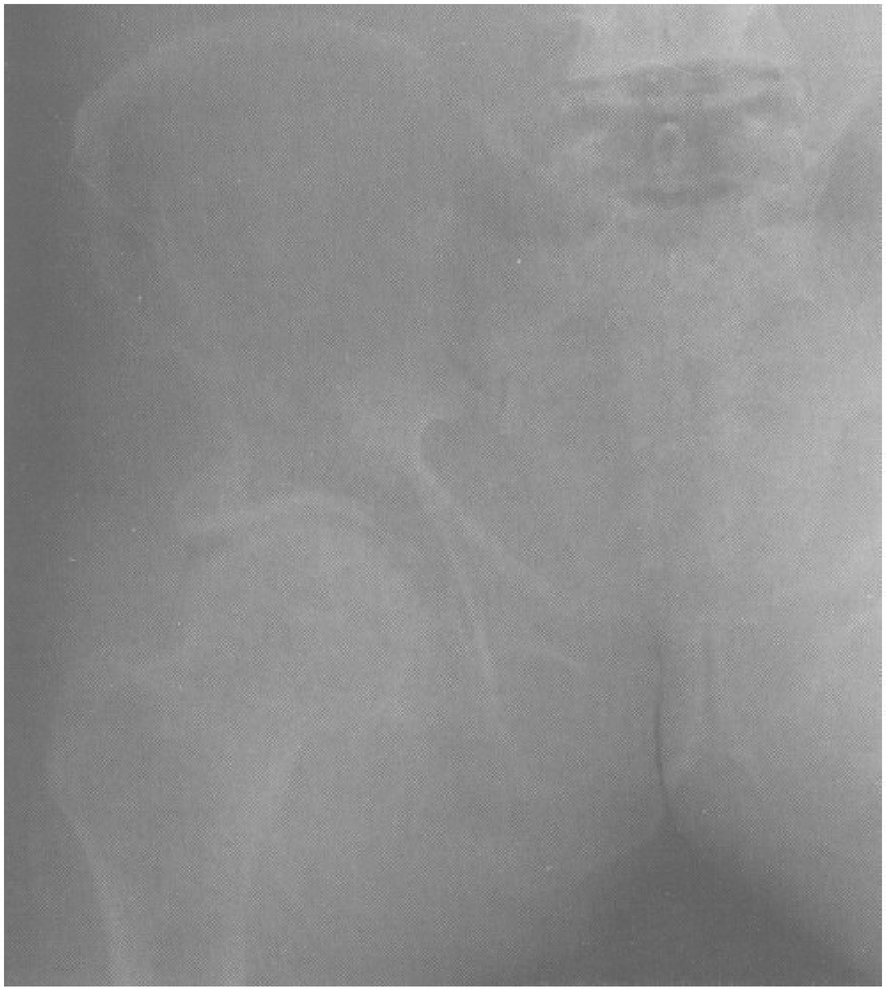
- amount of scattered radiation can be reduced with **raster**
- plates : alternating arrangement of lead lamella and material transparent to x-rays
- raster applied directly to x-ray film of amplifying foil



focusing line raster



focusing line raster



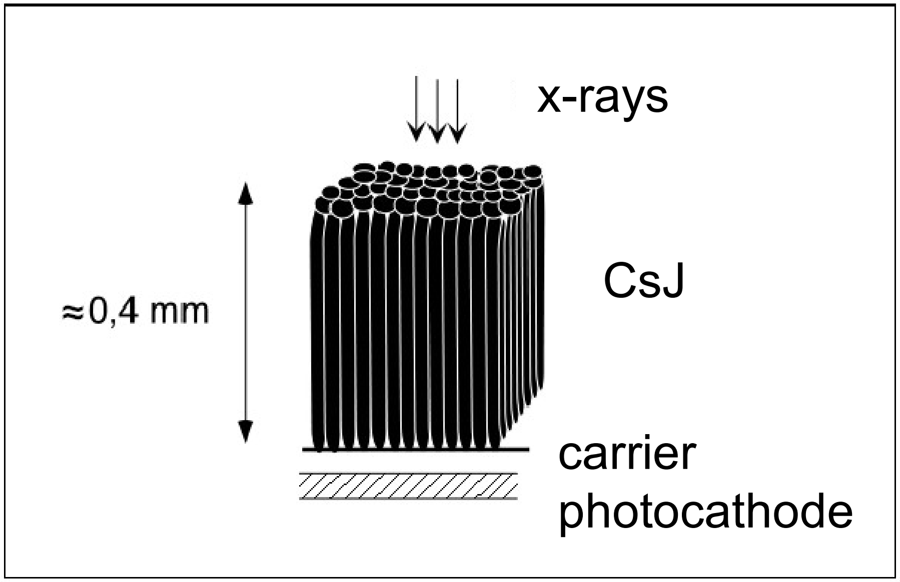
high amount of scattered radiation,
75 kV, no raster

low amount of scattered radiation,
75 kV, with raster

x-ray image amplifier

aim: visualization of dynamic processes during x-ray imaging
in the past: watch fluorescent screen, high radiation exposure
today: video-based image chain (optics, camera, monitor)

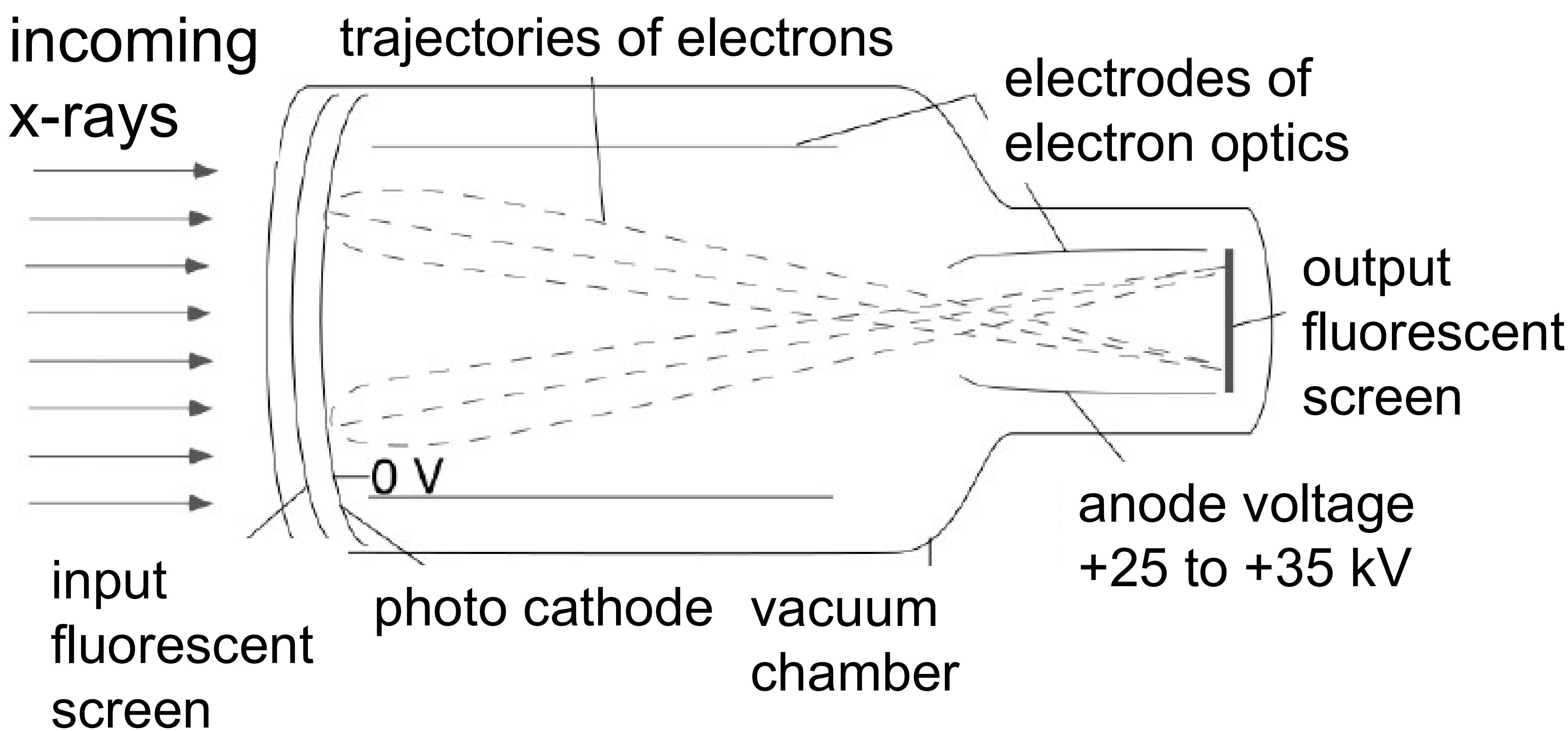
principle:
convert x-rays into visible light
photo effect; caesium iodide screen



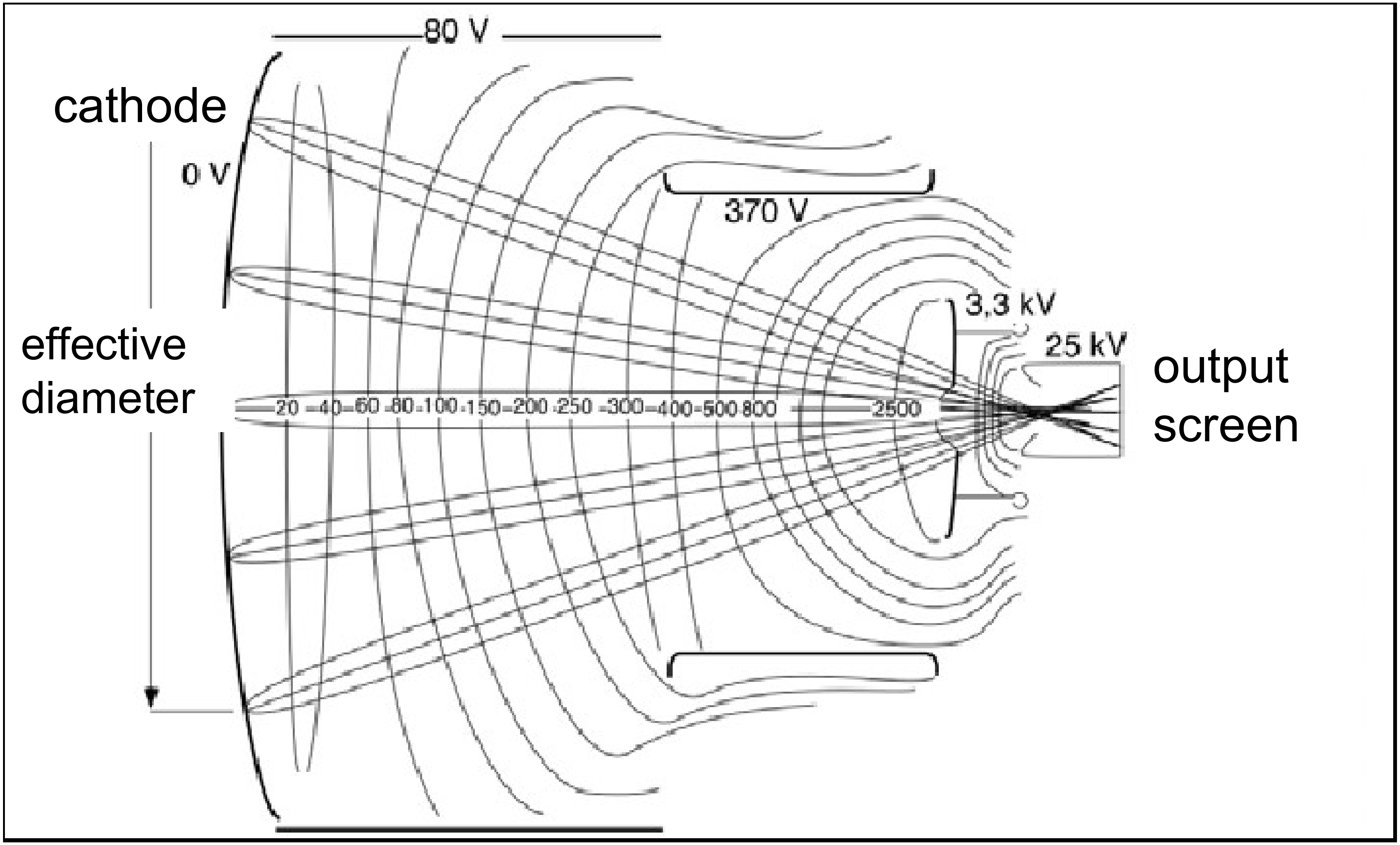
amplification with electron optics

recording of image on ZnCdS:Ag screen with video camera

x-ray image amplifier



distribution of potential in x-ray image amplifier



x-ray image amplifier

example:

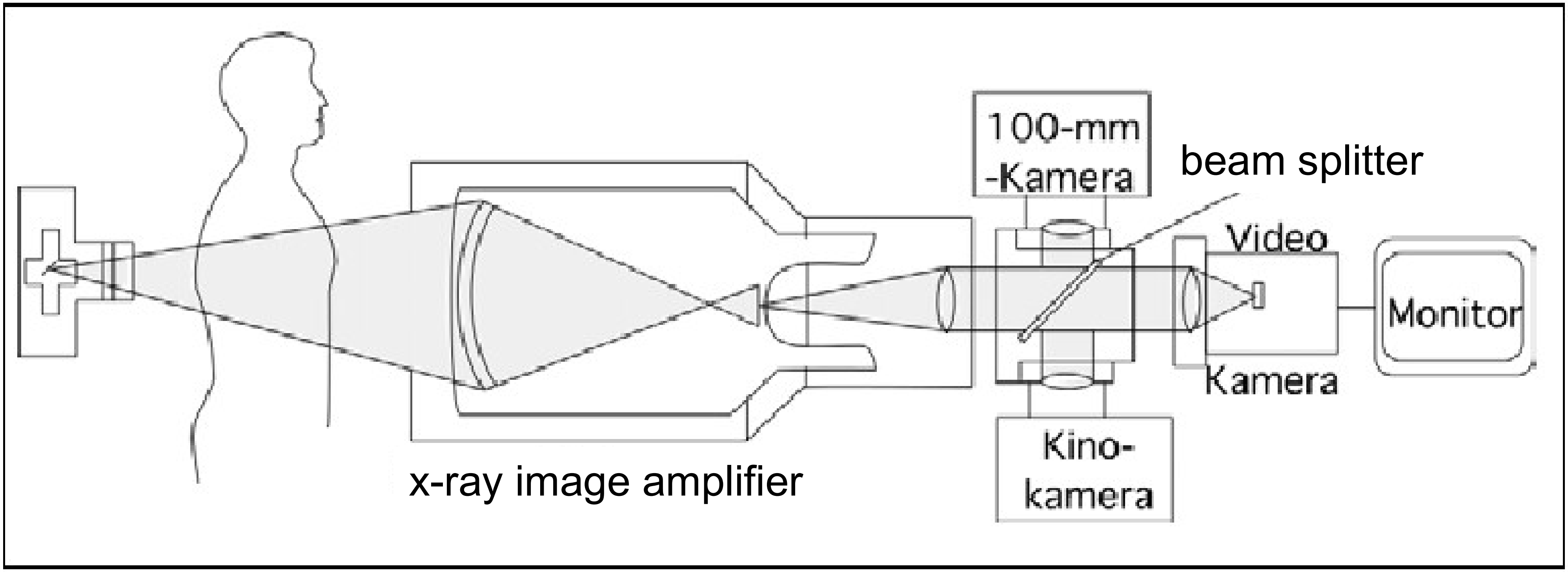
dose at fluorescent input screen: 0.2 µGy/s
(equals to $5 \cdot 10^5$ γ-quanta per cm and s)

absorption in input screen: ~ 60 %
- per γ-quant approx. 1000 photons on fluorescent input screen
- per electron approx. 1000 photons
- per photon at input screen approx. 100 photons at output screen

area of output screen about 10x smaller than input screen
⇒ increase of light density by x100

⇒ **total amplification: 10^4**

x-ray image amplifier



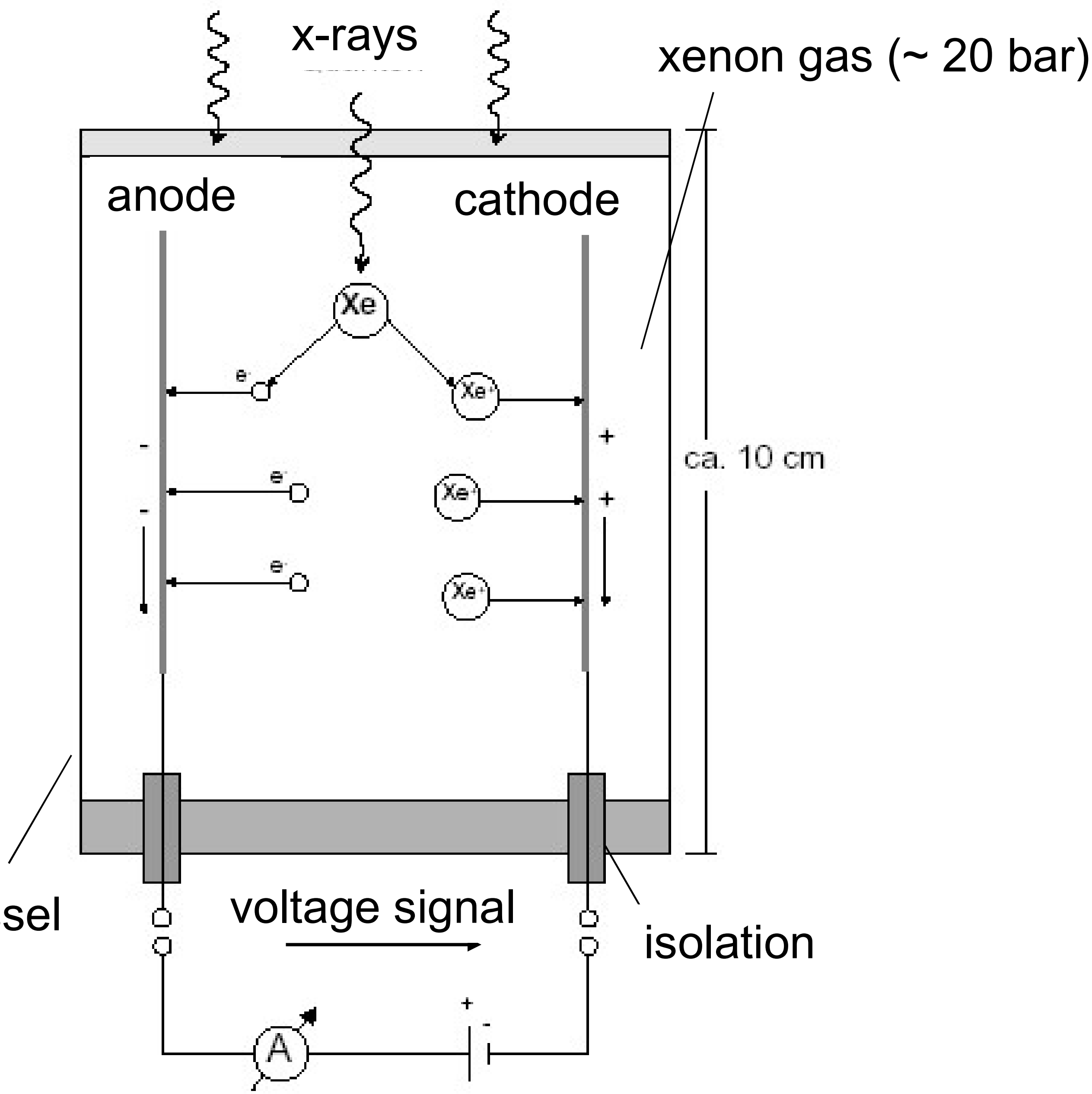
gas detectors

principle: ionization chamber

- x-rays ionize of xenon gas (high pressure chamber)
- walls of chamber = capacitor plates (high voltage)
- ionization generates charged particles
- charge separation
- output voltage directly prop. to intensity of x-rays

pros
- fast decay times (short acquisition times)
- insensitive to fluids and temperature

gas detectors



solid-state detectors

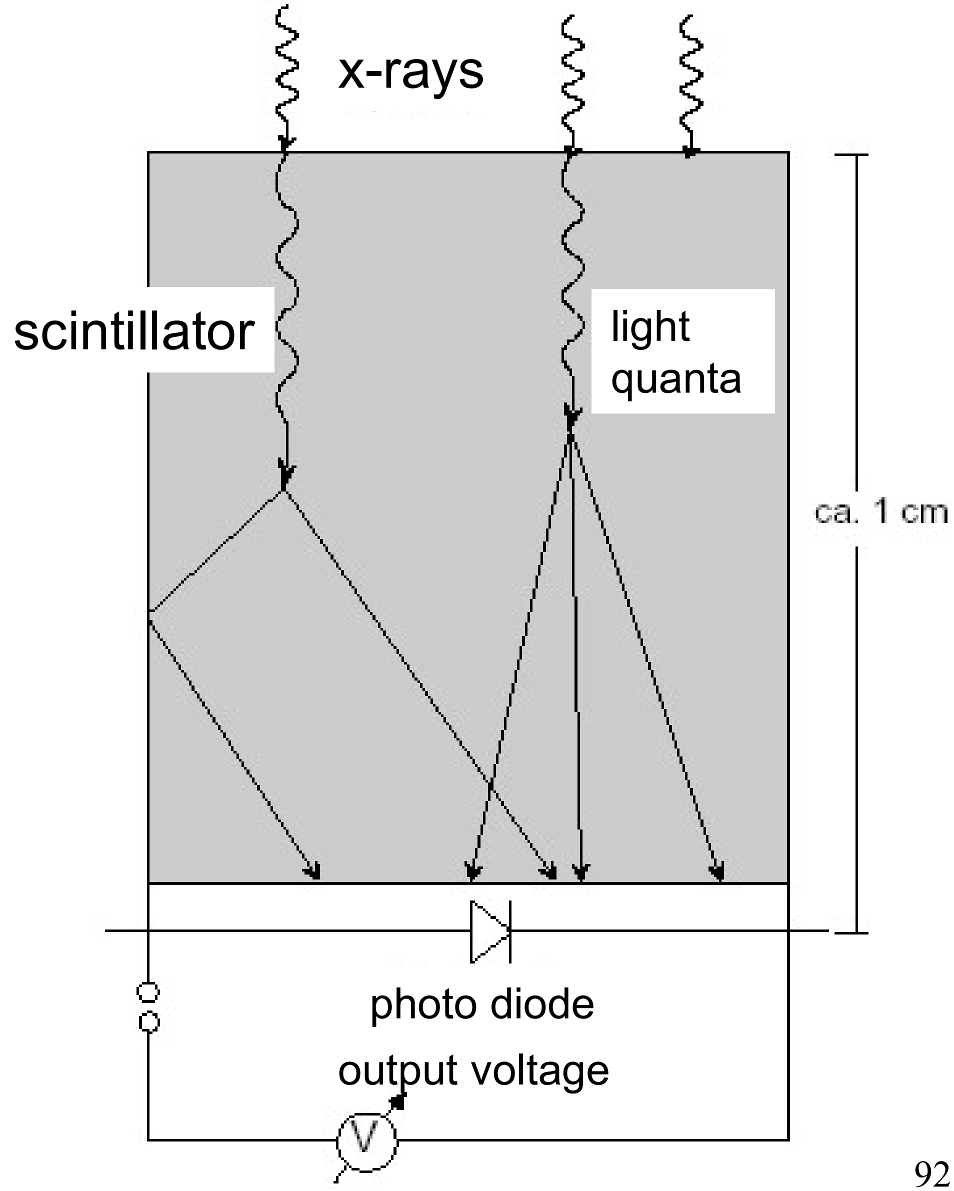
principle: scintillation

- x-rays excite e^- in crystal

- relaxation with emission of photons

- conversion to voltage changes with photo diode

- voltage directly prop. to energy of x-rays

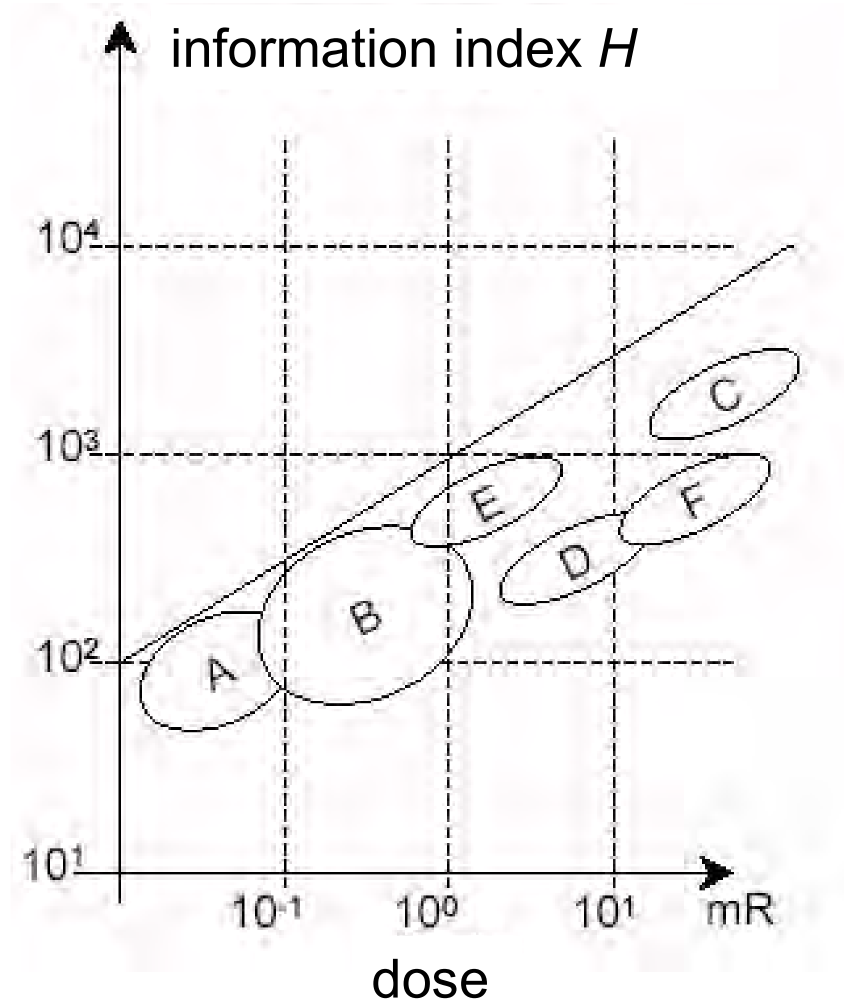


comparing different detectors

information index *H*:

$H = c \cdot r$

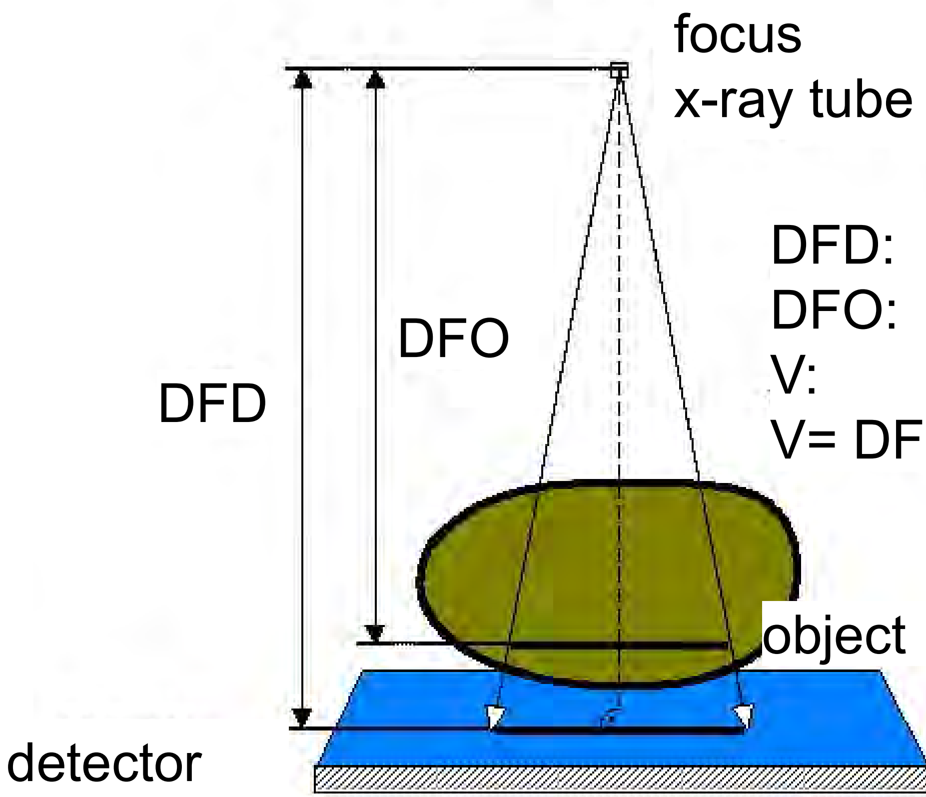
c = contrast sensitivity
r = resolution



A: x-ray image amplifier
B. intensifying screen (amplifying foil)
C: x-ray film
D: xeroradiography
E: electroradiography with Xe chamber
F: CT

impact of geometry **projection laws**

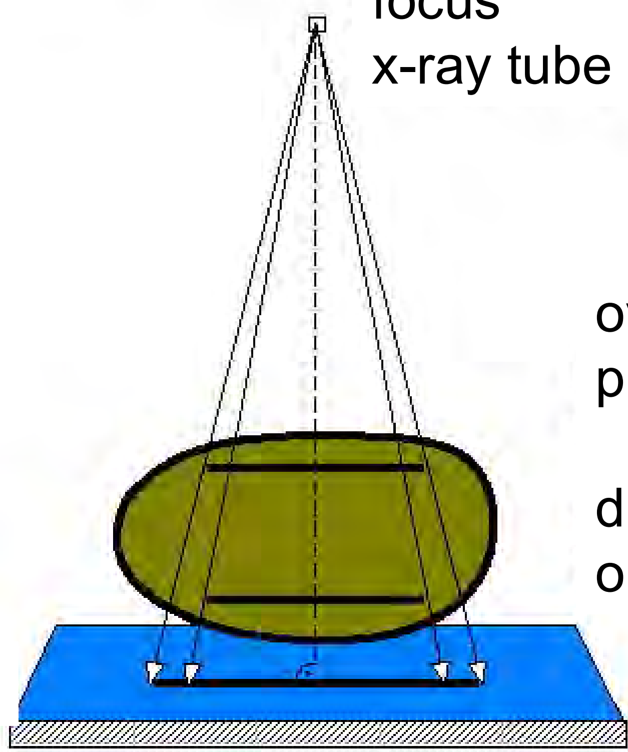
central projection



DFD: distance focus-detector
DFO: distance focus-object
V: amplification factor
V= DFD/DFO

impact of geometry **projection laws**

inverted field of view

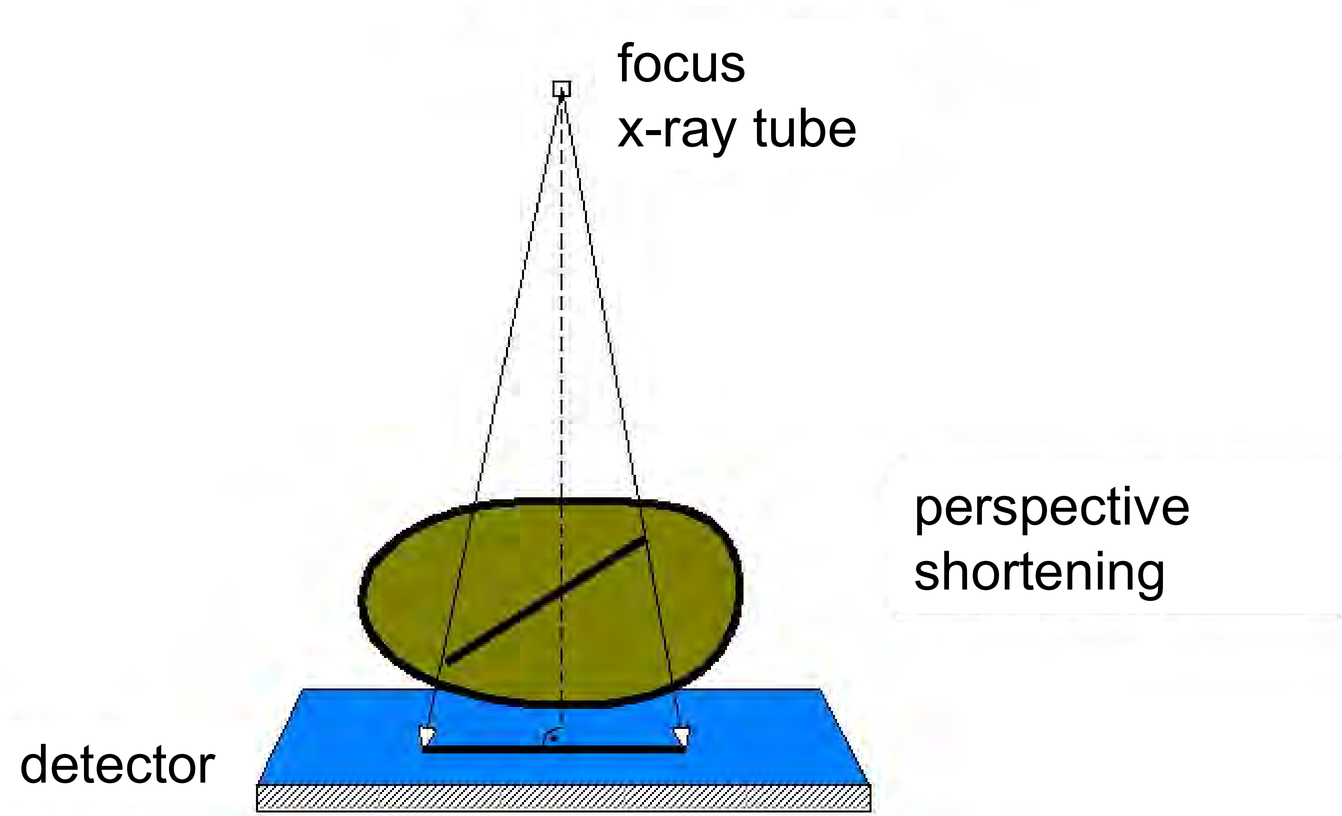


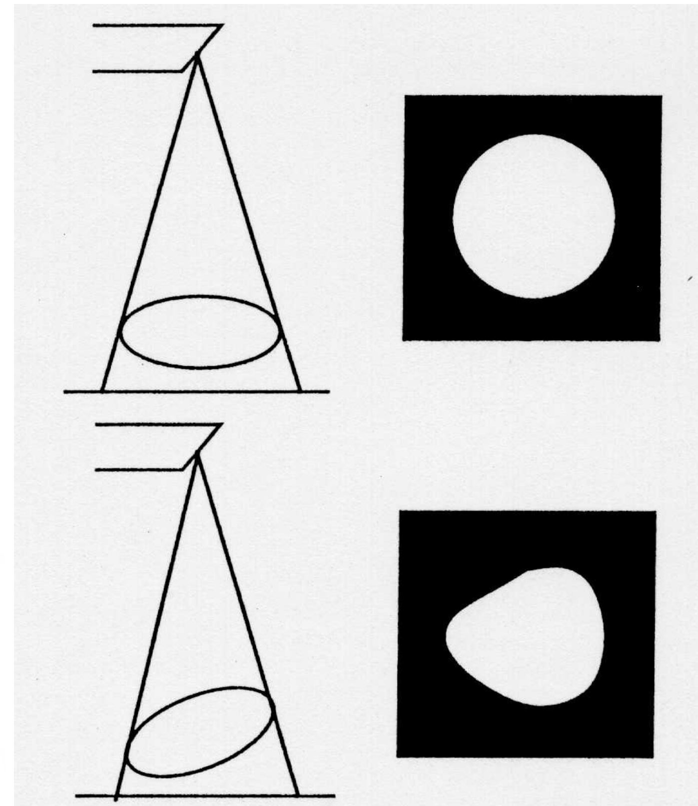
overlay of objects in image plane

different amplification of objects of equal length

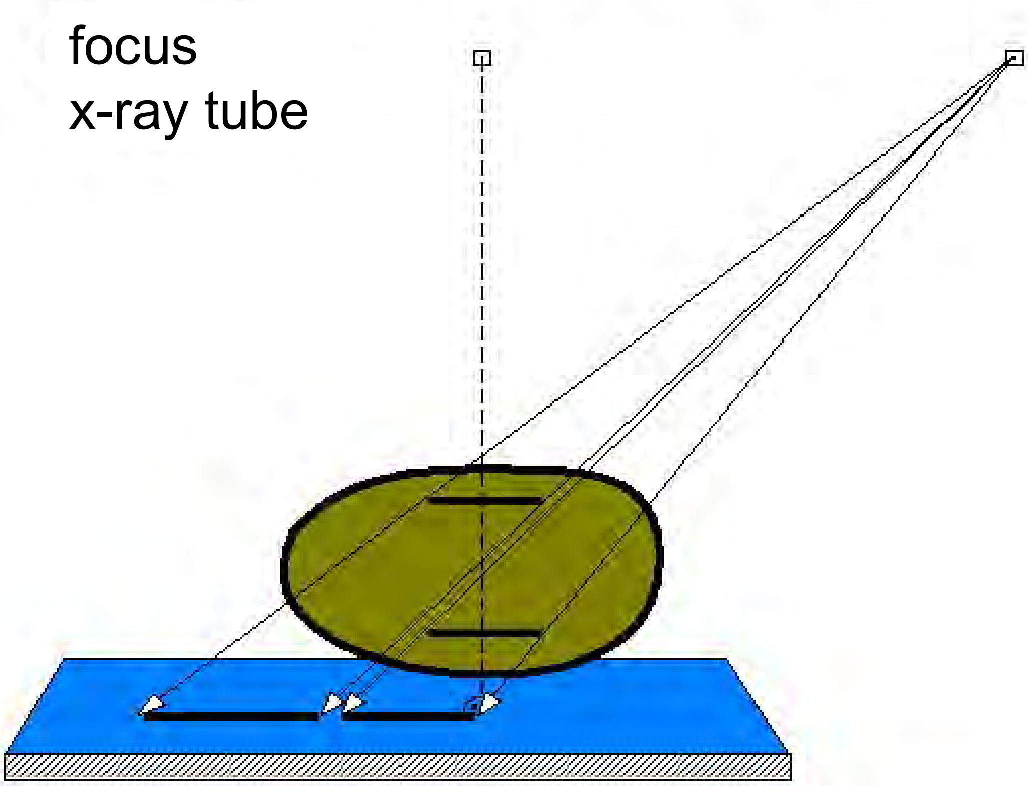
impact of geometry **projection laws**

distortions





disentangling of superposed objects



impact of geometry **projection laws**

inverse square law

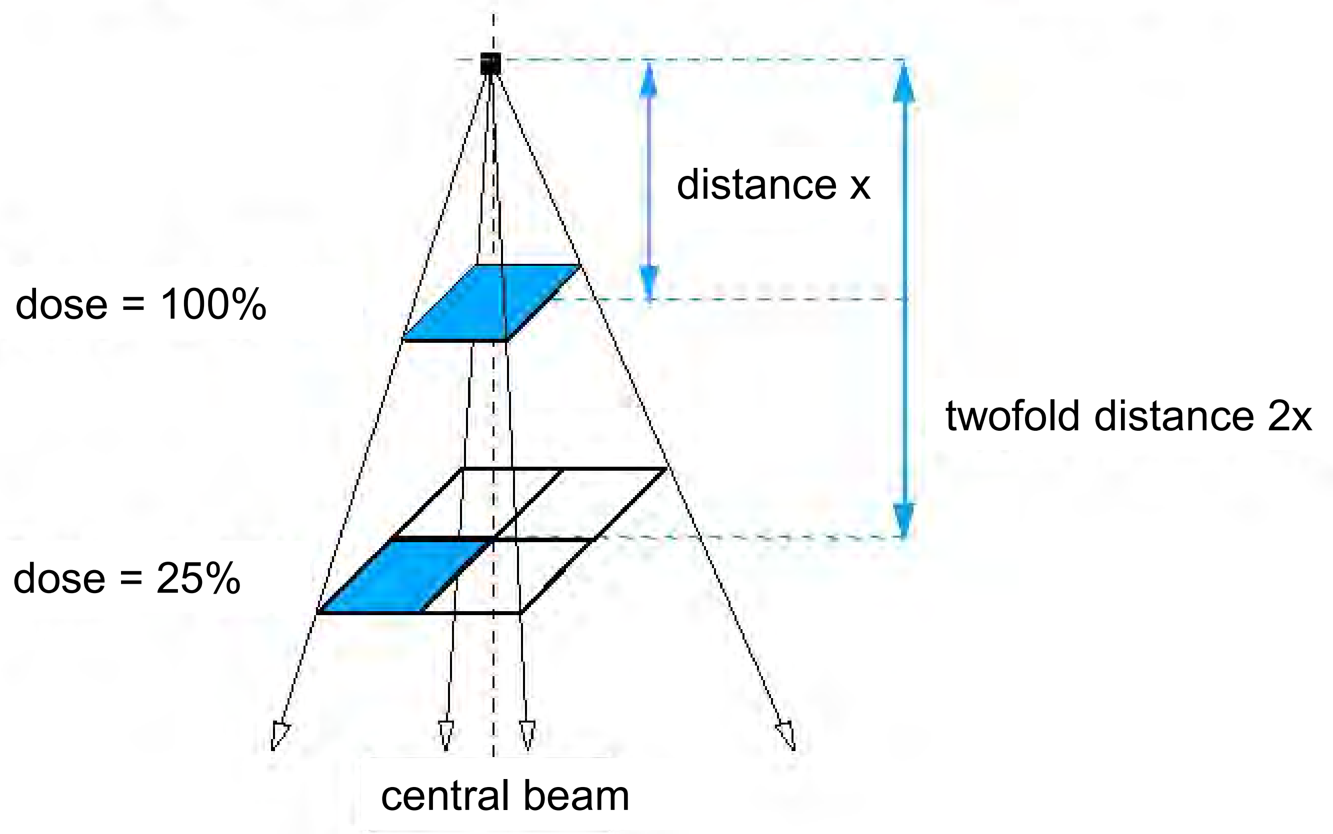


image quality

Imaging with x-rays

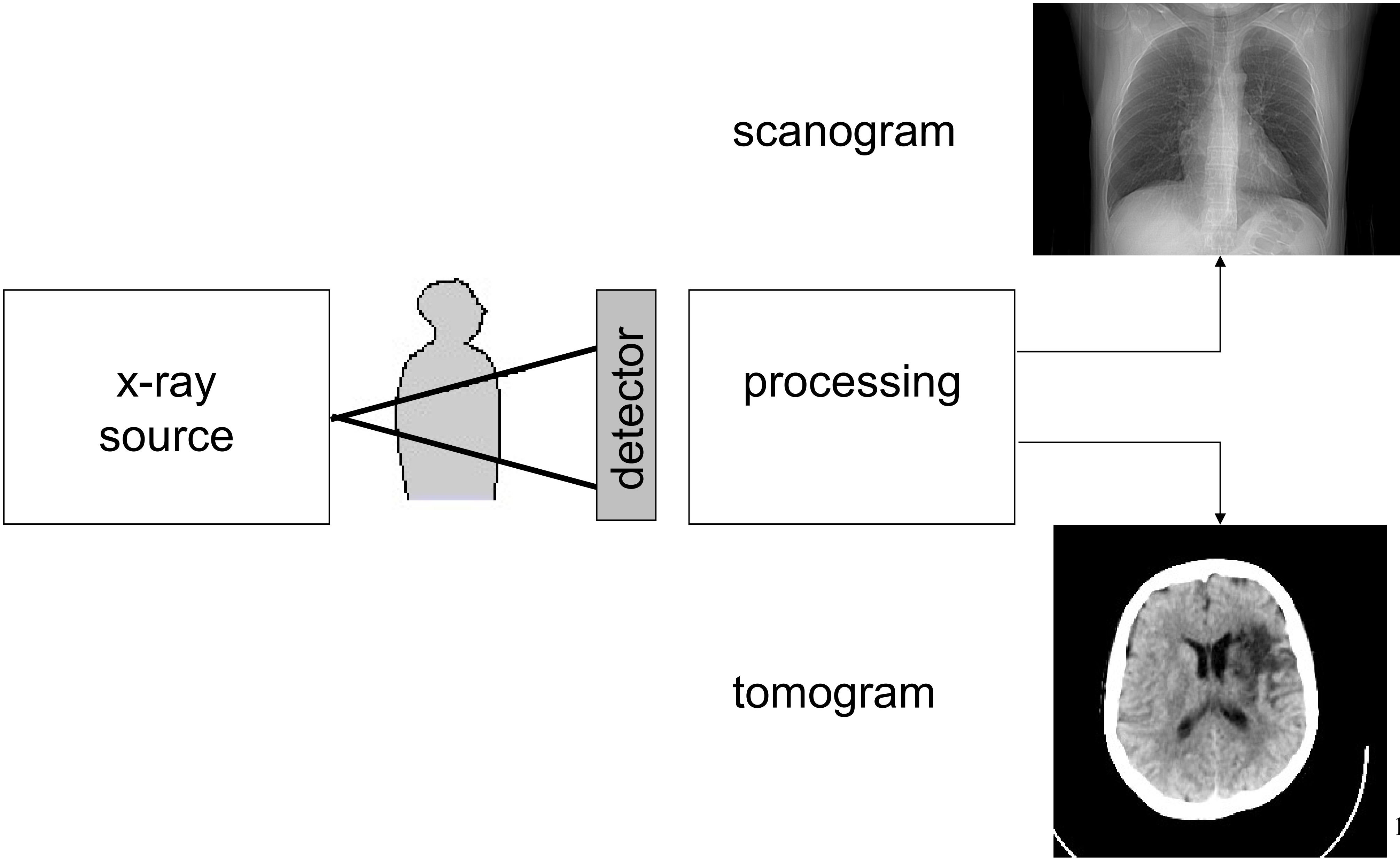


image quality of x-ray image amplifier **influencing factors**

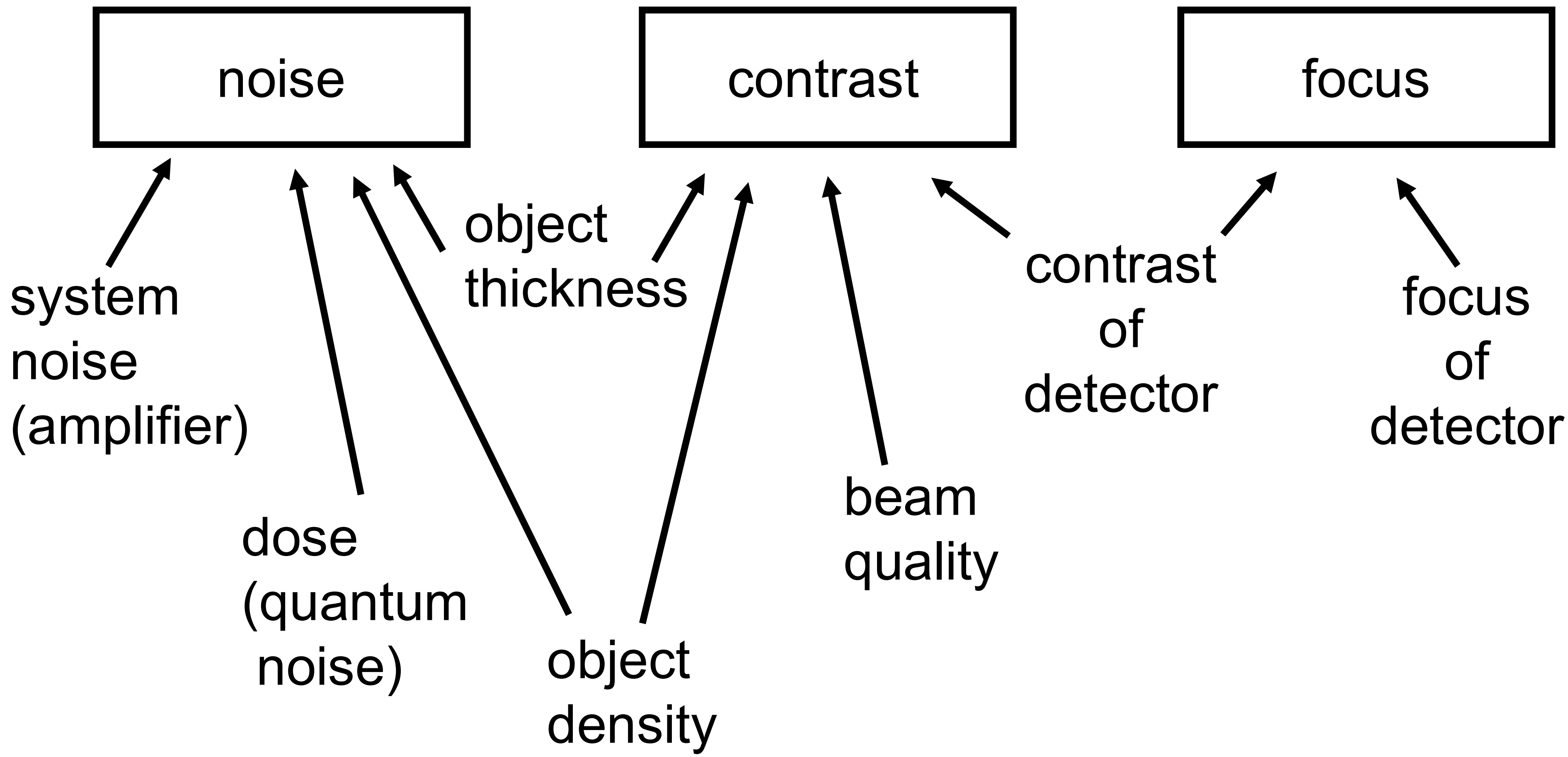
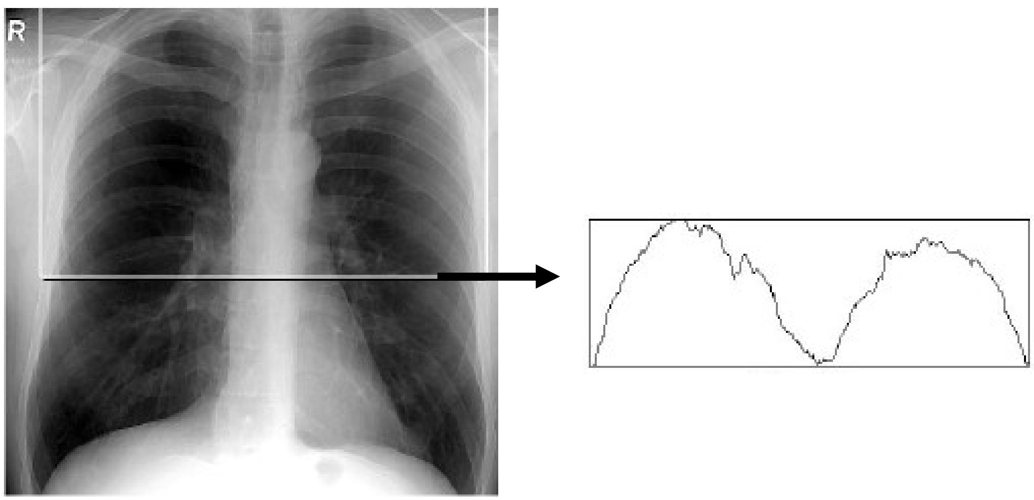


image quality of x-ray image amplifier

intensity profile



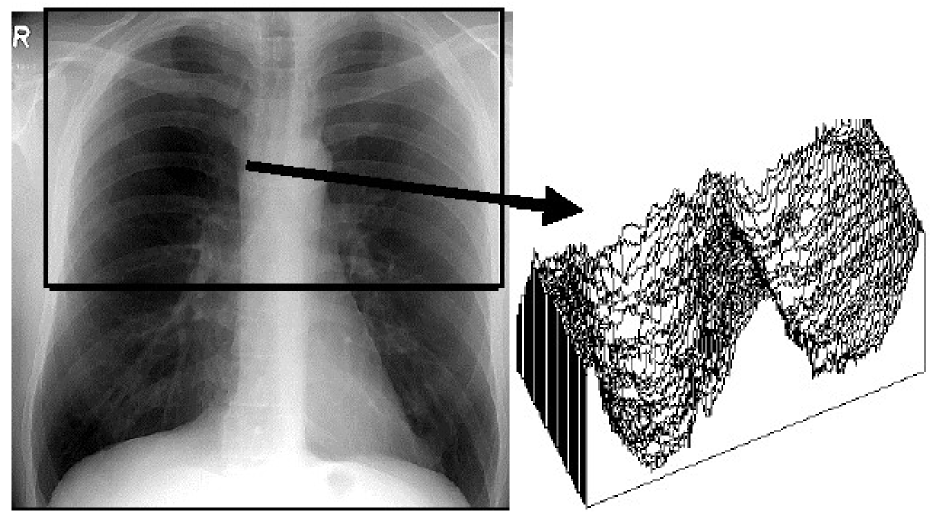
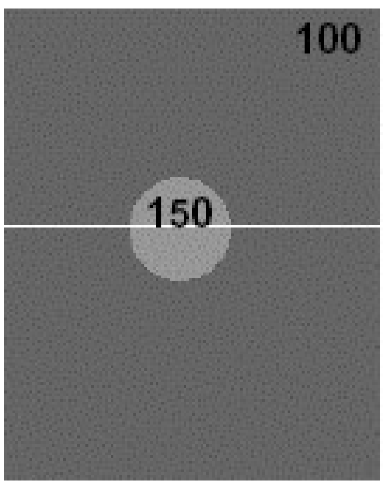


image quality of x-ray image amplifier

$$\text{contrast} = \frac{\text{object intensity} - \text{background intensity}}{\text{background intensity}}$$

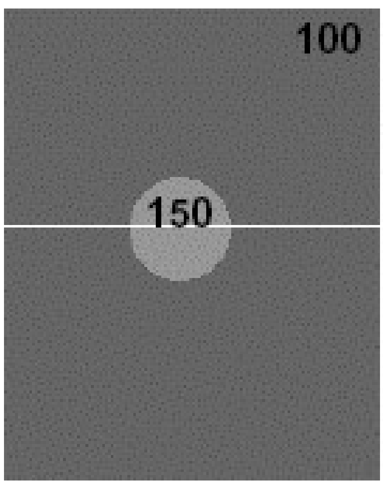


contrast = (150-100)/100 = 0,5

image quality of x-ray image amplifier

$$\text{modulation} = \frac{\text{object intensity - background intensity}}{\text{background intensity + object intensity}}$$

$$\text{modulation} \in [0,1]$$



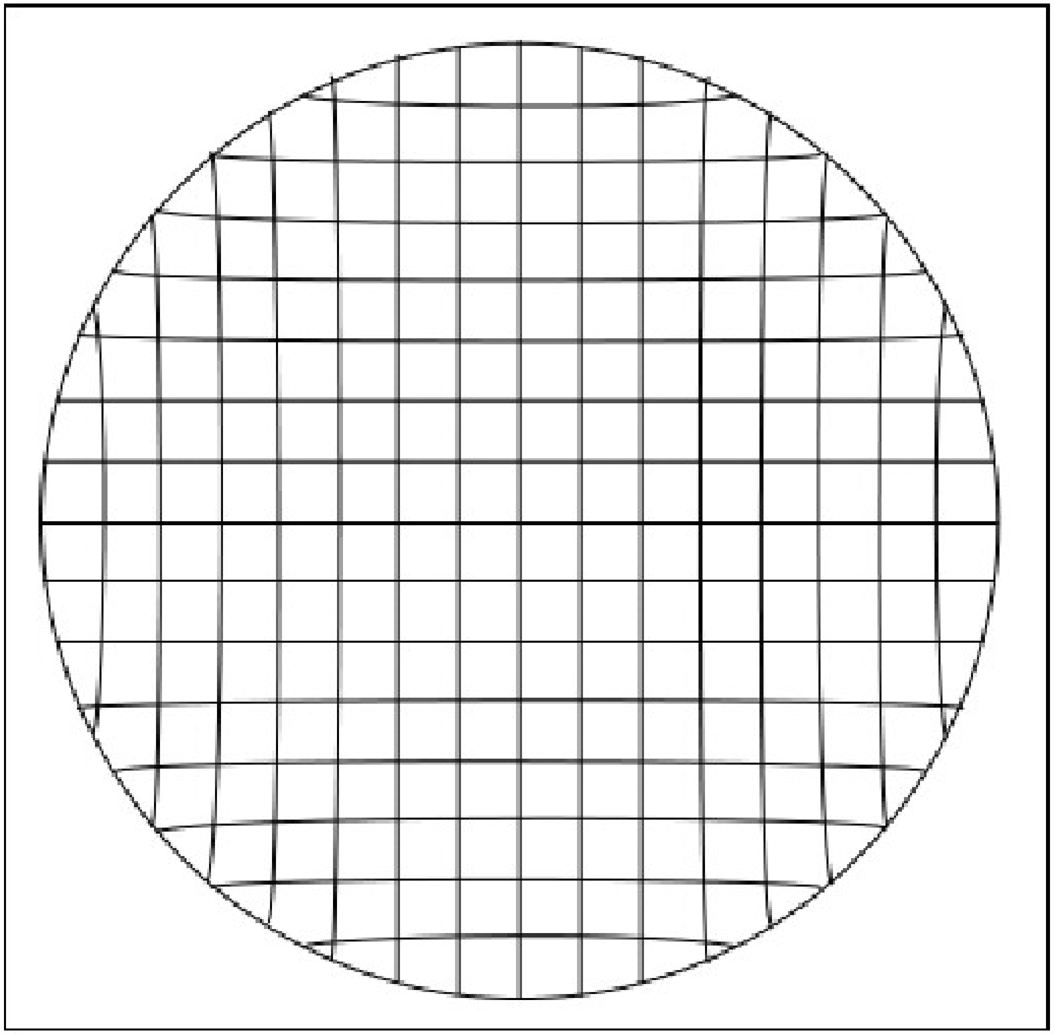
modulation = (150-100)/(100+150) = 0,2

image quality of x-ray image amplifier

quality criteria

- distortions
- uniform illumination
- conversion factor
  (brightness @output / dose rate @input)
- noise
- spatial resolution

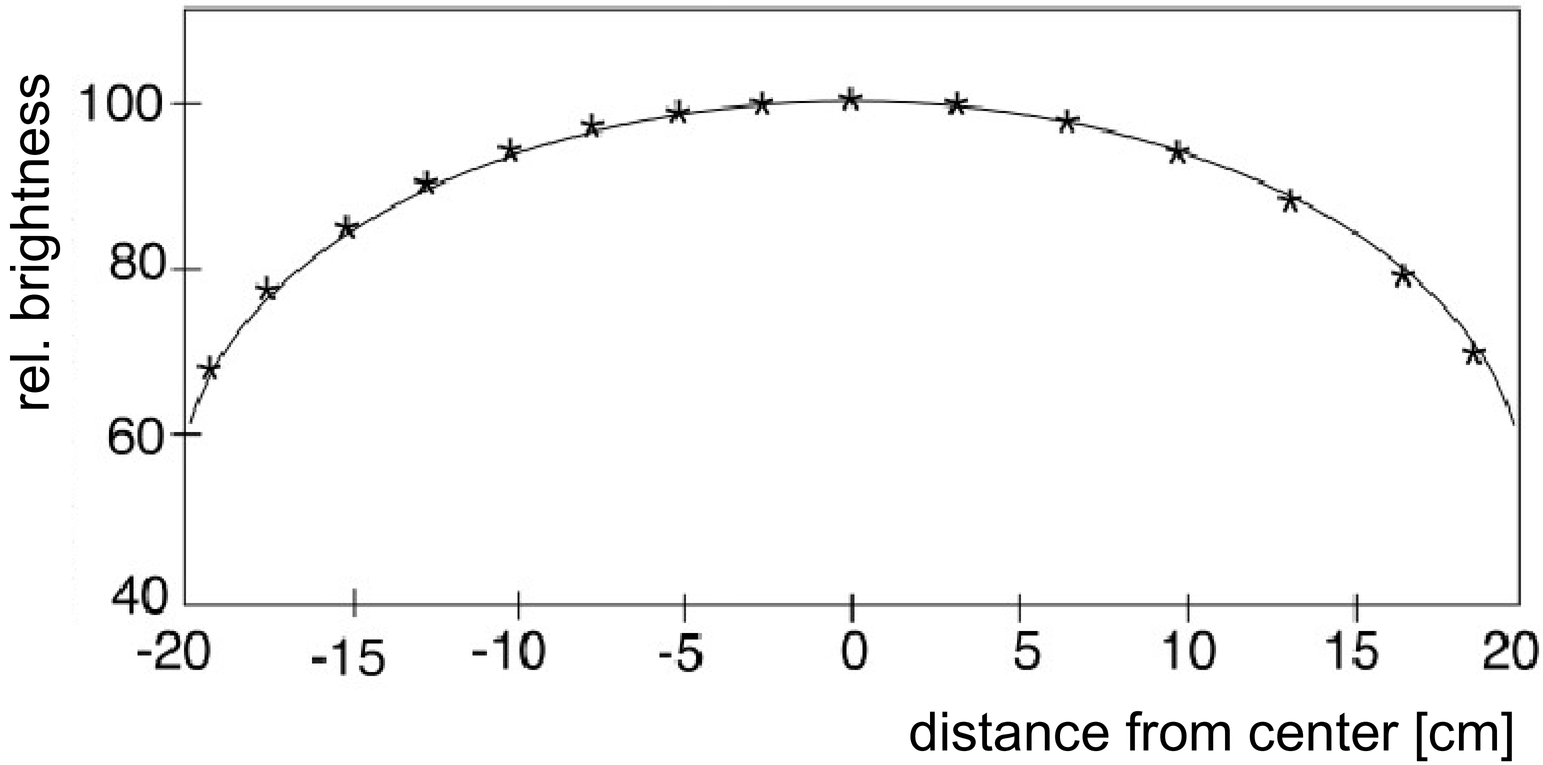
image quality of x-ray image amplifier



saddle-like distortions

- device-specific
- no strong impact on diagnosis
- identify with quadratic mesh
- digital correction

image quality of x-ray image amplifier



„vignetting“

- device specific effect

image quality of x-ray image amplifier — noise

number of detected γ-quanta (x) per unit area and unit time is Poisson distributed:

$$p(x) = \frac{m^x e^{-m}}{x!}$$

Poisson distribution characterized by mean value *m* only!

$$\sigma^2_{Poisson} = m$$

(Gaussian distribution: mean and standard deviation)

the higher the number of γ-quanta , the higher the standard deviation (the smaller the relative spread)

m	σ	σ (%)
10	3,16	31,60
100	10,00	10,00
1000	31,60	3,16

quantum noise:

number of γ-quanta ⇒ image grey level ⇒ noisy image

image quality of x-ray image amplifier — noise

estimating quantum noise (units)

number of quanta per absorbed dose

$$\text{dose} = \frac{\text{number of quanta}}{\text{unit area}}$$

unit: $1/\text{mm}^2$

$$\frac{\text{radiation energy}}{\text{unit area}} = \frac{\text{radiant power} \cdot \text{exposure time}}{\text{unit area}}$$

$$= \frac{\text{number of quanta} \cdot \text{energy of quanta}}{\text{unit area}}$$

unit: J/mm^2

image quality of x-ray image amplifier — noise

estimating quantum noise (technically)

number of quanta per absorbed dose

$$\text{ion dose} = \frac{\text{charge quantity (of given sign) due to ionization in air}}{\text{air volume in measurement chamber @ 760 Torr}}$$

unit: Röntgen (R)

$$\text{ion dose} = \frac{\text{charge quantity (of given sign) due to ionization in air}}{\text{mass of air in measurement chamber}}$$

unit: C/kg = As/kg

100 R = 25.8 mC/kg (air @760 Torr)

image quality of x-ray image amplifier — noise

estimating quantum noise (technically)

number of quanta per absorbed dose

$$\text{absorbed dose} = \frac{\text{energy deposited in body due to radiation}}{\text{mass of object}}$$

unit: J/kg = Gray = Gy

conversion ion dose [Coulomb/kg] → absorbed dose [Gy]
(air,100 keV)

1 Gy = 29.86 mC/kg

image quality of x-ray image amplifier **noise**

estimating quantum noise (technically) ***calibration curve***

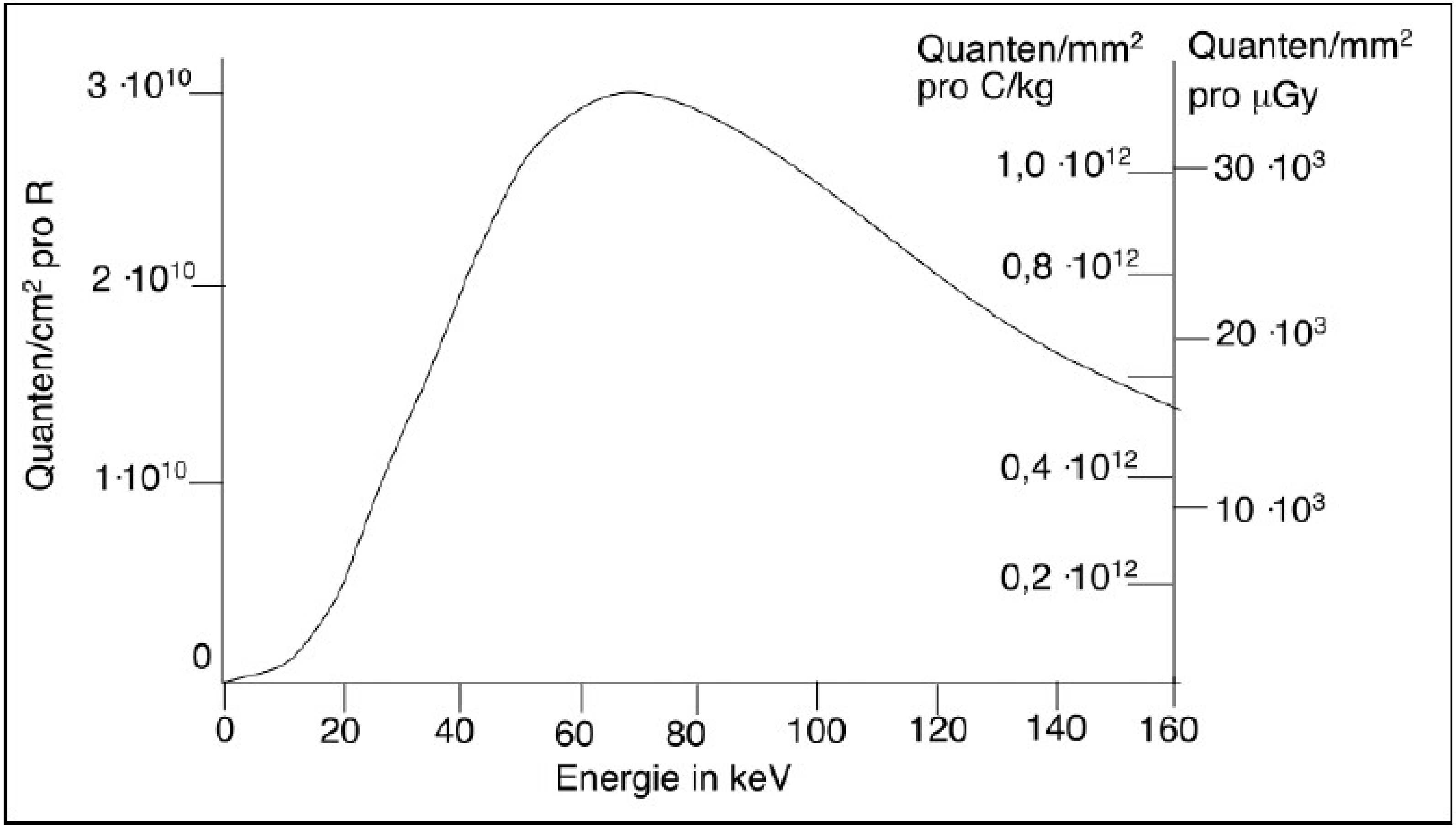


image quality of x-ray image amplifier — noise

estimating quantum noise — *an example*

mean x-ray energy:	80 keV
dose rate:	0.2 µGy/s
pixel size:	0.2 mm x 0.2 mm
exposure time/image:	0.2 s

80 keV $\Leftrightarrow$ $3{,}4 \cdot 10^4$ quanta/(mm µGy)

quantum noise (incoming γ-quanta):
$\Rightarrow$ 54 γ-quanta/pixel; std. dev.: 7.3; rel. std. dev.: 13.5 %

(due to the Poisson distribution; can not be optimized further !!)

image quality of x-ray image amplifier **noise**

estimating quantum noise ***an example***

(continued)

impact of measurement chain (I)
absorption in fluorescent screen @input: 10 %
eff. degree of absorption (CsJ (80keV)): 70 %

$\Rightarrow$ detected quanta (screen): $34 \pm 5{,}8 = 34 \pm$ 17,1 %

$\Rightarrow$ deterioration of signal/noise-ratio in fluorescent screen @input !!

image quality of x-ray image amplifier — noise

estimating quantum noise — *an example*

(continued)

impact of measurement chain (II)
conversion γ-quanta to visible light: 2600 photons / γ-quant
$\Rightarrow$ 2600 x 34 = 88400 photons/image/pixel

but: photon generation is also statistical process !
(e.g., assume: 2600 $\pm$ 100 photons/image/pixel)
with error propagation:
$\Rightarrow$ number of generated photons 88400 $\pm$ 15400 = 88400 $\pm$ 17,4 %

$\Rightarrow$ only minor change of standard deviation
only minor change of signal/noise ratio !

image quality of x-ray image amplifier **noise**

the signal/noise ratio @output
is **not** a good measure for the
quality of an imaging system!

more important is the factor
by which the system deteriorates
the signal/noise ratio!

image quality of x-ray image amplifier — noise

Detective Quantum Efficiency DQE

$$\text{DQE} = \frac{(\text{signal/noise ratio})^2 \text{ @output}}{(\text{signal/noise ratio})^2 \text{ @input}}$$

DQE $\in$ [0,1]

DQE=1 ideal system

for quantum noise only, we have:

$$\text{DQE} = \frac{\text{mean number of detected } \gamma\text{-quanta}}{\text{mean number of incoming } \gamma\text{-quanta}}$$

$\sigma^2_{output}/\sigma^2_{input} = n_{input}/n_{output}$ due to Poisson distribution

image quality of x-ray image amplifier **noise**

Detective Quantum Efficiency DQE

DQE of an imaging chain:

$DQE_{chain} = DQE_1 \cdot DQE_2 \cdot .... \cdot DQE_N$

(since output of component 1 = input of component 2 etc.)

DQE for above-mentioned example:

$DQE_{input\ screen} = 34/54 = 0.63$

after photon conversion:

$DQE_{conv} = (88400/15400)^2 / (34/5.8)^2 = 0.96$

image quality of x-ray image amplifier **noise**

when imaging with x-rays,
image quality is always limited

due to **balance** between

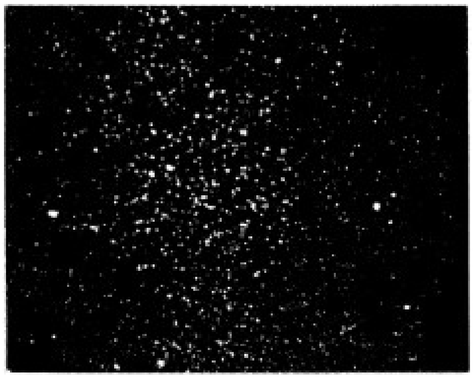
minimization of dose resp. dose rate
(radiation protection)

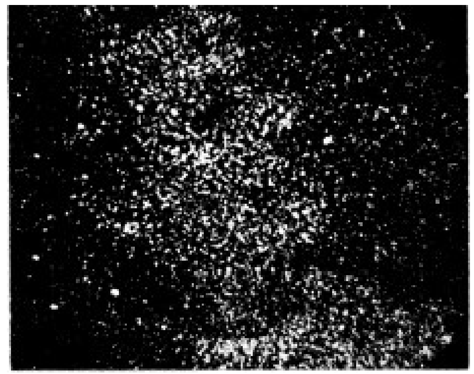
and

maximization of signal
(detectability of details)

image quality of x-ray image amplifier — noise

detectability of details


3000 photons


12000 photons


93000 photons


760000 photons


3,6 Mio. photons


28 Mio. photons

Rose: *Vision*, 1973

image quality of x-ray image amplifier

spatial resolution (image sharpness)

general definition:
separability of adjacent objects (Rayleigh criterion)

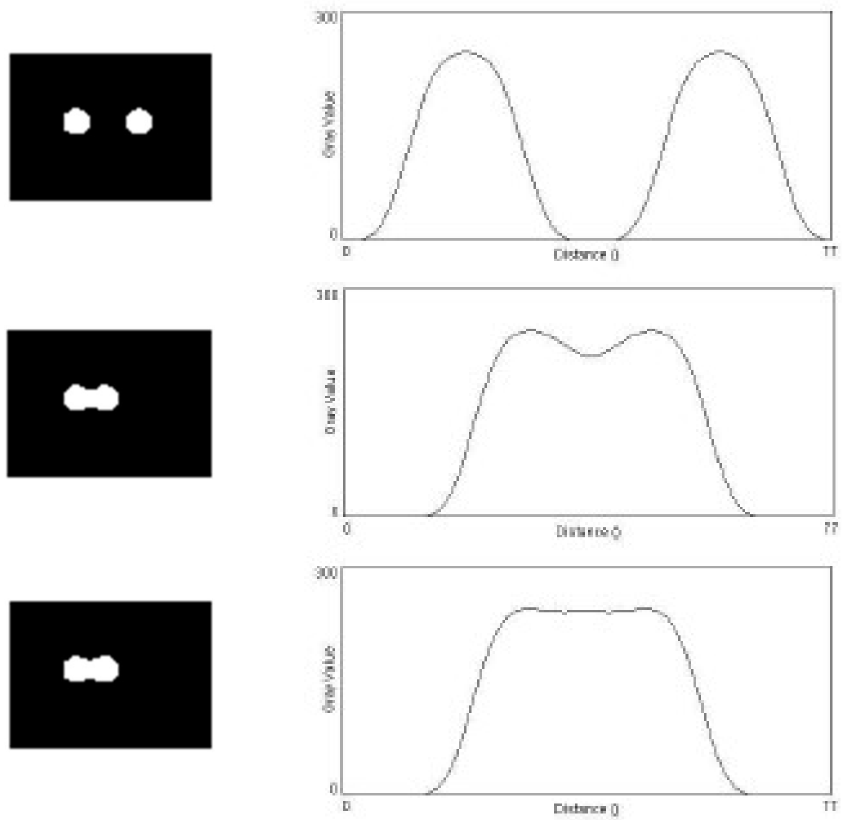
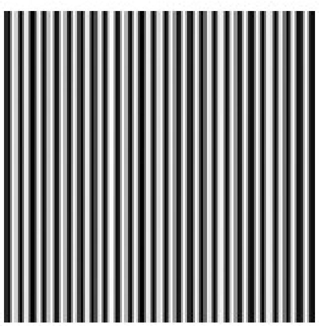


image quality of x-ray image amplifier

spatial resolution

special definition:
Modulation Transfer Function MTF

example: in x-direction sinusoidal modulated image @output

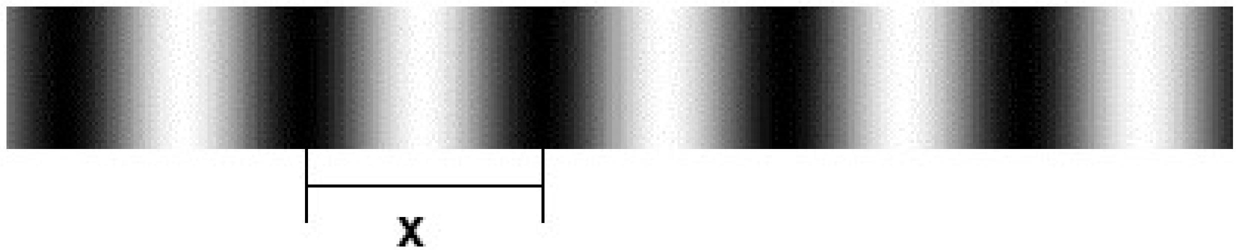

wave length: 20 pixel

$$g(x) = \bar{g} + K_0 \cdot \sin(2\pi \cdot u \cdot x)$$

$g(x)$: grey level of original at position x
$\bar{g}$: mean grey level of original
K_0: amplitude of grey-level modulation
$u=1/\lambda$ spatial frequency of grey-level modulation
λ wave length of grey-level modulation

image quality of x-ray image amplifier

def.: spatial frequency

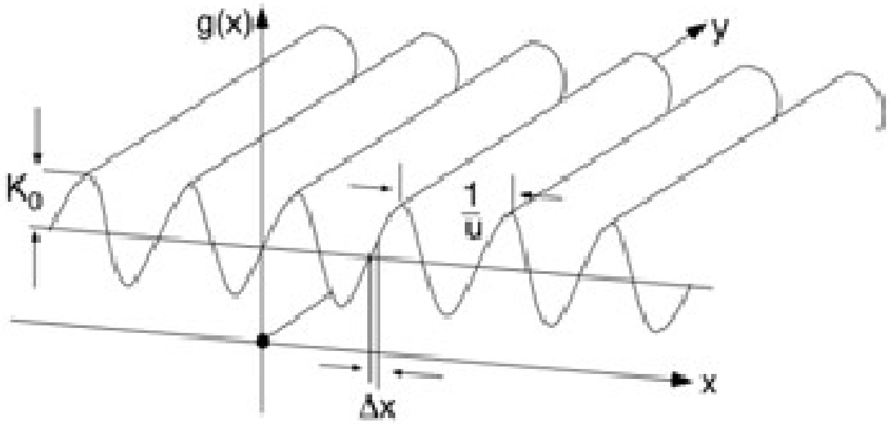


spatial frequency = number of periodically recurring light-dark-modulations (so called line pairs, Lp) per unit length

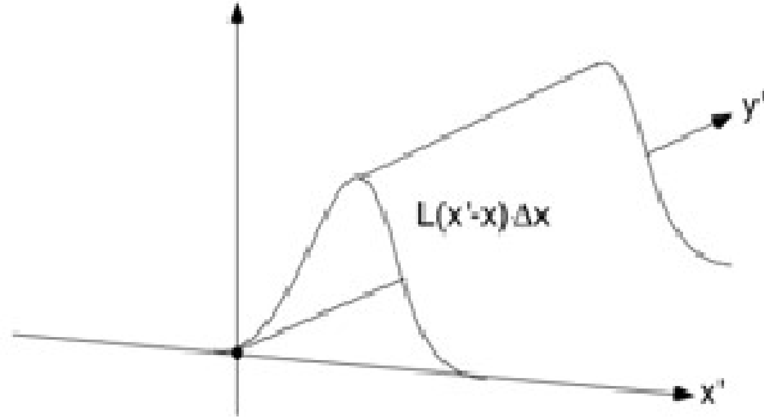
$u = 1/x \quad [mm^{-1}]$

Modulation Transfer Function MTF

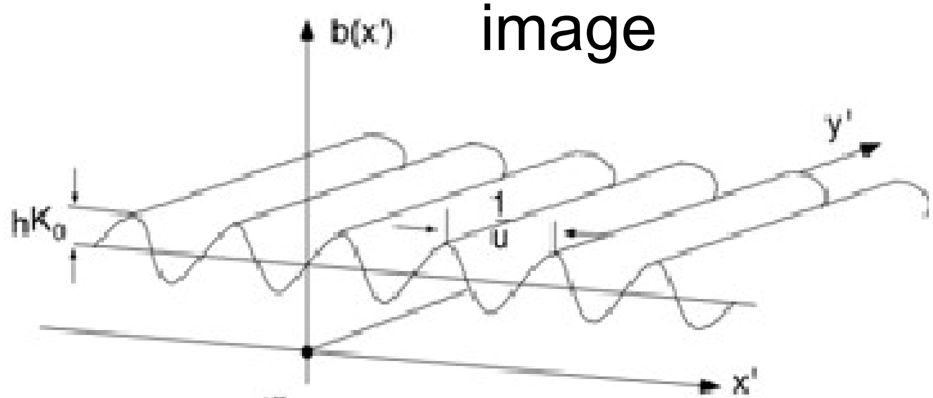
object



imaging system
(**point spread function**)



image



$$b(x') = \int_{-\infty}^{+\infty} g(x)\, L(x'-x)\,dx$$

Modulation Transfer Function MTF

$$b(x') = \int_{-\infty}^{+\infty} g(x) \cdot L(x'-x)dx$$ with $L(x'-x)=$ point spread function (PSF)

$$b(x') = \int_{-\infty}^{+\infty} g(x'-x) \cdot L(x)dx$$

with chosen function g(x)

$$b(x') = \int_{-\infty}^{+\infty} \{\overline{g} + K_0 \cdot \sin(2\pi u(x'-x))\} \cdot L(x)dx$$

$$= \overline{g} \int_{-\infty}^{+\infty} L(x) \cdot dx + K_0 \int_{-\infty}^{+\infty} \sin(2\pi u(x'-x)) L(x)dx.$$

Modulation Transfer Function MTF

$$\int_{-\infty}^{+\infty} L(x)\; dx = 1$$ (normalization of point spread function)

$$\sin(\alpha+\beta) = \sin\alpha\cos\beta + \cos\alpha\sin\beta$$

$$\begin{aligned} b(x') = {} & \overline{g} + K_0 \cdot \int_{-\infty}^{+\infty} L(x) \sin 2\pi u x' \cdot \cos 2\pi u x \cdot dx \\ & - K_0 \cdot \int_{-\infty}^{+\infty} L(x) \cos 2\pi u x' \cdot \sin 2\pi u x \cdot dx \\ = {} & \overline{g} + K_0 \cdot \sin 2\pi u x' \int_{-\infty}^{+\infty} L(x) \cdot \cos 2\pi u x \cdot dx \\ & - K_0 \cdot \cos 2\pi u x' \cdot \int_{-\infty}^{+\infty} L(x) \sin 2\pi u x \cdot dx \end{aligned}$$

Modulation Transfer Function MTF

symmetric PSF: L(x)=L(-x)

with $$\int_{-\infty}^{+\infty} L(x)\sin 2\pi ux \cdot dx = 0$$

and $$\eta(u) = \int_{-\infty}^{+\infty} L(x)\cos 2\pi ux \cdot dx$$

$$b(x') = \bar{g} + K_0 \cdot \eta(u) \cdot \sin 2\pi ux'$$

η depends on spatial frequency of original

$\eta \in [-1,1]$ (normalization of PSF and $-1 < \cos 2\pi ux < 1$)

$\eta \to 1$, if $\lambda = 1/u \gg L(x)$ (unbiased transmission of large λ)

$\eta \to 0$, if $\lambda = 1/u \ll L(x)$ (complete loss of image information)

MTF and contrast

"contrast" of original:

$$\frac{\max[g(x)]-\min[g(x)]}{\max[g(x)]+\min[g(x)]}=\frac{\bar{g}+K_0-(\bar{g}-K_0)}{\bar{g}+K_0+\bar{g}-K_0}=\frac{K_0}{\bar{g}}$$

"contrast" of image:

$$\frac{\max[b(x)]-\min[b(x)]}{\max[b(x)]+\min[b(x)]}=\frac{\bar{g}+K_0\cdot\eta(u)-(\bar{g}-K_0\cdot\eta(u))}{\bar{g}+K_0\cdot\eta(u)+\bar{g}-K_0\cdot\eta(u)}=\eta(u)\frac{K_0}{\bar{g}}$$

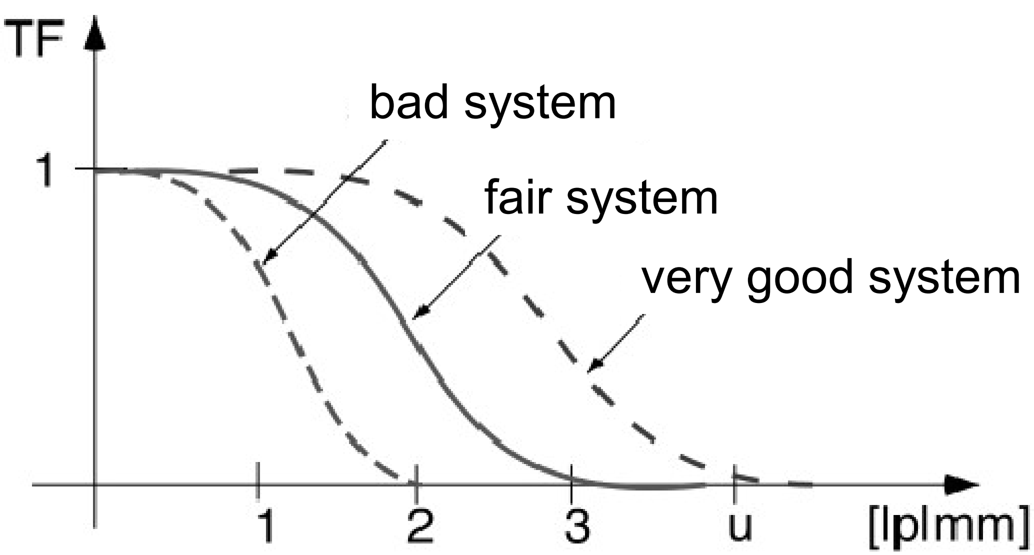
MTF and contrast

for sinusoidal (!!) signals, we have:

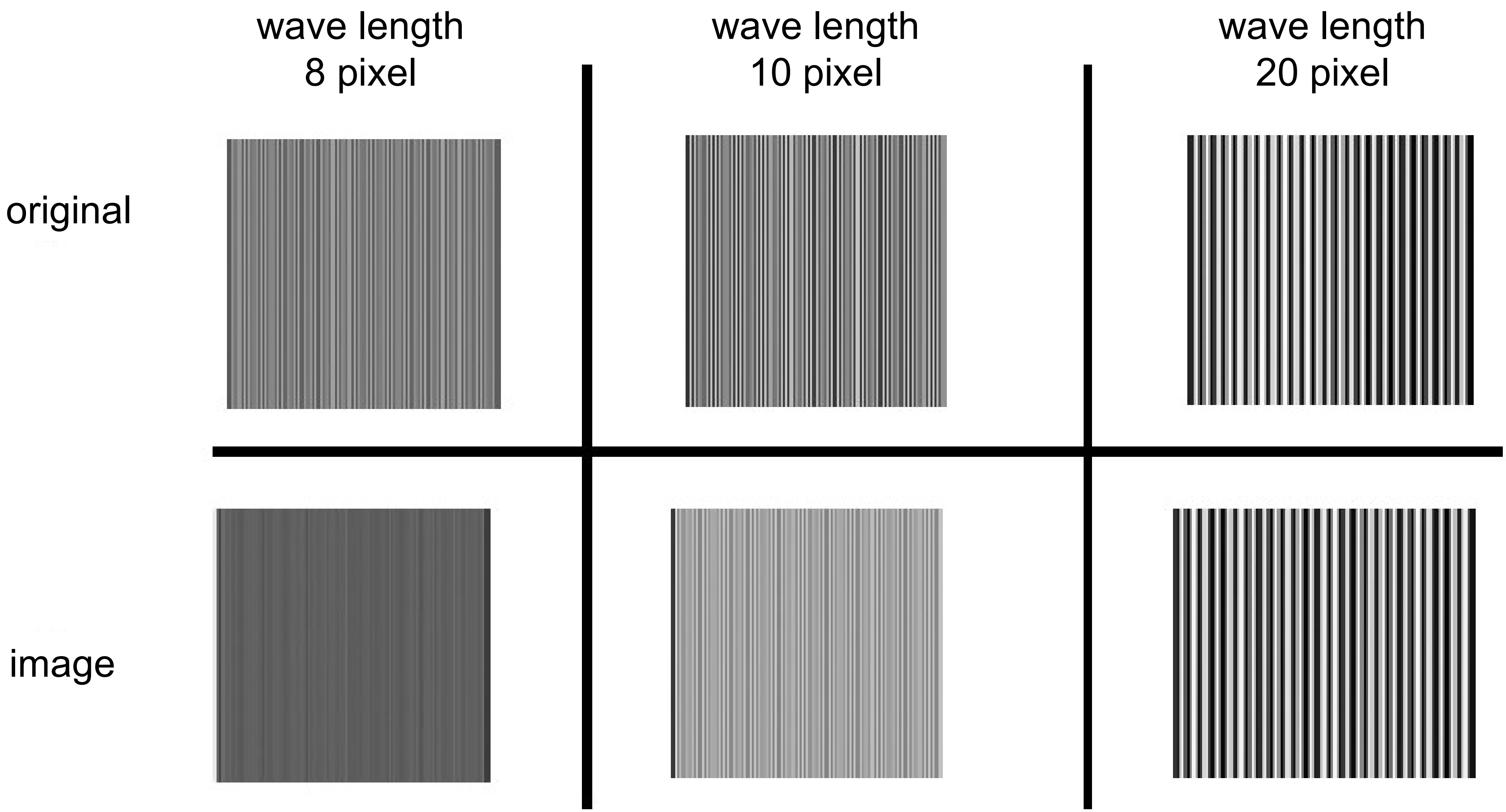
$$\mathrm{MTF}(u) = \frac{\text{“contrast” of image @output at frequency } u}{\text{“contrast” of original at frequency } u}$$

$$\mathrm{MTF}(u) = \eta(u)$$

[u]: line pairs lp/mm



imaging of sine-modulated original

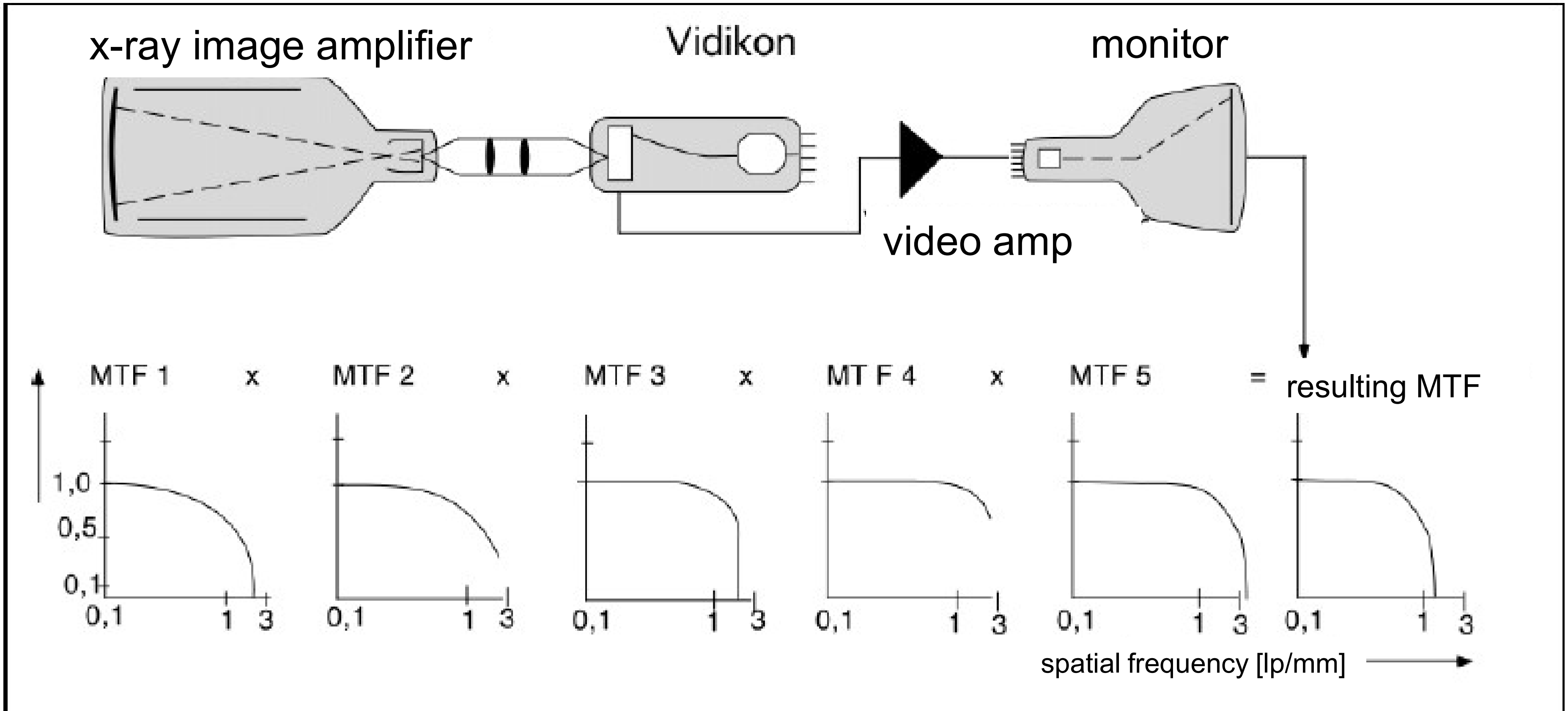


MTF of an imaging chain:

x-ray image amplifier

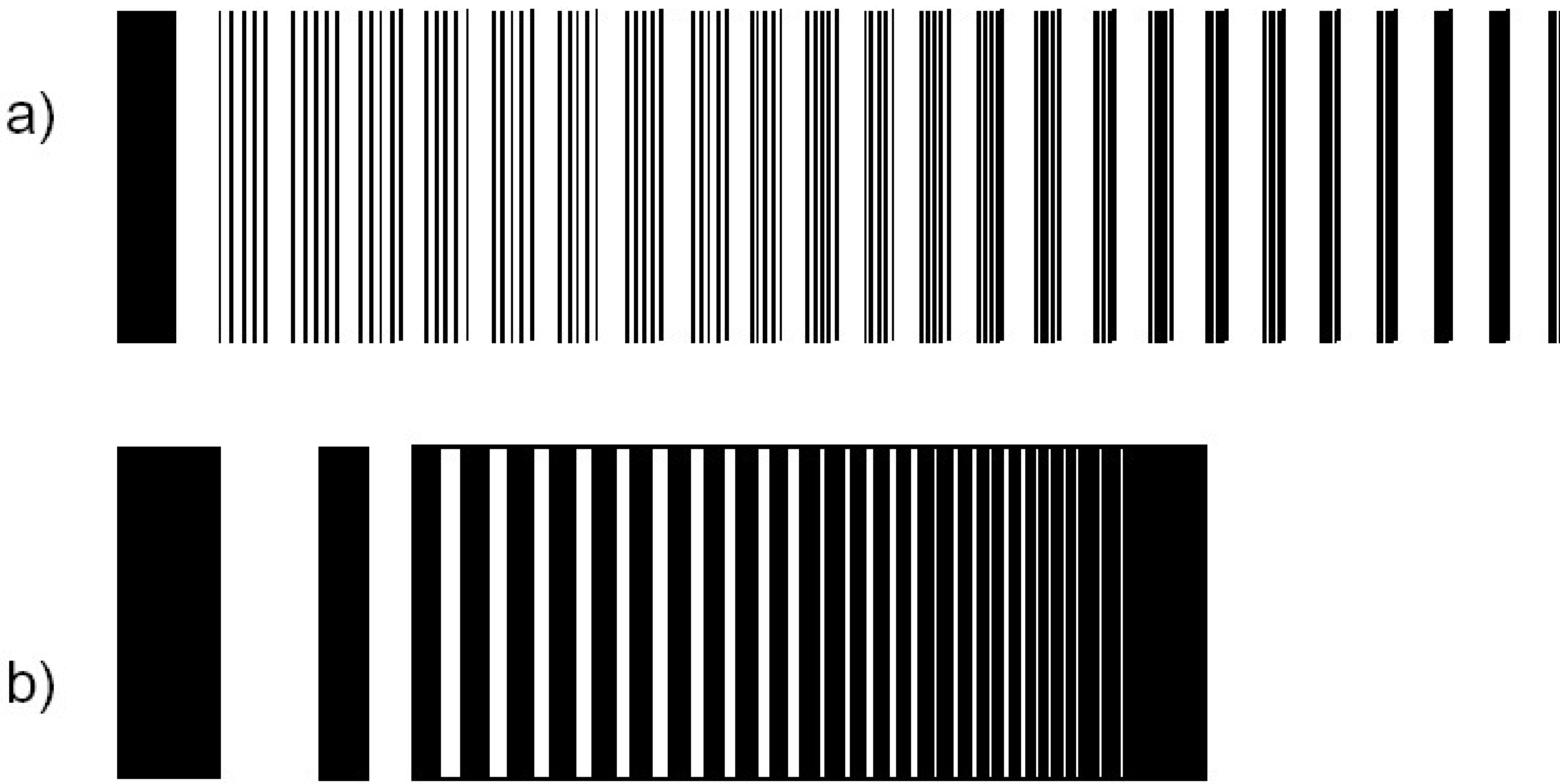
$MTF_{chain} = MTF(u)_1 \cdot MTF(u)_2 \cdot .... \cdot MTF(u)_N$

(since output of component 1 = input in component 2 etc.)



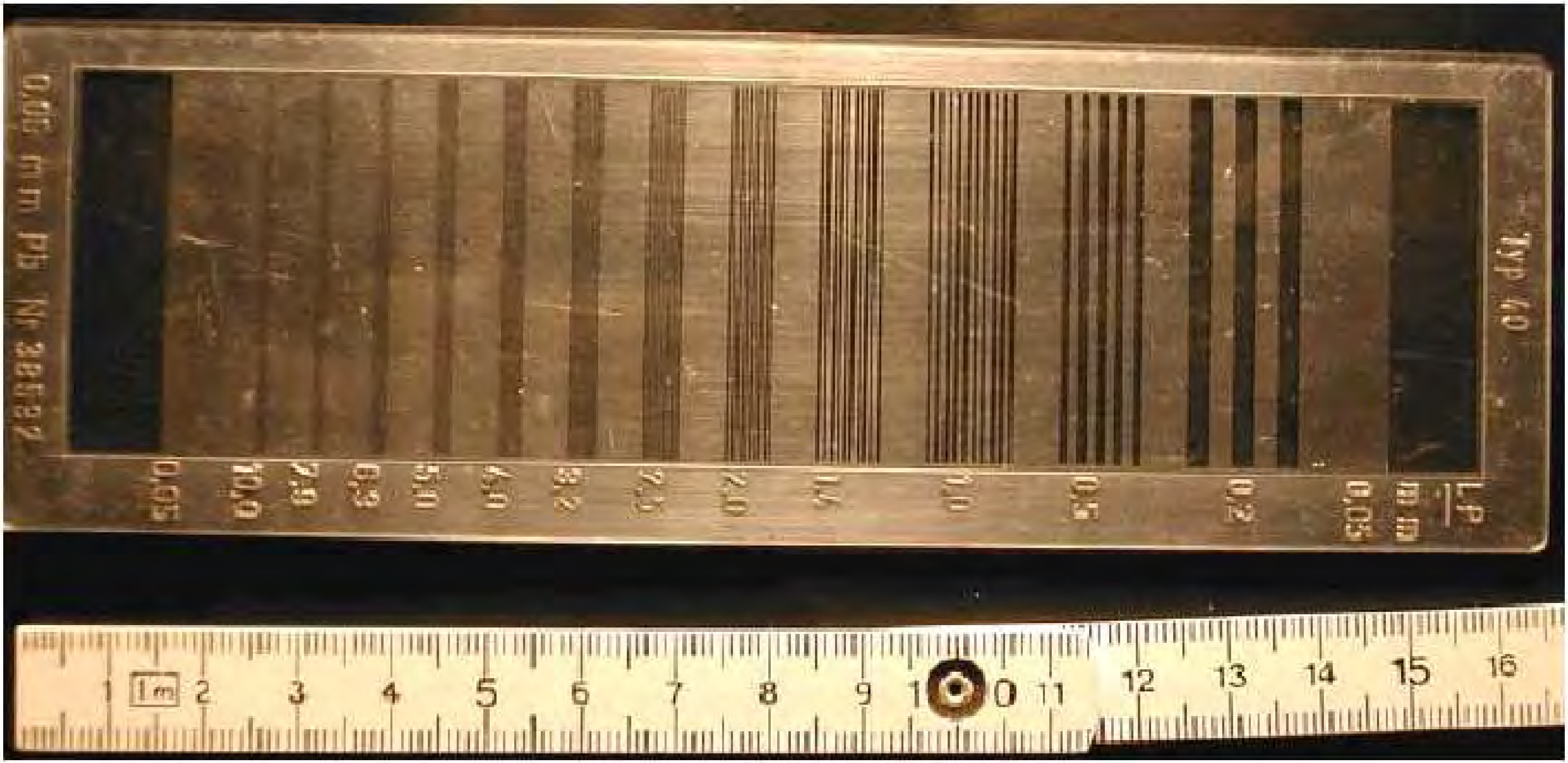
experimental assessment of MTF

lead line test pattern



experimental assessment of MTF

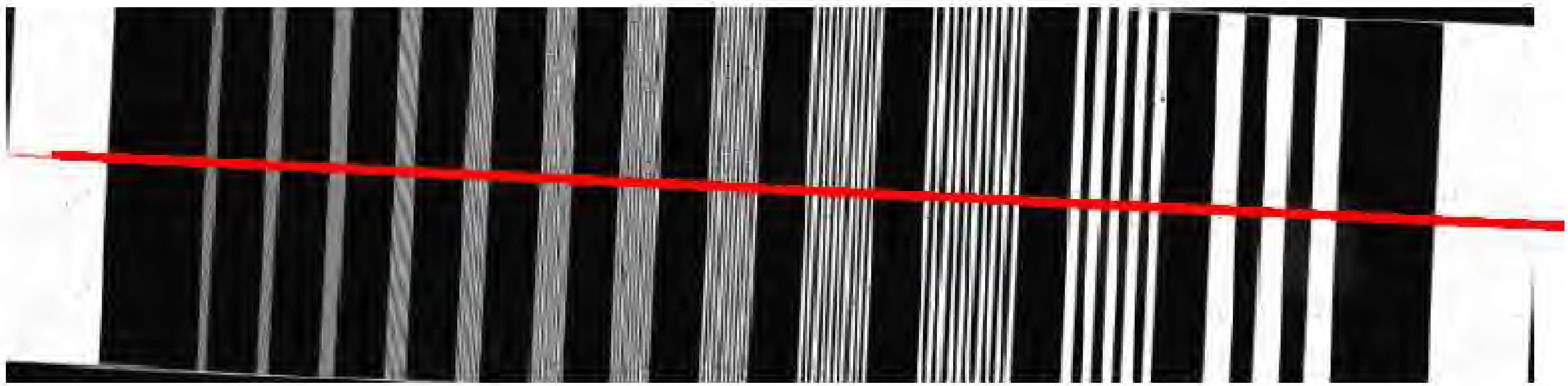
lead line test pattern



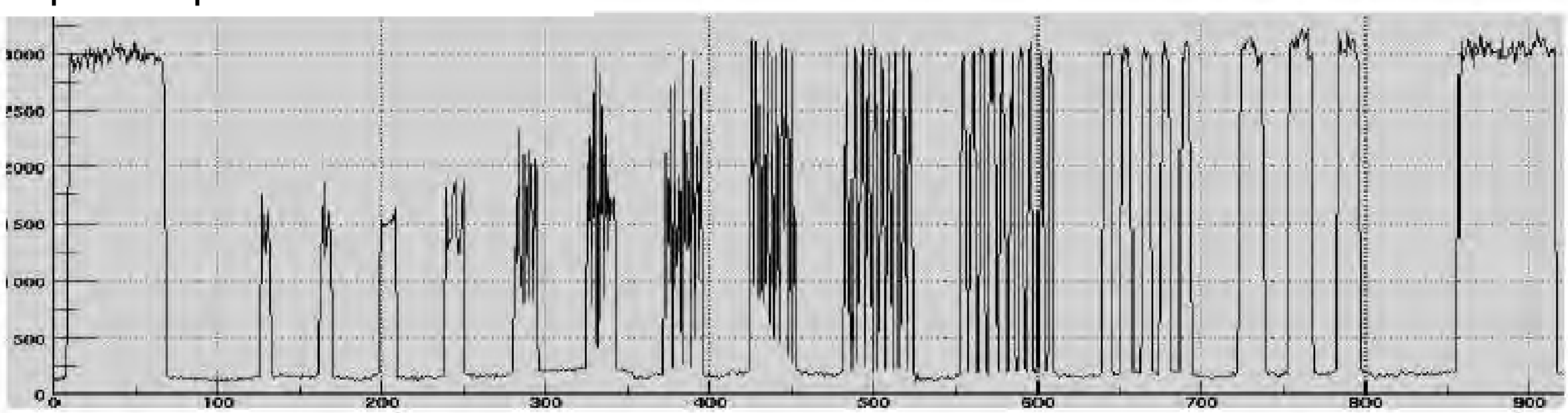
experimental assessment of MTF

lead line test pattern

x-ray image of test pattern



amplitude profile



experimental assessment of MTF

lead line test pattern

provides information about fundamental frequency AND about odd higher harmonics

this follows from the Fourier series expansion of a rectangular function:

$$\mathrm{MTF}(u) = \frac{\pi}{4}\left|R(u) + \frac{R(3u)}{3} - \frac{R(5u)}{5} + \frac{R(7u)}{7} - \ldots\right|$$

where

$$R(u) = \frac{\text{contrast @output}}{\text{contrast @input}}$$

rectangular function

noise and spatial resolution

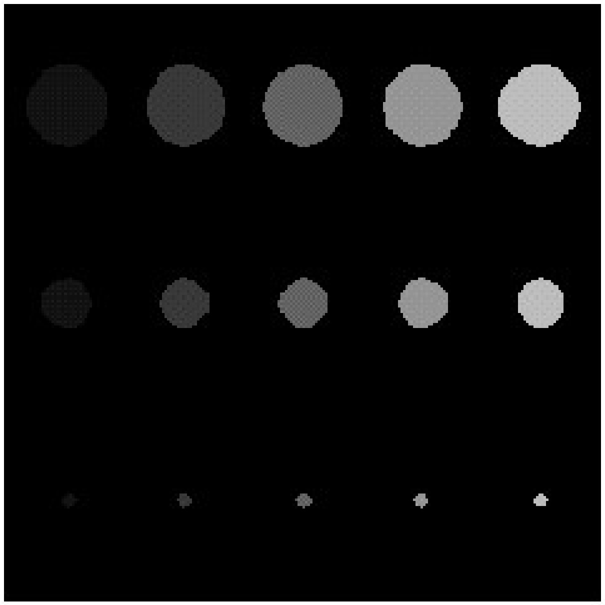
improving DQE

results in

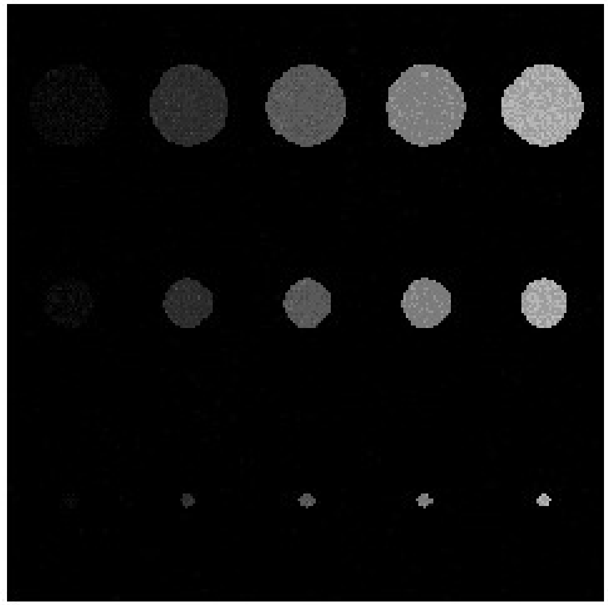
a deteriorated MTF

and vice versa !

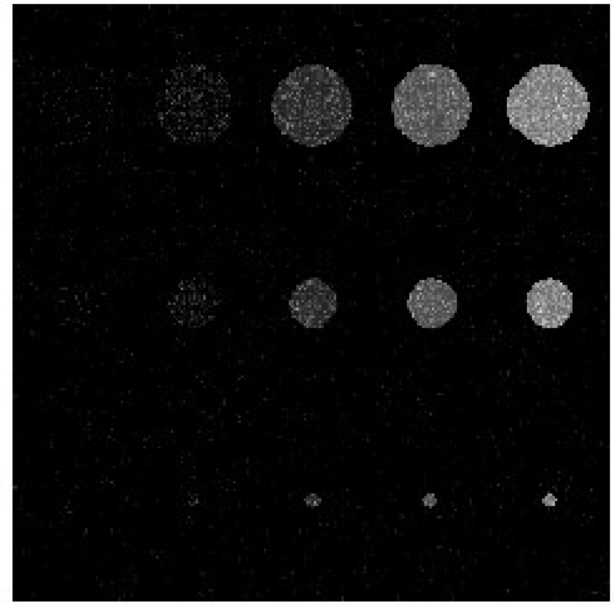
image quality of x-ray image amplifier — noise



no noise



256 quanta/pixel
noise +/- 16



16 quanta/pixel
noise +/- 4