Master‘s Program in Medical Physics

Physics of Imaging Systems

Basic Principles of X-Ray Diagnostic III

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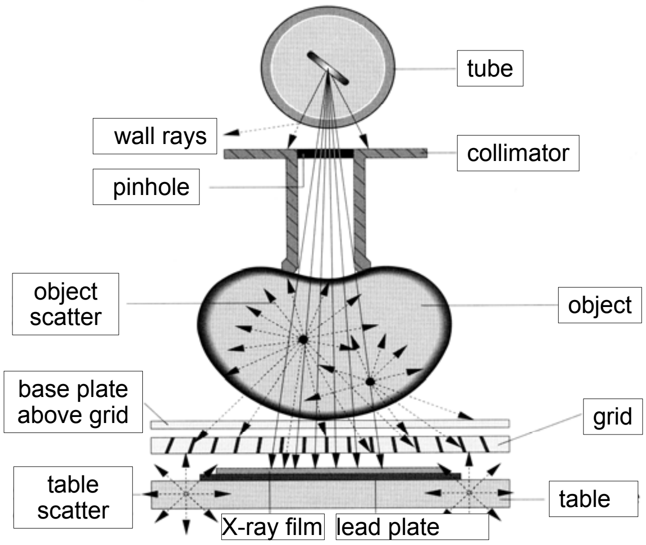
Scattered Radiation



- „scattered radiation fog“ in absorption image
- homogeneous additional film exposure
- contrast reduction
- reduction of SNR of the imaged details

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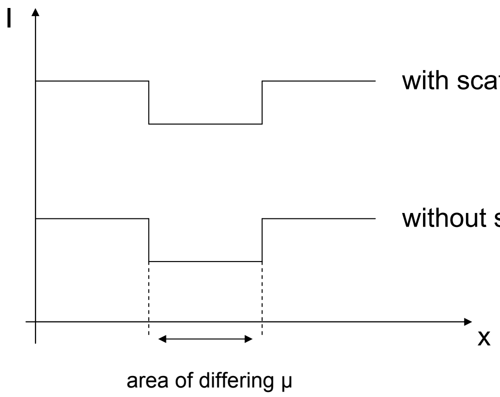
12/9/2008 | Page 3

Scattered Radiation: Intensity Distribution

- intensity distribution behind an object with differing attenuation coefficient μ



source: Dössel. “Bildgebende Verfahren in der Medizin” 2000

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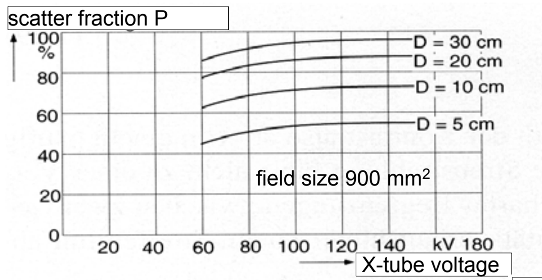
12/9/2008 | Page 4

Scattered Radiation Fraction I

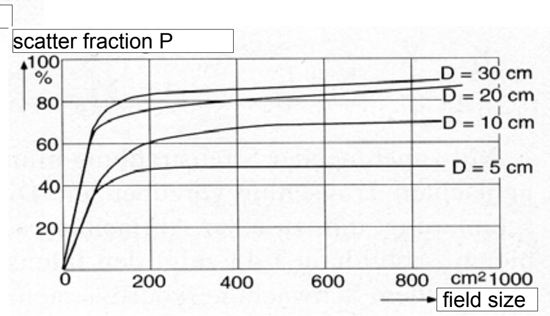


- scattered radiation fraction as a function of tube voltage, patient thickness D, field size

$$P = \frac{I_s}{I_p + I_s}$$

I_s = scattered radiation intensity, i.e. X-ray intensity in the detector plane generated in a non straightforward way

I_p = primary radiation intensity, i.e. X-ray intensity in the detector plane generated in straightforward way from source to detector



source: Dössel. “Bildgebende Verfahren in der Medizin” 2000

Scattered Radiation Fraction II

scattered radiation fraction P and exposure prolongation factor V
(for scatter raster) at different X-ray examinations

object	$U_{Rö}$ [kV]	P [%]	V (Pb 8/40)
head p.-a.	70	45	2.6
lung (medium) p.-a.	120	55	2.6
lung thick p.-a.	120	65	3
pelvis (medium) p.-a.	80	80	5
pelvis (thick) p.-a.	80	85	6
pelvis lateral	80	90 - 95	7 – 8.5

source: Morneburg. "Bildgebende Systeme für die medizinische Diagnostik", 1995

Scattered Radiation: Solutions I

- amount of scattered radiation up to 85% → reduction necessary !

1. distance:

- increase distance between object and detector (scattered X-rays do not hit detector, but: less intensity → higher dose and geometric problem !)

- reduce scatter volume (compression and collimation at object)

2. slit before and after object:

- scattered radiation is shielded, line-scanning can be performed by slit movement across the object (but: complex mechanics, more space, and high intensity → dose problem !)

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12/9/2008 | Page 7

Scattered Radiation: Solutions II

3. filter:

- possibility 1: homogeneous filter between patient and detector = larger distance and absorption of low-energy radiation, selectivity: 1.5 - 2

- possibility 2: filtering in time domain (time-of-flight !), technique has not been fully developed

4. raster

- walls with alternating thin lead foils and penetrable material

- raster is mounted such to penetrate non scattered radiation and to absorb scattered radiation in the lead foils

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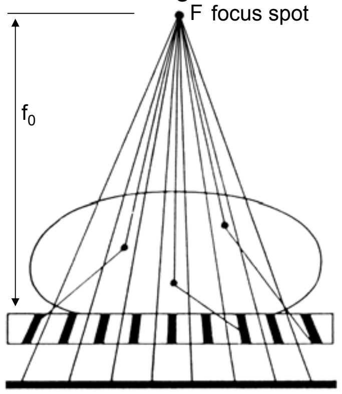
12/9/2008 | Page 8

Scatter Radiation Raster

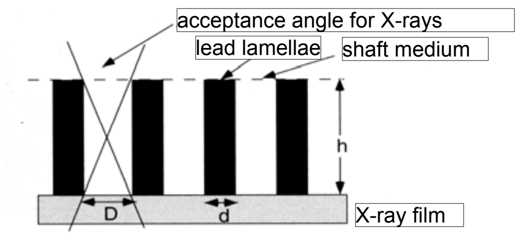
focusing raster



f_0 focus distance

non-focusing raster

lamellae-raster



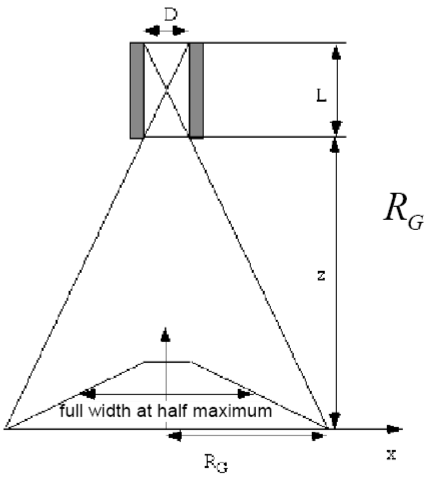
typ. values: d = 0,07 mm, D = 0,18 mm, h = 1,4 mm
between 40 and 75 lines/cm

using higher raster walls scatter radiation can be absorbed more effectively
(but: alignment of lamellae has to be adjusted more carefully to avoid absorption of primary radiation, high intensity necessary → dose problem !)

source: Dössel. "Bildgebende Verfahren in der Medizin" 2000

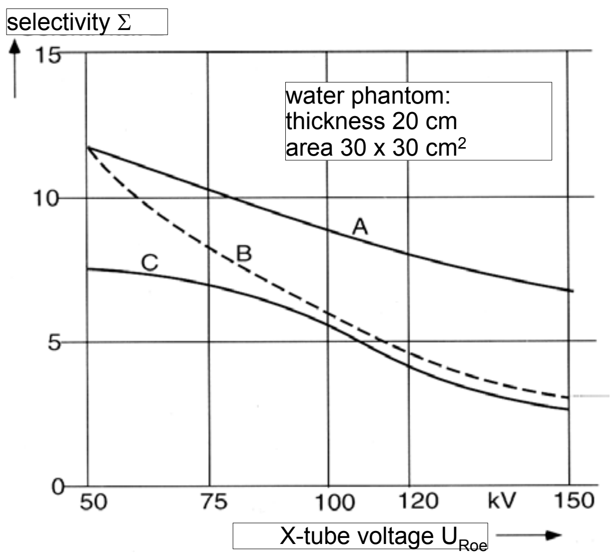
Scatter Radiation Raster Collimation

Collimator



$$R_G = \frac{D}{L}\left(z + \frac{L}{2}\right)$$

Raster Selectivity



$$\Sigma = \frac{T_p}{T_s}$$ selectivity

T_p primary radiation transparency
T_s scattered radiation transparency

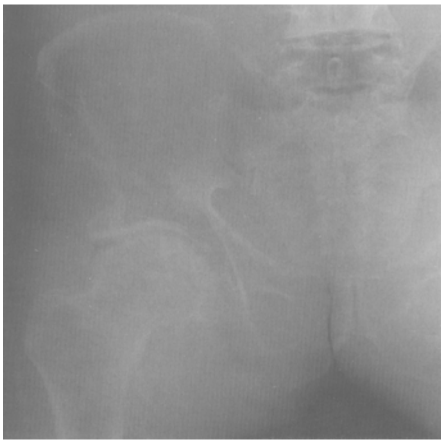
(effective raster: S = 10 (at 75 kV))

r shaft ratio (L/D)
d lamellae thickness

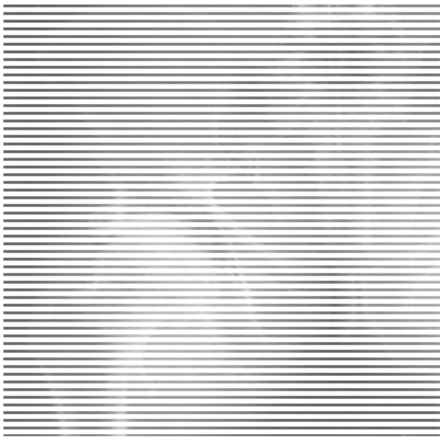
raster A: r = 12, d = 0,07 mm
raster B: r = 12, d < 0,07 mm
raster C: r = 8, d = 0,07 mm

source: Morneburg. “Bildgebende Systeme für die medizinische Diagnostik” 1995

Scatter Reduction: Example



large scattered radiation fraction, strong „scattered radiation fog“, 75 kV, without raster



small scattered radiation fraction, 75 kV, raster with shaft ratio 8

source: Morneburg. “Bildgebende Systeme für die medizinische Diagnostik” 1995

X-Ray Image Intensifier

X-Ray Image Intensifier

X-Ray Image Intensifier: Principle

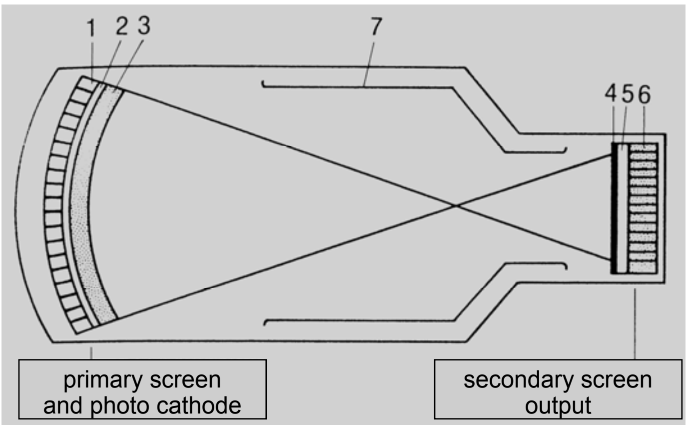
electronic image intensifier for X-ray exposure with screen photograph system or as a part of a television unit, both to reduce dose and adaptation time

principal:

a primary screen (photo cathode in a high vacuum tube) creates an electronic relief corresponding to the X-ray intensity distribution; electrons of this distribution are accelerated and focused on a secondary screen creating a reversed and reduced image with 100 to 1000 higher image intensity which can be observed by an inverting and magnifying optics

Roche Lexicon Medicine, 4. Edition © Urban & Fischer Verlag, München 1999

X-Ray Image Intensifier: Schema



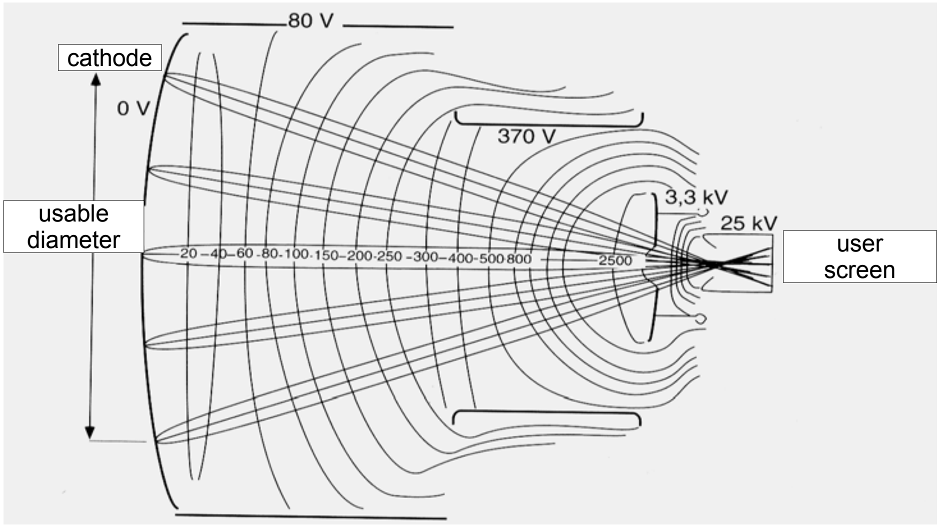
primary screen

(1) aluminum calotte
(2) Na-activated CsI
(3) Photo cathode (Cs-Sb)

secondary screen

(4) Al-layer
(5) fluorescence layer (ZnS, CdS, Ag)
(6) fiber optics plate
(7) electrodes for electron optics

Electric Potential Distribution



source: Dössel. “Bildgebende Verfahren in der Medizin” 2000

Enhancement of Light Density

1. linear reduction of electron image of 1:10
   creates an area reduction of 1:100

   light density is increased by the same factor !

2. photo electrons are accelerated in the electric field
   and can produce more photons in the lightening layer
   due to their higher energy

increasing the brightness by a factor 1000 means:

visual function at observing secondary screen
is performed by uvulas instead of rods → improved visual acuity
and contrast of the human eye !

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12/9/2008 | Page 17

X-Ray Image Intensifier: Tubes

classical RII made of glass. Notice the glass cover, the electrode system made of steel, the primary screen at the upper part, and the secondary screen at the lower part

RII with ceramic technology

RII with enamel technology

source: Electromedica 70 (2002) issue 1

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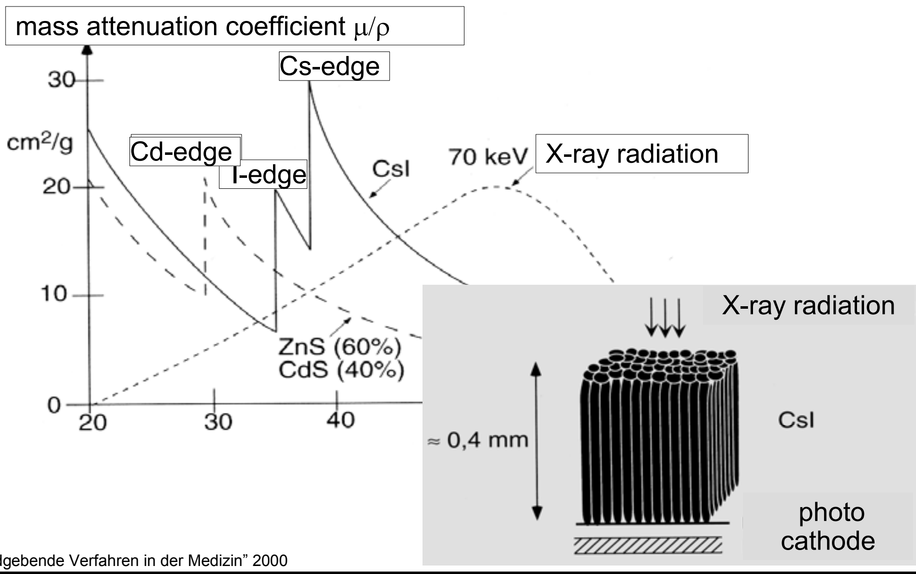
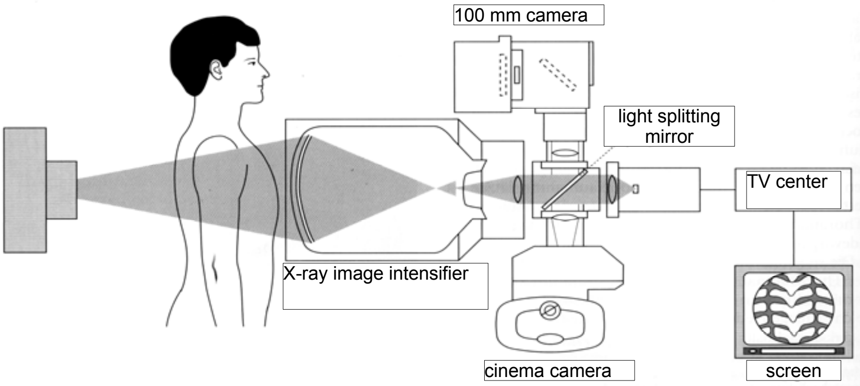
12/9/2008 | Page 18

Mass Attenuation Coefficient: ZnCdS, CsI



source: Dössel. “Bildgebende Verfahren in der Medizin” 2000

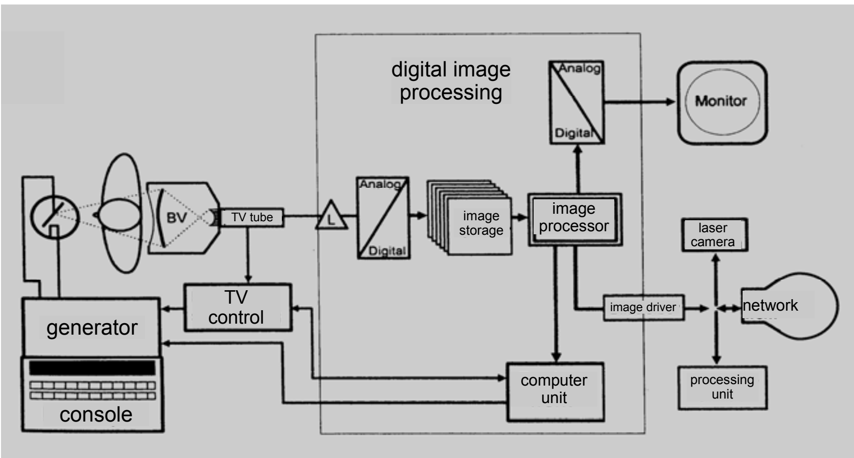
3-Channel X-Ray Diagnostic Unit



partially transmitting light splitting mirror
positioned in the parallel beam direction enables a simultaneous observation
at a television screen at cinema- or single-image-mode

source: Morneburg. "Bildgebende Systeme für die medizinische Diagnostik" 1995

Digital Radiography



digital X-ray examination (fluoroscopy) with digital subtraction

source: Laubenberger and Laubenberger. „Technik der medizinischen Radiologie", Deutscher Ärzte-Verlag 1999

Digital Examination Techniques

advantage:

- reduced exposure dose
- fast result (no film development)
- data processing possible (zoom, contrast enhancement, filter, etc.)
- data archiving (PACS/RIS - systems)
- fast forwarding (online, E-mail)
- ecologically friendly (no chemicals)

disadvantage:

- high initial cost (!)
- digital images can be easily manipulated (!)
- high sophisticated technique
- unusual X-ray image

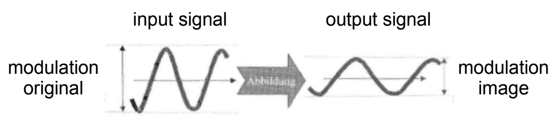
Resolution: Modulation Transfer Function

- resolution is limiting the information of the image

- possible definitions:

  - smallest distance to separate two objects
    disadvantage: not exact since resolution can be dependent on the shape of the objects

  → introduction of **M**odulation **T**ransfer **F**unction (MTF)



www.pi4.physik.uni-erlangen.de/Giersch/ SeminarSS2003/Bildqualitaet.pdf

Modulation Transfer Function MTF I

- most imaging processes can be described by convolution operation

$$G = C * g$$

G: image, C: convolution mask determining the imaging process, g: original

- in frequency space (Fourier space) the convolution is described by point wise multiplication

$$\hat{G} = \hat{C} \cdot \hat{g}$$

- small structures: high frequencies define resolution → MTF

$$MTF(\omega) = | \hat{C}(\omega) / \hat{C}(0) |$$

- MTF: quality function that describes how structures of certain size (defined by their frequency) are suppressed
  - $MTF(\omega) = 1$: no suppression
  - $MTF(\omega) = k$: suppression to k

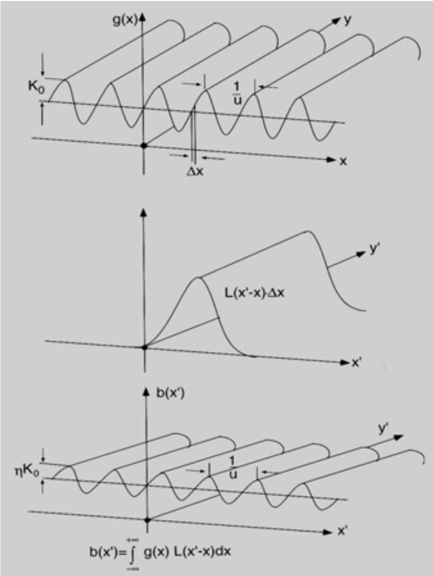
Modulation Transfer Function MTF II

MTF defines how good objects with different details and contrast (object contrast) can be represented by intensity contrast in the image (image contrast) → modulation of spatial frequency [Lp/mm]

characteristic quantities:

(1) visual limit of resolution: limiting visual detectable representation of a high contrast object
(2) spatial frequency where modulation is less than 4% (limiting frequency)
(3) modulation at spatial frequency 1 Lp/mm (characteristic modulation, requested resolution according to RöV guideline)
(4) spatial frequency where modulation is less than 2% (limiting resolution, about 4 Lp/mm)

MTF Example



imaging of a lamellar object g(x) into image b(x´) using convolution function L(x)

$g(x)$ = grey value of the original at position x

$\bar{g}$ = mean grey value of the original

K_0 = amplitude of grey value modulation

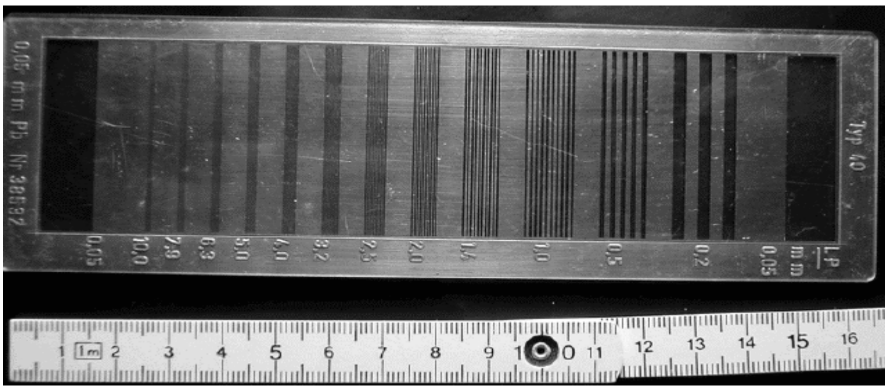
$u = 1/\lambda$ = spatial frequency of grey value modulation

λ = wavelength of grey value modulation

source: Dössel. "Bildgebende Verfahren in der Medizin" 2000

Measurement of MTF

- for MTF measurement lattice with different lattice parameters are used routinely



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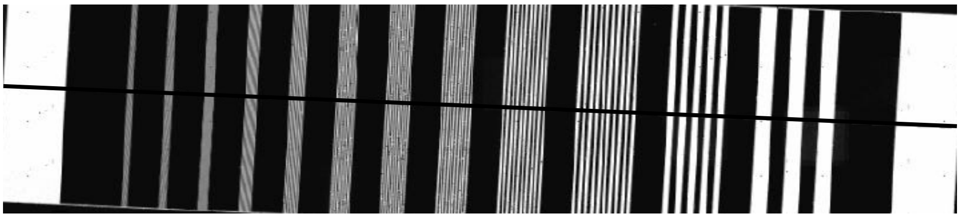
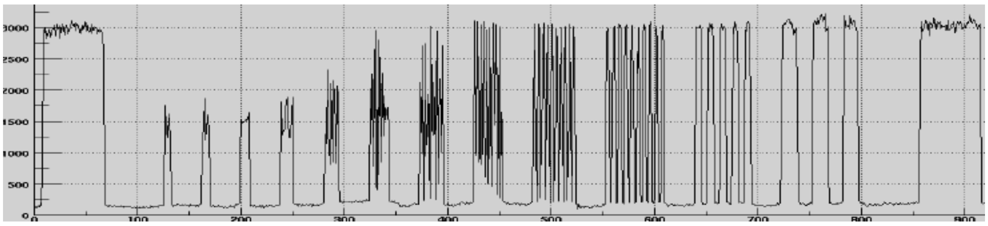
Measurement of MTF: Results

- X-ray image of lattice



- profile



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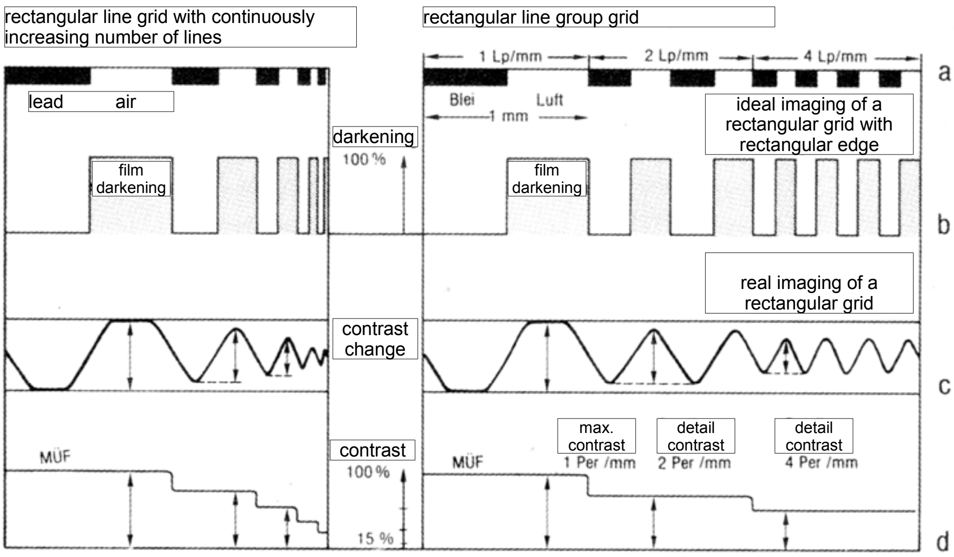
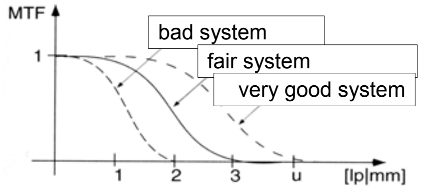
Measurement of MTF: Principle



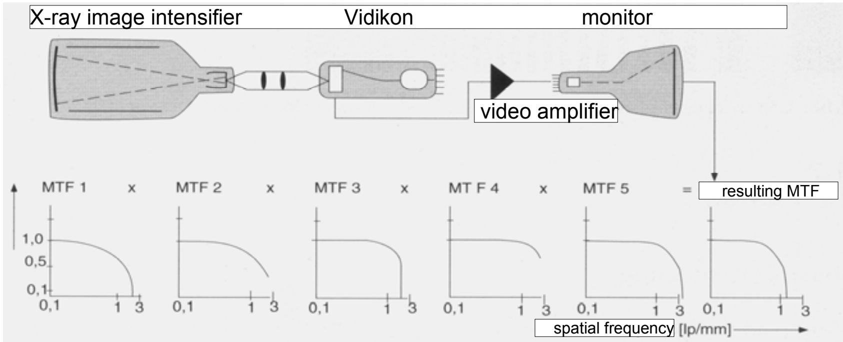
source: Laubenberger and Laubenberger. „Technik der medizinischen Radiologie“, Deutscher Ärzte-Verlag 1999

MTF of Total System



- typical characteristics of MTF(u) = | Ĉ(u) / Ĉ(0) |
  relevant details in medicine: 0 - 2 Lp/mm

$$MTF_{system} = MTF_{comp1}(u) \times MTF_{comp2}(u) \times \ldots$$



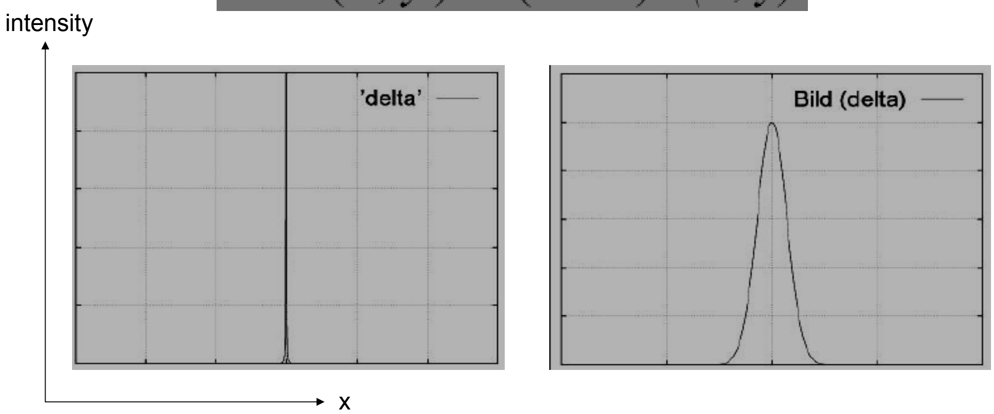
MTF of an image intensifier system

source: Dössel. “Bildgebende Verfahren in der Medizin” 2000

MTF and Point Spread Function PSF

- PSF is the response function of the system to a delta function $\delta(x,y)$

$$PSF(x, y) = (\mathrm{Bild})\delta(x, y)$$



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MTF Definition Using PSF

MTF definition by Fourier transformation of PSF

$$\mathrm{MTF}(u,v) = \left|(\mathrm{FT})\ \mathrm{PSF}(x,y)\right|$$

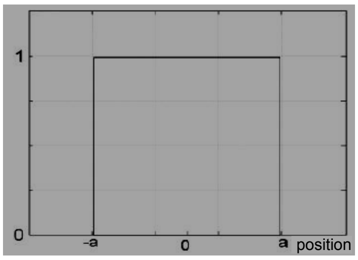
with normalization using value at position (0,0)

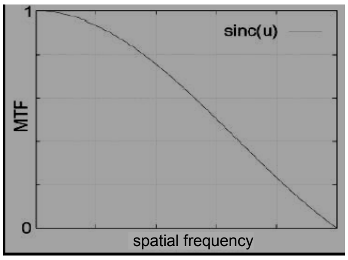
- MTF is defined by the absolute value of the complex transfer function at (0,0) and normalized to 1
- each part of the imaging system reduces the resolution → multiplication with respective MTF
- high frequency parts in the image are often the result of noise
  Poisson noise (source quantum noise)
  Gaussian noise (detector, background, etc.)

Theoretical MTF of Pixel Detector

- the theoretical MTF of a pixel detector is the pixel aperture function in the position space, i.e. the Fourier transform of the pixel geometry

$$\mathrm{FT}(\mathrm{kasten}(a,x)) = \int_{-a}^{a} dx \exp(2\pi i u x) = \frac{\sin(\pi a u)}{\pi u} = \mathrm{sinc}(a,u)$$





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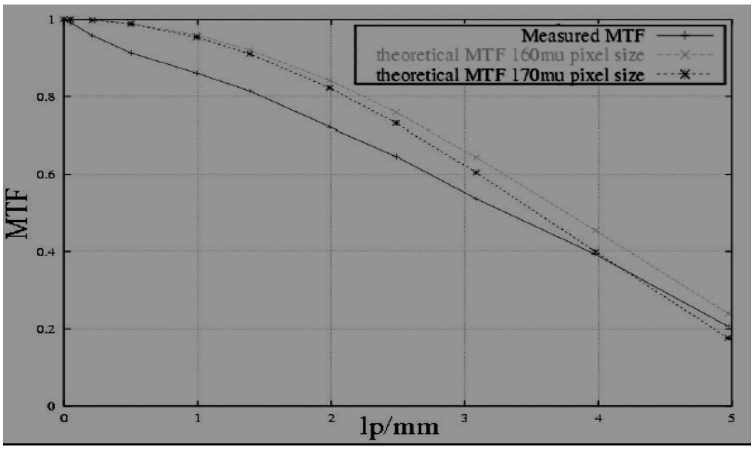
Comparison: Theoretical & Measured MTF

- comparison: theoretical MTF → real detector
(example Medipix 1, pixel size 170 μm)



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Contrast

quantitative difference

- high contrast regime:
  contrast is much more larger than noise
  e.g. bones or metallic objects (possible problems of dynamic range)
  visibility of object is limited by spatial resolution

- low contrast regime:
  contrast is only very little above noise
  e.g. soft tissue in CT or radiography
  visibility of object is limited by signal-to-noise ratio (SNR)

Noise

- noise is not reproducible
  (in contrast to imaging errors and artifacts)
- noise can only be described statistical
- e.g. quantum noise (source)
  film core noise (detector)
  conversion noise (detector)
  electronic noise (detector)
- exception: „Fixed Pattern Noise“: detector shows always a reproducible sensitivity distribution
  → can be easily eliminated by calibration
- quantum noise (source): number of detected quant's / area / time are statistically fluctuating (Poisson distribution) → can not be avoided !
- electronic noise (detector): technical problem, improvement by shortening and shielding of cables, cooling, early signal digitization
  (Gaussian noise = additive noise)

Gaussian and Poisson Noise

$$N(x;\mu,\sigma)=\frac{1}{\sqrt{2\pi}\sigma}e^{-\frac{(x-\mu)^2}{2\sigma^2}}$$

Gaussian noise

- with μ the mean value and σ the standard deviation
- Gaussian noise is additive,
  i.e. signal = original signal + noise
- Gaussian noise is typical for environment noise
- Gaussian noise is also typical for detector noise (electronics etc.)

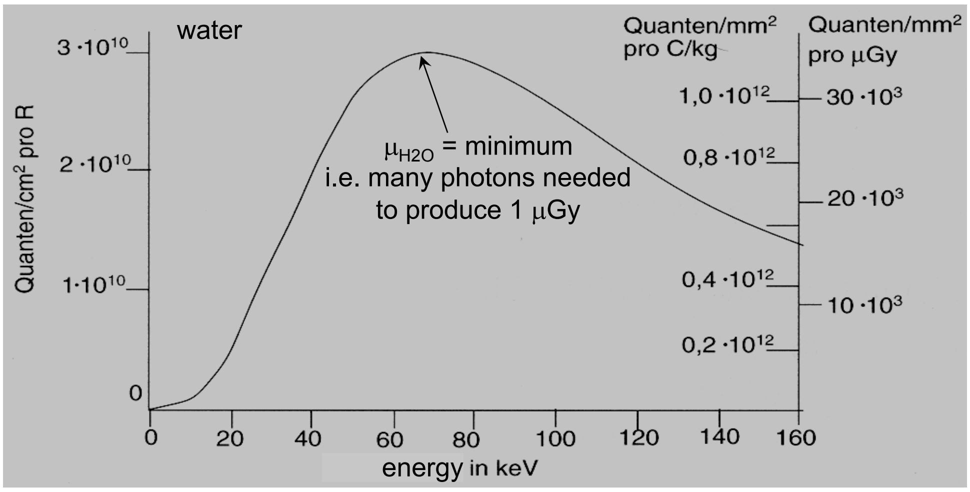
$$P(X=n)=\frac{(\lambda)^n}{n!}e^{-\lambda}$$

Poisson noise

- probability P of n events in a time interval, with λ the average value and $\sqrt{\lambda}$ the standard deviation
- Poisson process: appearance of event (emission of photon) is independent from previous one
- Poisson noise occurs e.g. when electron is emitted from cathode and during absorption
- each pixel value in the image is therefore a stochastic variable

Number of Quant's

- conversion factor number of quant's / mm^2 / µGy as a function of quantum energy



source: Dössel. "Bildgebende Verfahren in der Medizin" 2000

Number of Quant's: Example

- given:
- X-ray energy: 80 keV
- dose: 0.2 μGy/s
- pixel size: 0.2 x 0.2 mm^2
- exposure t: 0.2 s per image

- from former diagram: 80 keV = 3.4 10^4 quant's/mm^2 μGy
- calculation: 272 quant's per pixel and per second and
  54 quant's per pixel and per image

- Poisson statistic:
  statistical error of coming quant's corresponds
  to standard deviation $\sigma = \sqrt{54} = 7.3$

  → quantum noise = 54 ± 7.3 = 54 ± 13.5%

source: Dössel. "Bildgebende Verfahren in der Medizin" 2000

Detective Quantum Efficiency DQE

DQE = ratio of detected photons relative to incoming photons

$$\mathrm{DQE} = \mathrm{SNR}^2_{out} / \mathrm{SNR}^2_{in}$$

- signal: ~ N (number of quant's), noise: ~ √N
  → SNR = signal / noise = N / √N = √N

- describes the degree of efficiency to transfer incoming X-rays into an image signal

- ideal detector has a DQE of 100% !

- the DQE of film-detector-systems is significantly larger than film-foil-systems

- a large DQE is an indispensable prerequisite for potential dose reduction without losing image quality

Noise Equivalent Quanta NEQ

• a non ideal detector has a bad SNR

$$\mathrm{SNR}_{\text{non-ideal}} = \mathrm{SNR}_{\text{out}} = \frac{S_{\text{non-ideal}}}{\sigma_{\text{non-ideal}}} < \sqrt{N_{\text{in}}}$$

• the measured SNR can be allocated by a quantity *N´* called NEQ with the following definition

$$N' = \mathrm{NEQ} = \mathrm{SNR}^2_{\text{out}} < N_{\text{in}}$$

• NEQ is the number of photons needed by an ideal detector to reach a given SNR

DQE and NEQ

• division of NEQ by the number of required photons results in DQE

$$DQE = \frac{NEQ}{N_{in}} = \frac{SNR_{out}^2}{SNR_{in}^2}$$

• fraction of X-ray quant's which are converted into light quant's for imaging

DQE Overview

system properties of different digital radiography units

technique	pixel size (mm)	resolution Lp/mm	dynamic	quantum efficiency (70 kV, 0 Lp/mm)
film / foil (400)	–	5	1:30	20%
image intensifier	0,15–0,4	3,3–1,3	1:100	20%
storage foil	0,2	2,5	1:40 000	25%
selenium barrel (Philips: Thoravision)	0,2	2,5	>1:10 000	60% (60 kV)
CCD technique (Swissray)	0,17	3	1:4000	40%
flat bed (scintillator) (Sterling)	0,139	3,6	>1:10 000	43%
Trixel	0,143	3,5	>1:10 000	60%
General Electric	0,2	2,5	>1:10 000	80%

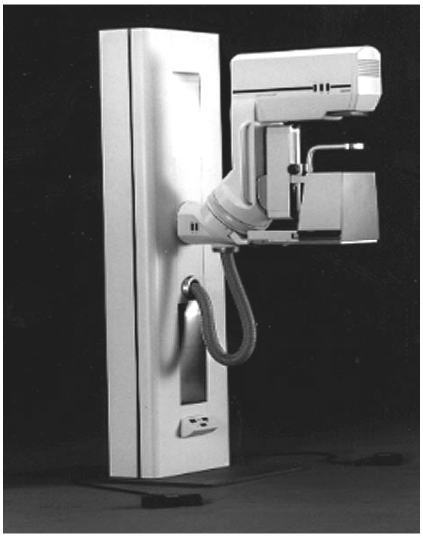
Busch et al. Radiologe 1999

X-Ray Applications

X-Ray Applications

Mammography Systems

Philips mammo Diagnost 3000



Siemens Type 300

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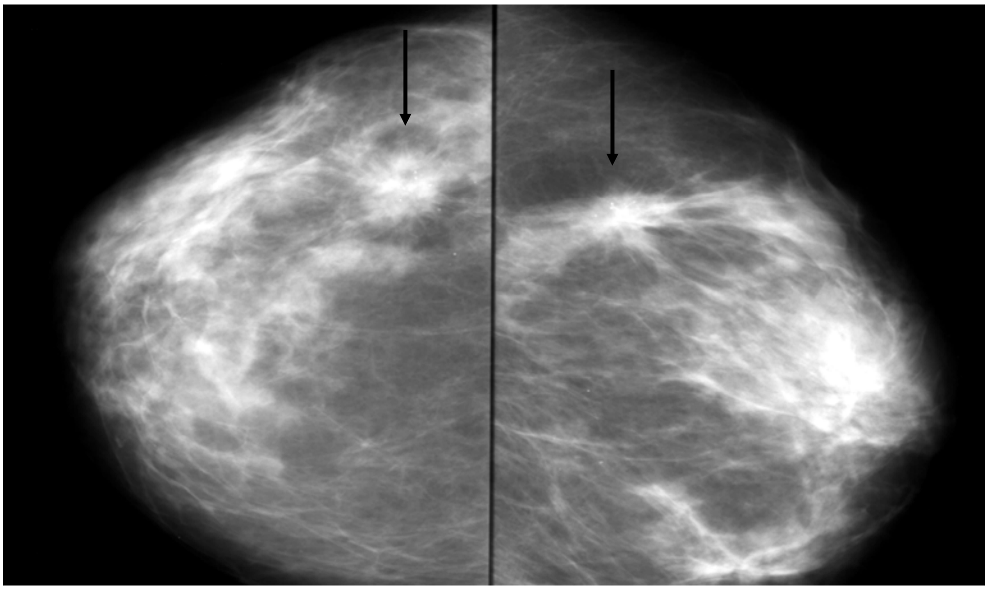
Mammography Images



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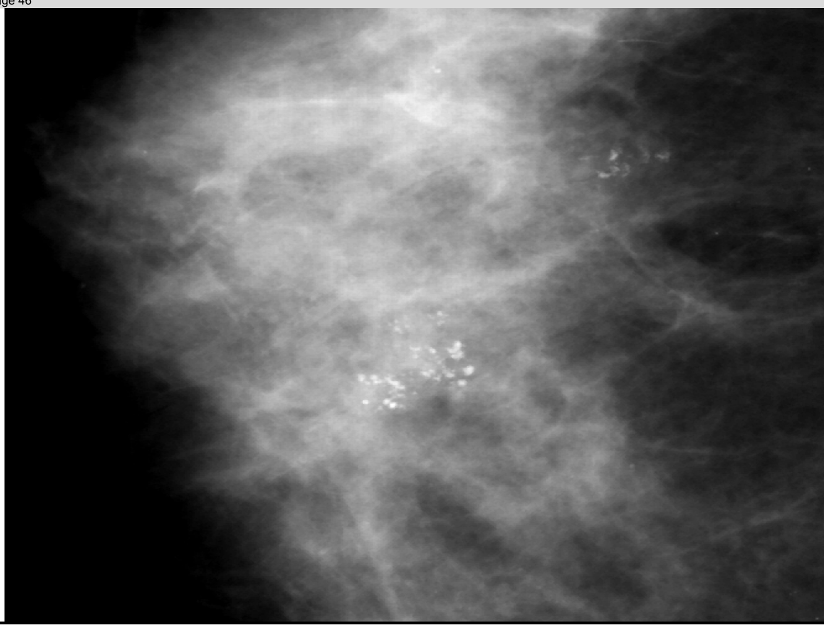
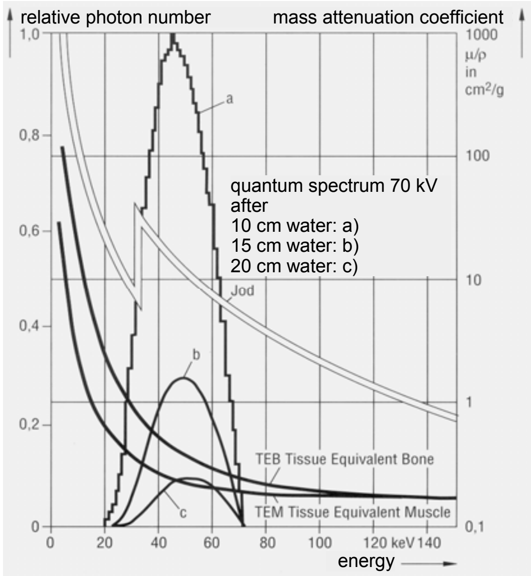
12/9/2008 | Page 46

Mammography Tumor Calcification



X-Ray Contrast Agent

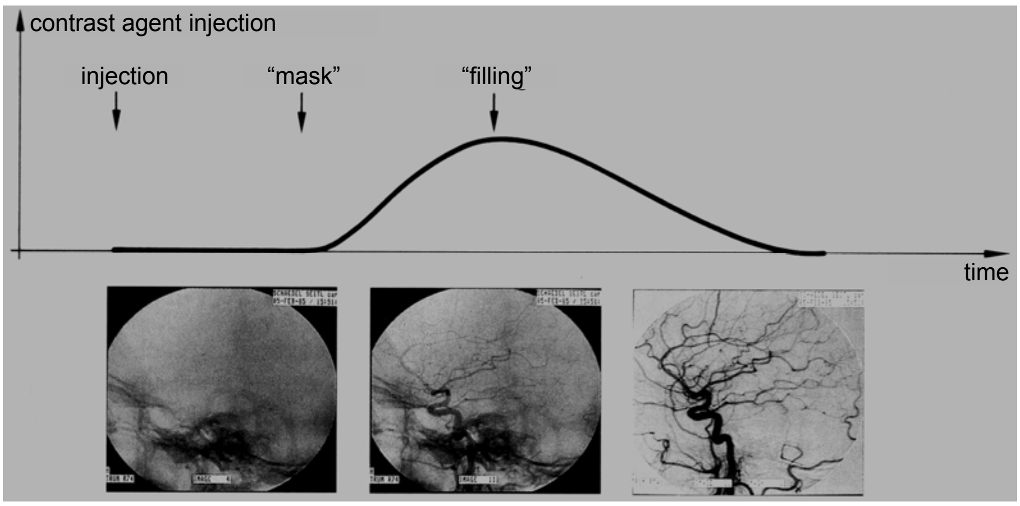


- X-ray negative contrast agent:
  air, CO_2, N_2O
- X-ray positive contrast agent:
  tri-iodine-benzoin-acid or similar (vessels)
  barium-sulfate $BaSO_4$ (gastrointestinal)
- tube voltage as a function of mass

- absorption coefficient for iodine with application for X-ray angiography of extremities
- optimal tube voltage for iodine contrast agent is about 63 kV

source: Morneburg. "Bildgebende Systeme für die medizinische Diagnostik" 1995

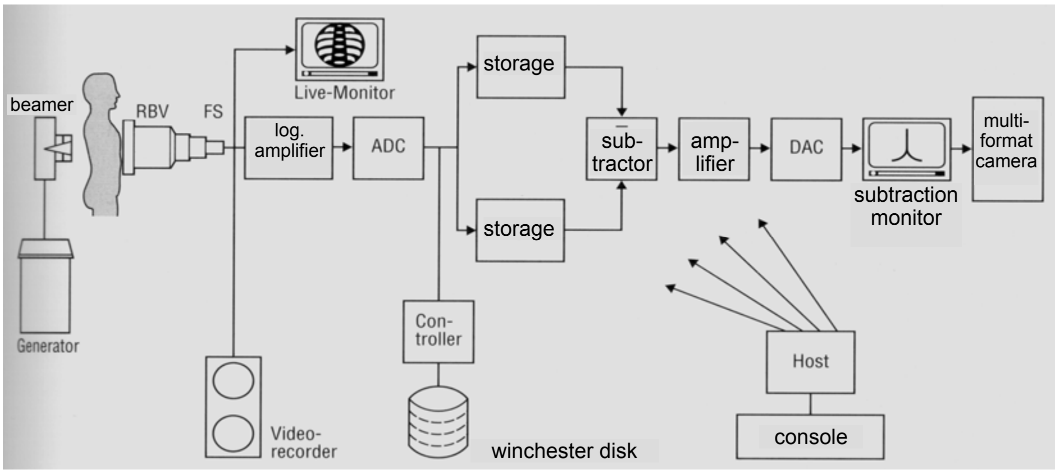
Contrast Agent Time Course



mask filling subtraction

source: Morneburg. "Bildgebende Systeme für die medizinische Diagnostik" 1995

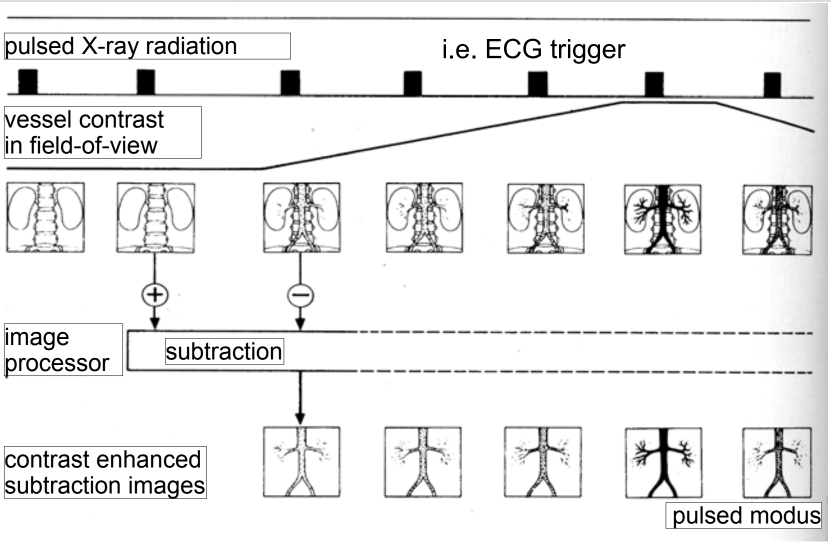
Digital Subtraction Angiography DSA



principle of DSA system

source: Morneburg. "Bildgebende Systeme für die medizinische Diagnostik" 1995

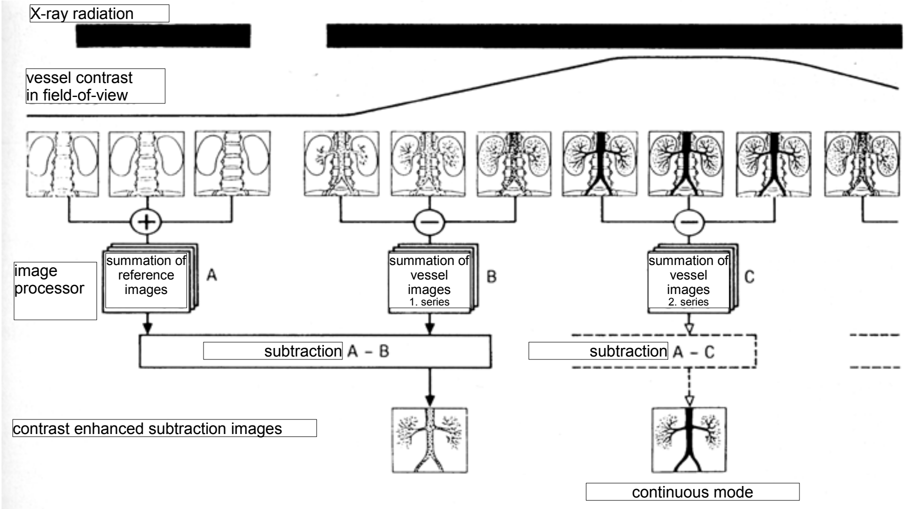
Pulsed DSA



maximum 6 images per second with high dose at single pulse !

source: Laubenberger and Laubenberger. „Technik der medizinischen Radiologie", Deutscher Ärzte-Verlag 1999

Continuous DSA



fixed mask which integrates some images, image artifacts due to vessel pulsation are compensated by different weightning at image addition

source: Laubenberger and Laubenberger. „Technik der medizinischen Radiologie“, Deutscher Ärzte-Verlag 1999

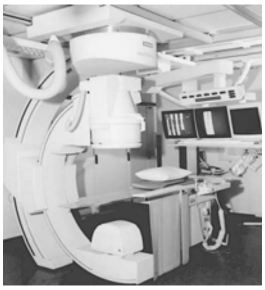
Dose Aspects

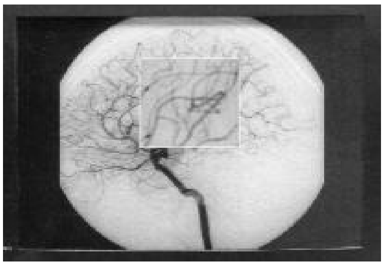
but:

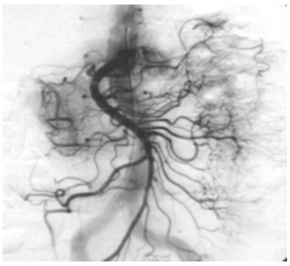
- radiation exposure is significant higher at continuous mode than at pulsed mode with short high-dose X-ray pulses

- same holds for stressing of the X-ray tube !

DSA System







DSA of arteria mesenterica superior (branch of abdominal artery)

Angiography: Examples

lung angiography

pelvis angiography

hand angiography (DSA)