RADIOLOGY RESEARCH

Henry Ford

Health System

AAPM Summer School 2004

Digital Radiography Performance
MTF, NPS, DQE, and Diagnostic Value

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Intro:
5 steps

Radiographic Image Formation

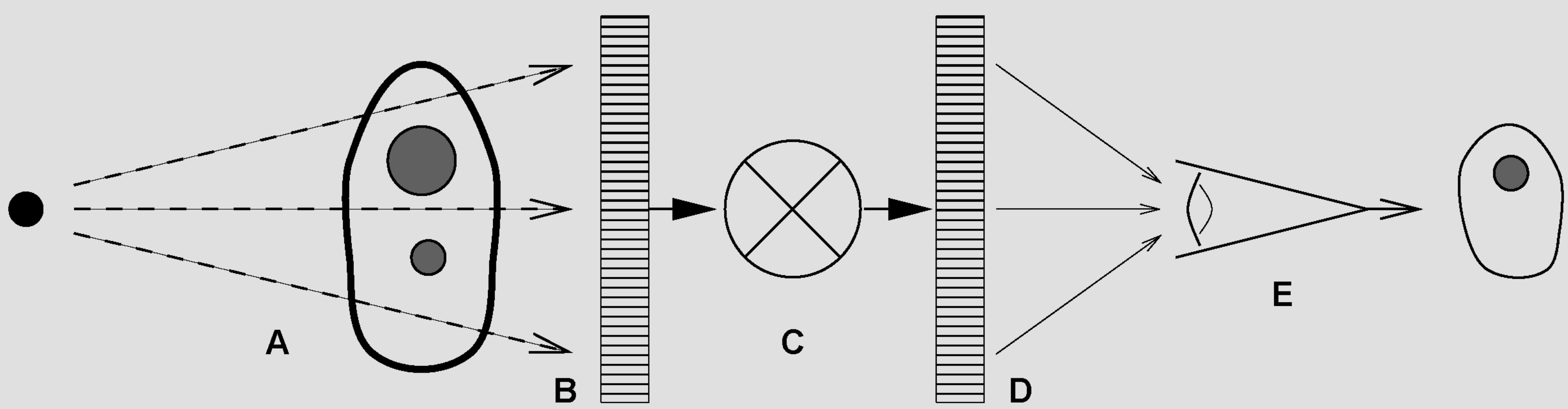
A. Object: Subject contrast and scatter …

B. Detector: Modulation and noise transfer, scatter…

C. Processing: grayscale, equalization, edge restoration….

D. Display: Modulation, noise, grayscale, color …

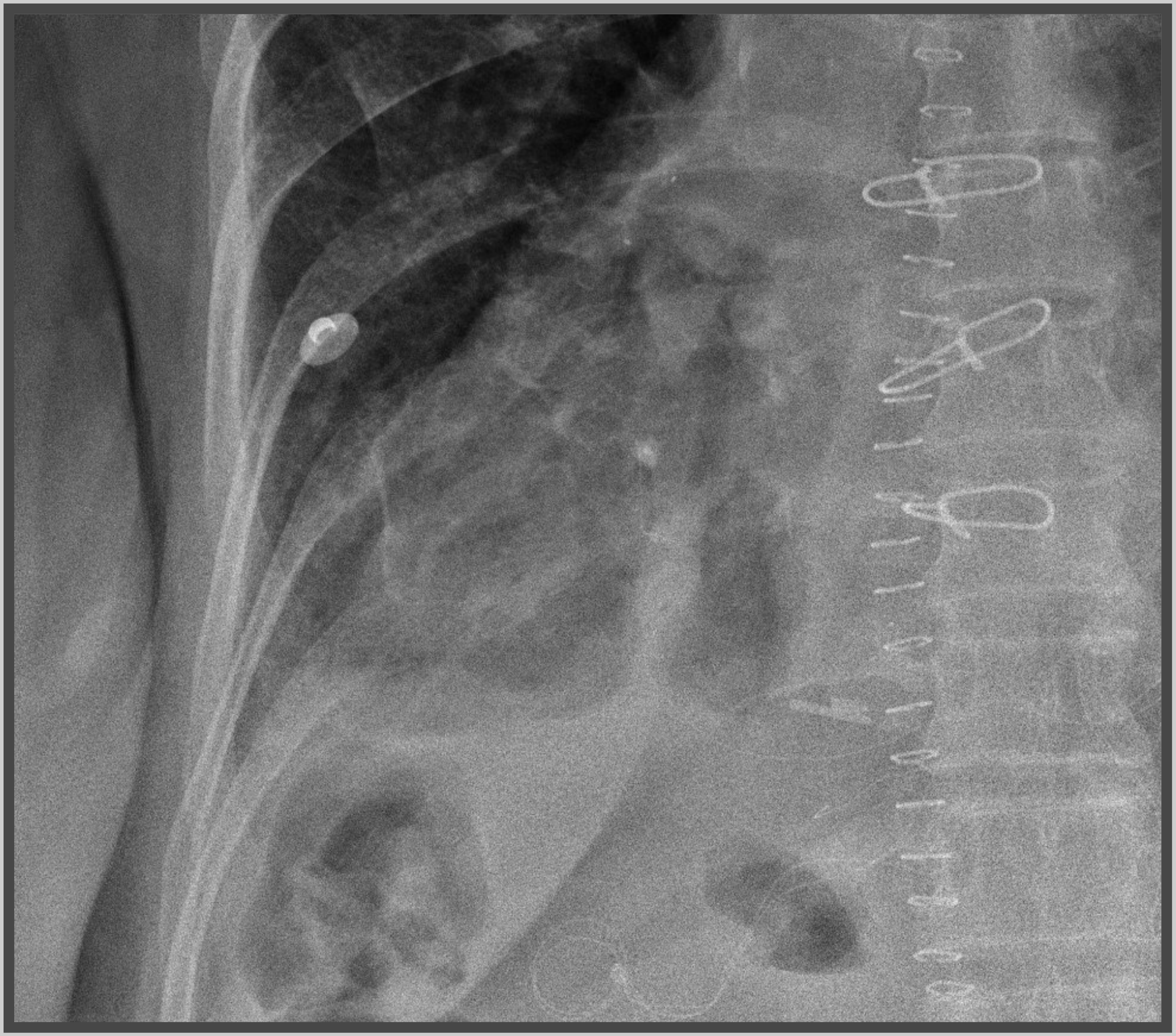
E. Vision: Visual response, recognition ..



Good detector resolution permits the recording of detailed object structures.

Intro:
Detail

Intro:
Noise



Good detector efficiency minimizes the appearance of quantum mottle.

Intro:
Aims

Learning Objectives

This presentation describes methods to assess the resolving capability (MTF) and quantum noise (NEQ) of digital radiography detectors.

- Signal transfer (MTF):
  Evaluation of modulated signal transfer using the edge response method.

- Noise transfer (NPS):
  Evaluation of image noise in relation to exposure using 2D fourier transform methods .

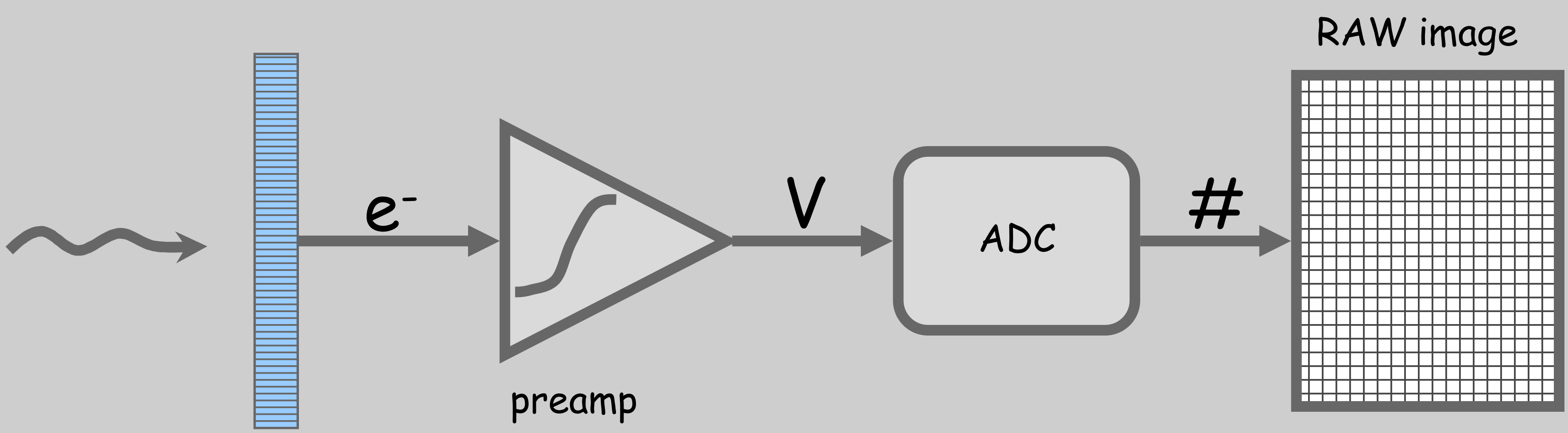
- Signal to noise (DQE):
  Evaluation of overall detector efficiency in terms of the signal to noise transfer.

Part
1

1 - Export Unprocessed Data

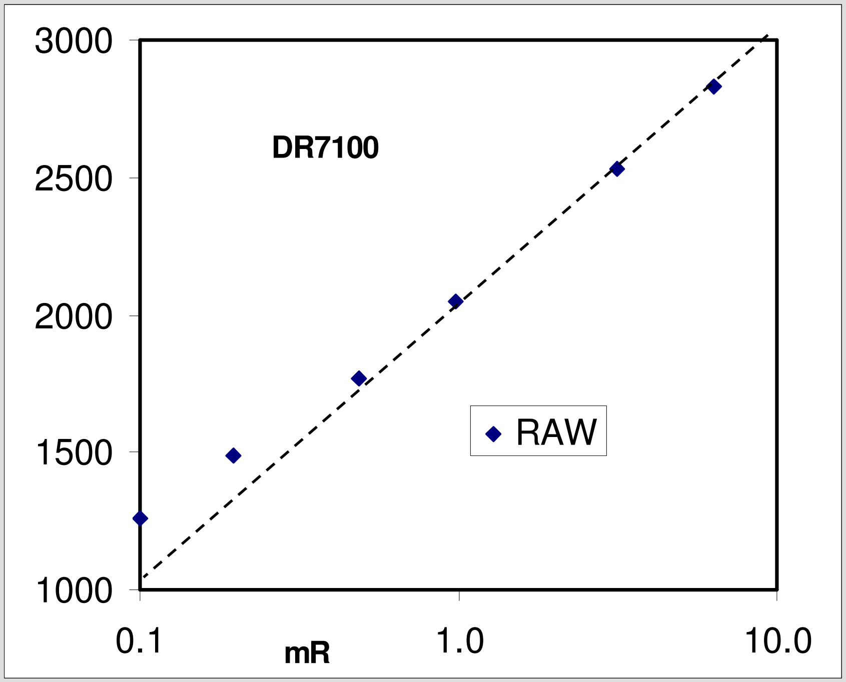
Raw Image Data

- For CR and DR systems, radiation energy deposited in the detector is converted to electrical charge.
- Preamplifier circuits then convert this to a voltage which is digitized using analog to voltage converter (ADC) to produce RAW image values.
- Non-linear preamplifiers are used so that the raw image values represent a wide range of exposures. Alternatively a non-linear input LUT can transform the ADC values.



Raw Log format

- Most RAW image values are proportional to the log of the exposure incident on the detector.
- Small relative changes in exposure due to small tissue structures produce a fixed change in RAW values regardless of the total tissue transmission.



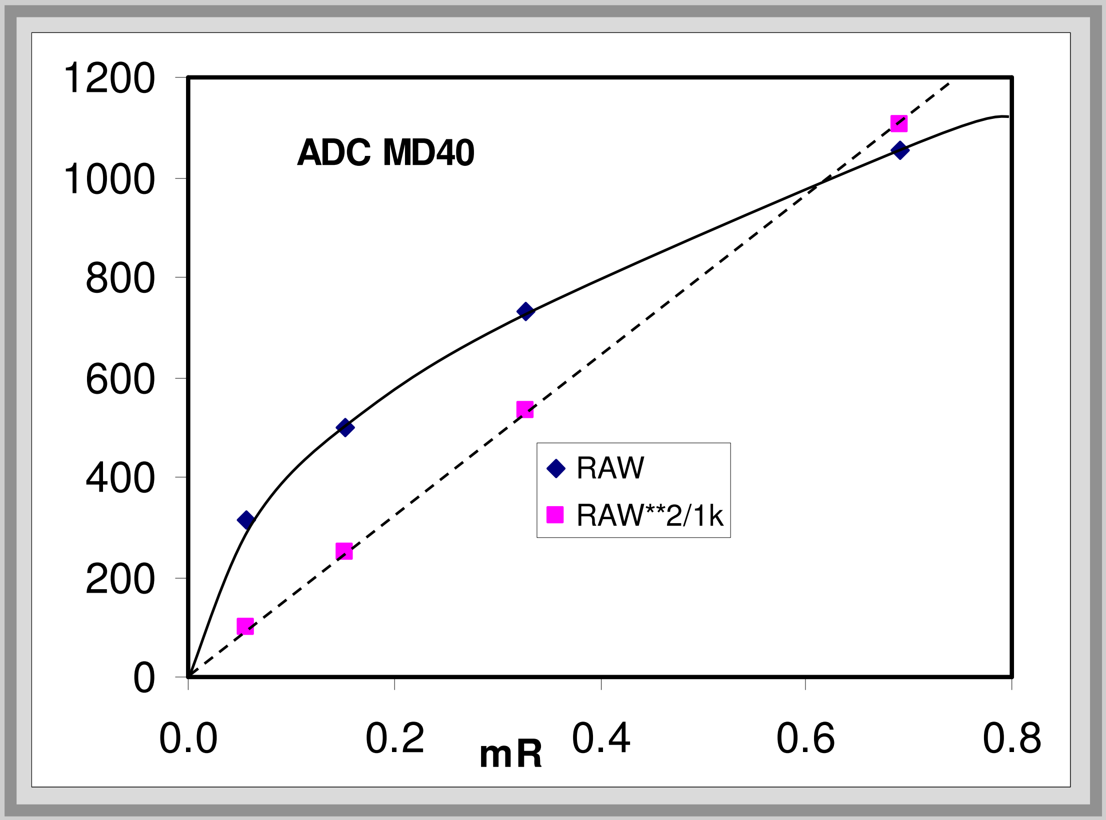
For RAW values stored as a 12 bit number (0 – 4095), a convenient format has a RAW change of 1000 for every factor of 10 change in exposure.

$$RAW = 1000 \log_{10}(mR) + 2000$$

Raw
$mR^{1/2}$

- One major manufacture uses preamplifiers that make the RAW values proportional to the square root of exposure.
- The relative noise of the RAW values is constant for all incident exposures, however the tissue contrast is not.

RAW = 1250 $mR^{1/2}$



Linear Systems Analysis

Prerequisites for Linear Systems Analysis

- **Linearity:**

  The signal response in relation to the incident x-ray intensity is linear or can be linearized.

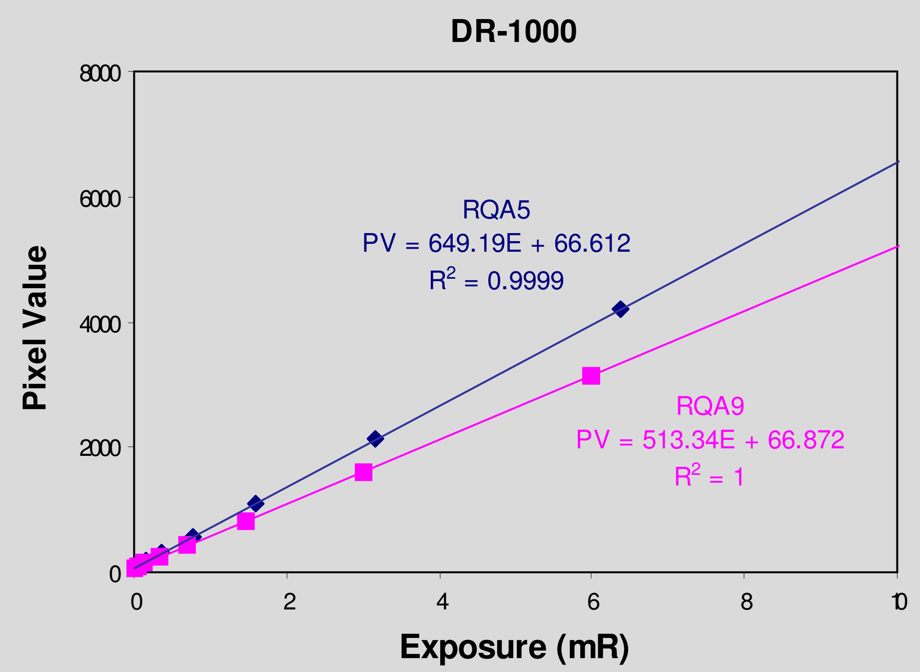
- **Stationarity:**

  The spatial response of the detector does not vary for x-rays incident at different positions.

For all performance measurements, data must be acquired and transformed to image values that are linear with exposure and have a zero offset.

A system that provides 12 bit log signal data over a range of 10^4 can be transformed by;

$$I(j,k) = 10^{Q(j,k)/1024}$$

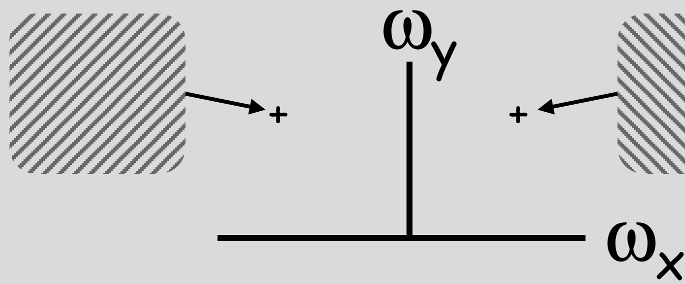


Part
2

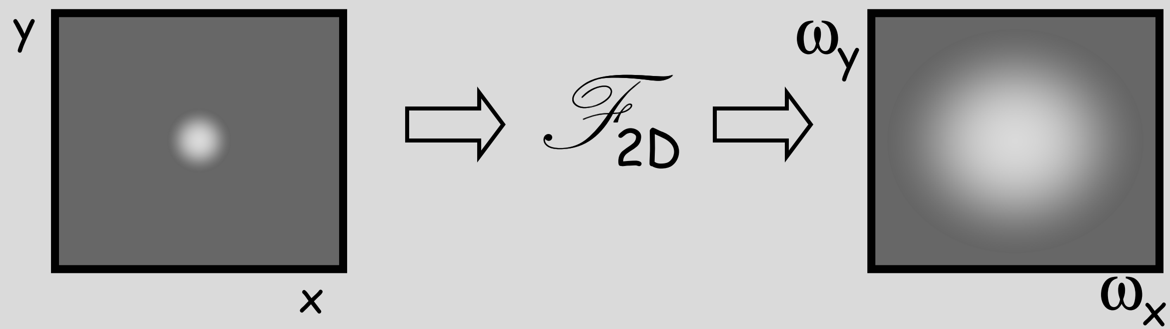
2 - Modulation Transfer

MTF:
2D MTF
and PSF

- The 2D MTF is defined as the modulation transfer for 2D sinusoidal patterns of varying spatial frequency and orientation.



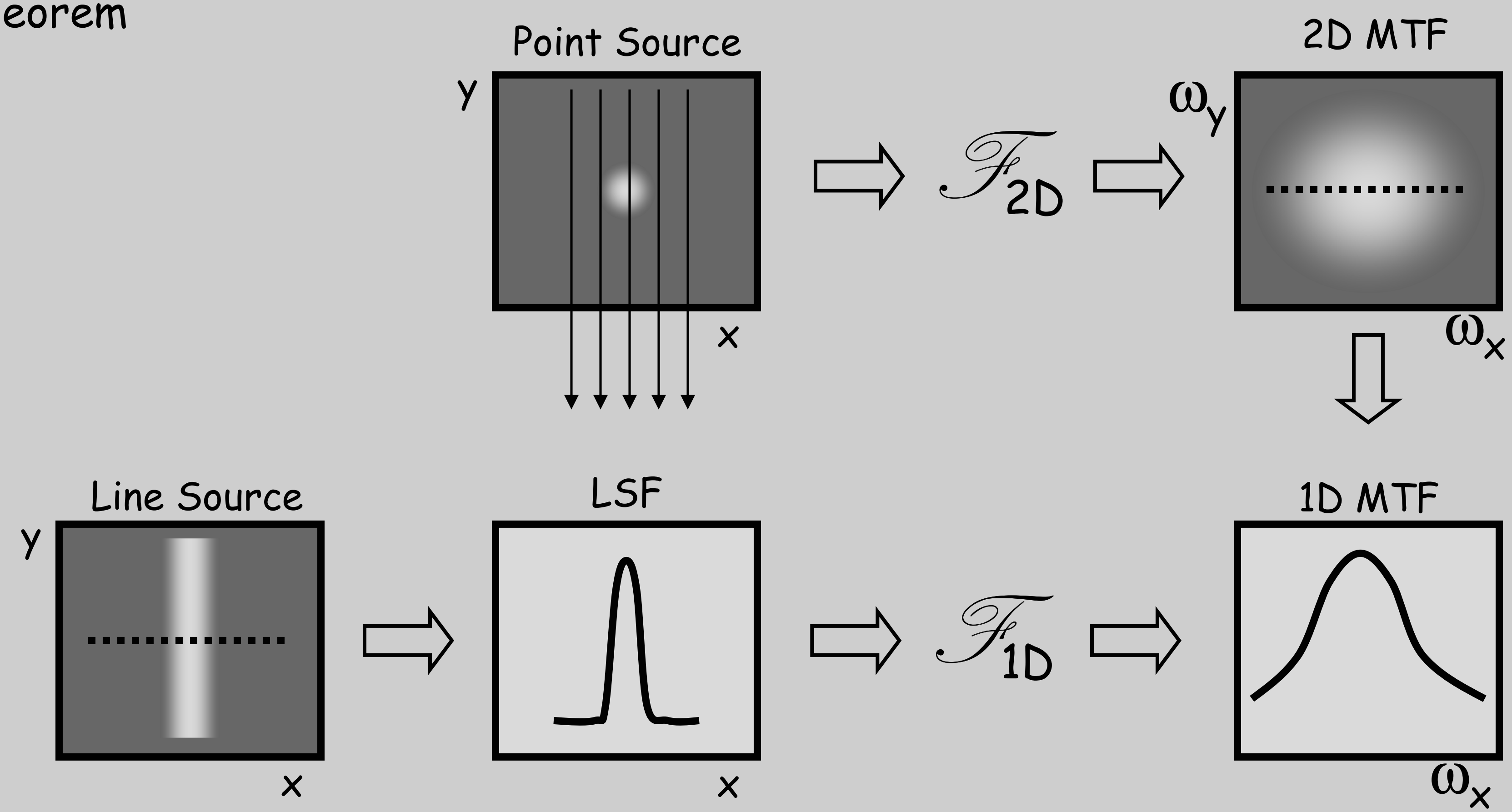
- The 2D MTF may be computed as the magnitude of the 2D Fourier transform of the systems point spread function, PSF.



- The PSF describes the systems response to x-rays normally incident to the detector at a specific point.

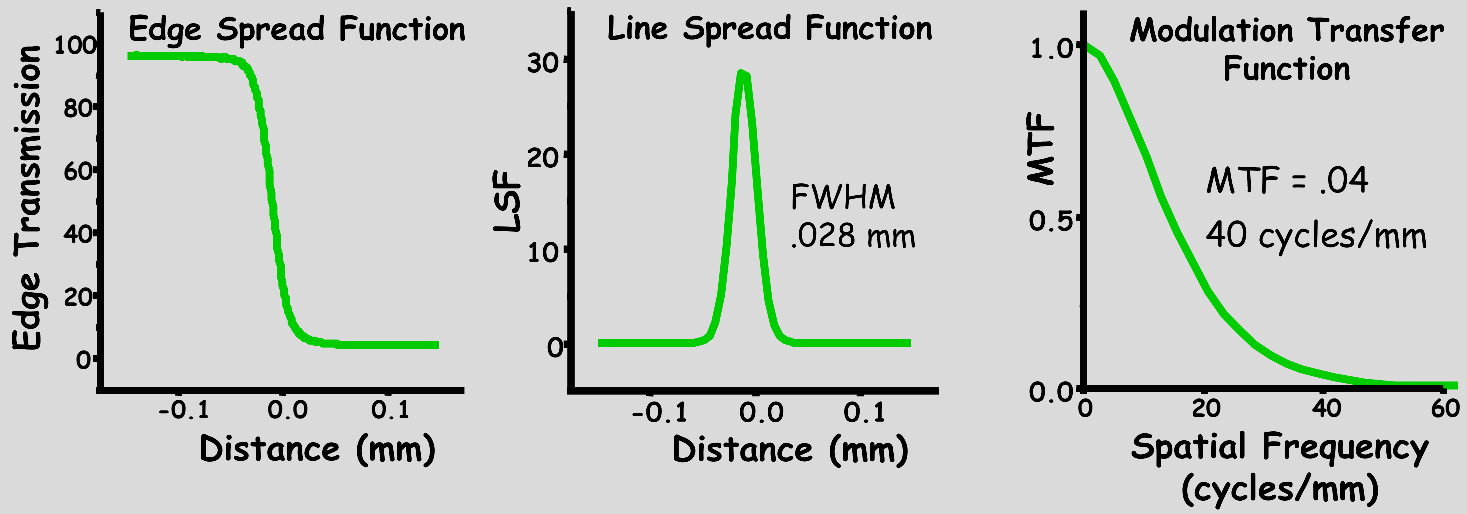
MTF: Central Ordinate Theorem

- The projection of a PSF is equal to the LSF for a line source oriented parallel to the projection direction.
- The 1D MTF is equal to the values of the 2D MTF along a line through the origin and perpendicular to the line source



•K. Rossman, Radiology, Vol 93, 1964

MTF:
Edge
Spread
Function



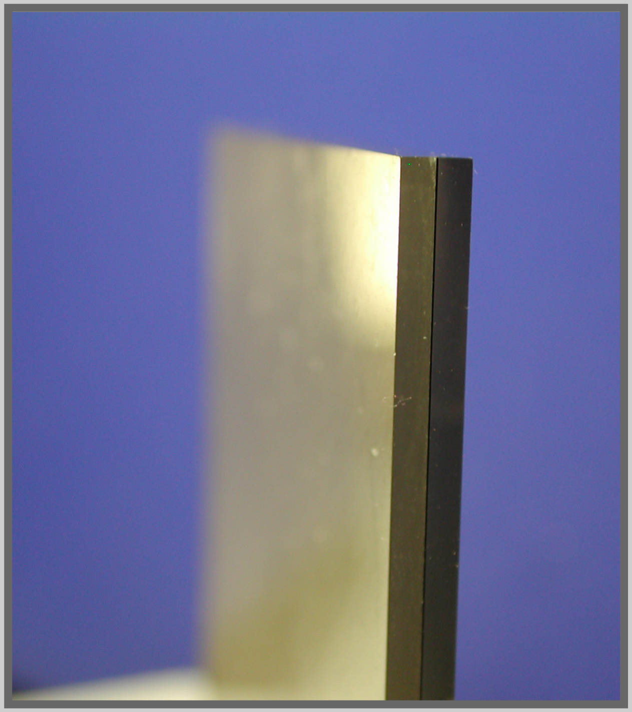
- The edge spread function, ESF, represents the values of a 2D edge response image along a line perpendicular to the edge.
- The ESF is equal to a definite integral of differential lines sources over the limits of the transmissive space.
- The LSF is simply deduced as the derivative of the ESF

•Samei, Med. Phys., V 25, N 1, 1998

MTF:
Edge
Phantom

Edge Test Devices

- Presently, edge test devices are not sold.
- However, they can be simply fabricated.
  - Laminate a metal foil between two lucite plates.
  - Precision mill to the final shape.
  - Lap all edges to obtain a straight edge;
    (use a sub-micron grit and flat lapping stone)



•Edge Test Device

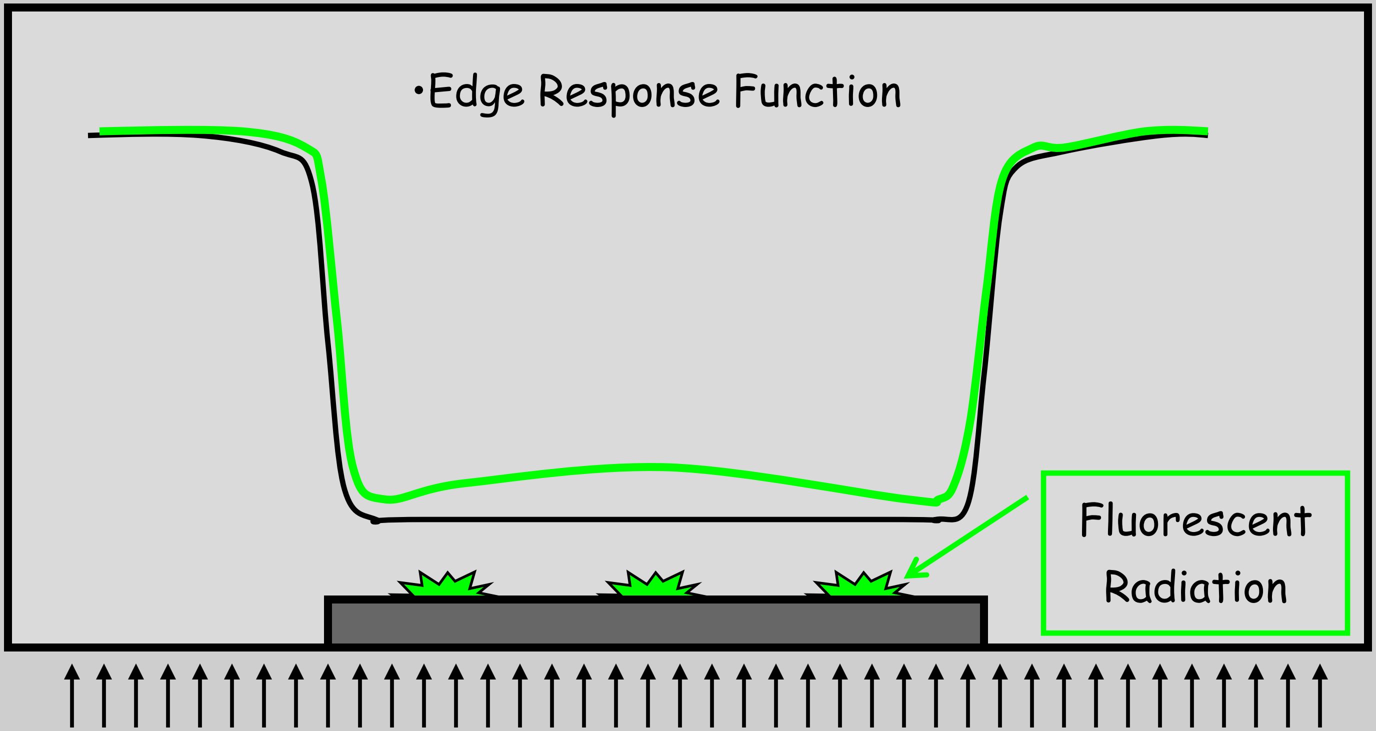
5 x 5 cm, 0.1-mm-thick polished Platinum-Iridium foil laminated between two 1-mm slabs of Lucite

•Image Acquisition

Device placement at the center of the field of view

Acquisition at ~ 4-6 mR (w/o grid)

MTF:
Edge
Phantom
Material

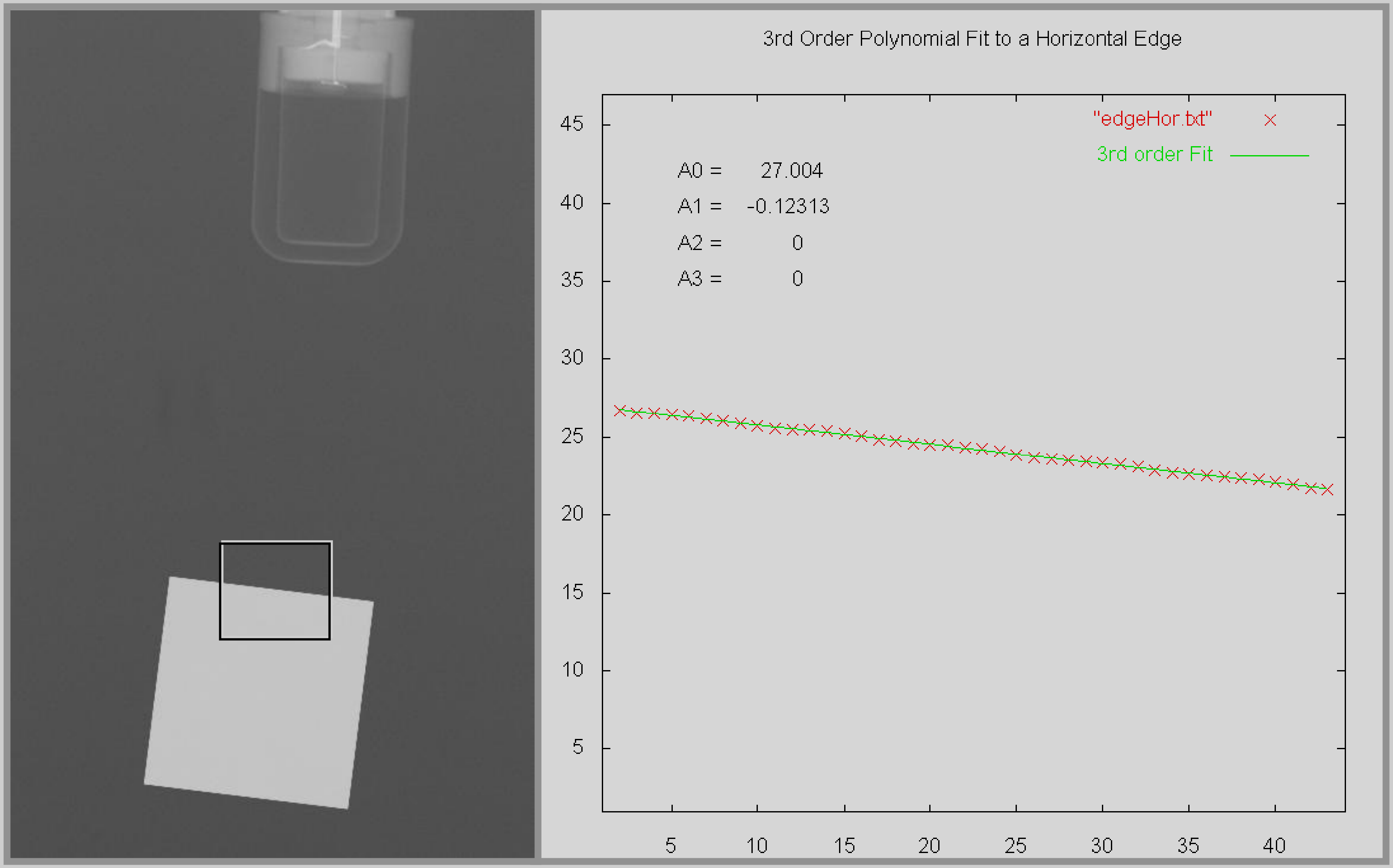


- If the kVp of the x-ray beam is above the binding energy of the edge material, fluorescent radiation can cause low spatial frequency distortion of the edge response.
- Edges should be made of very high Z material to minimize this effect;

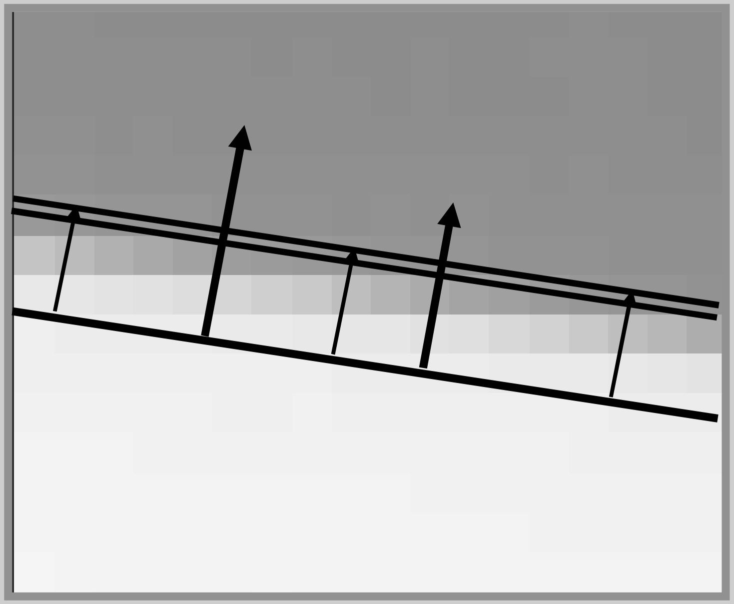
  Lead, Platinum-Iridium, depleted Uranium
- For low energies, a Brass-Aluminum laminate would be appropriate

- Edge positions are estimated for each column using a numerical derivative of the column data.
- A polynomial fit is performed on the edge estimates for all column in a defined region of interest

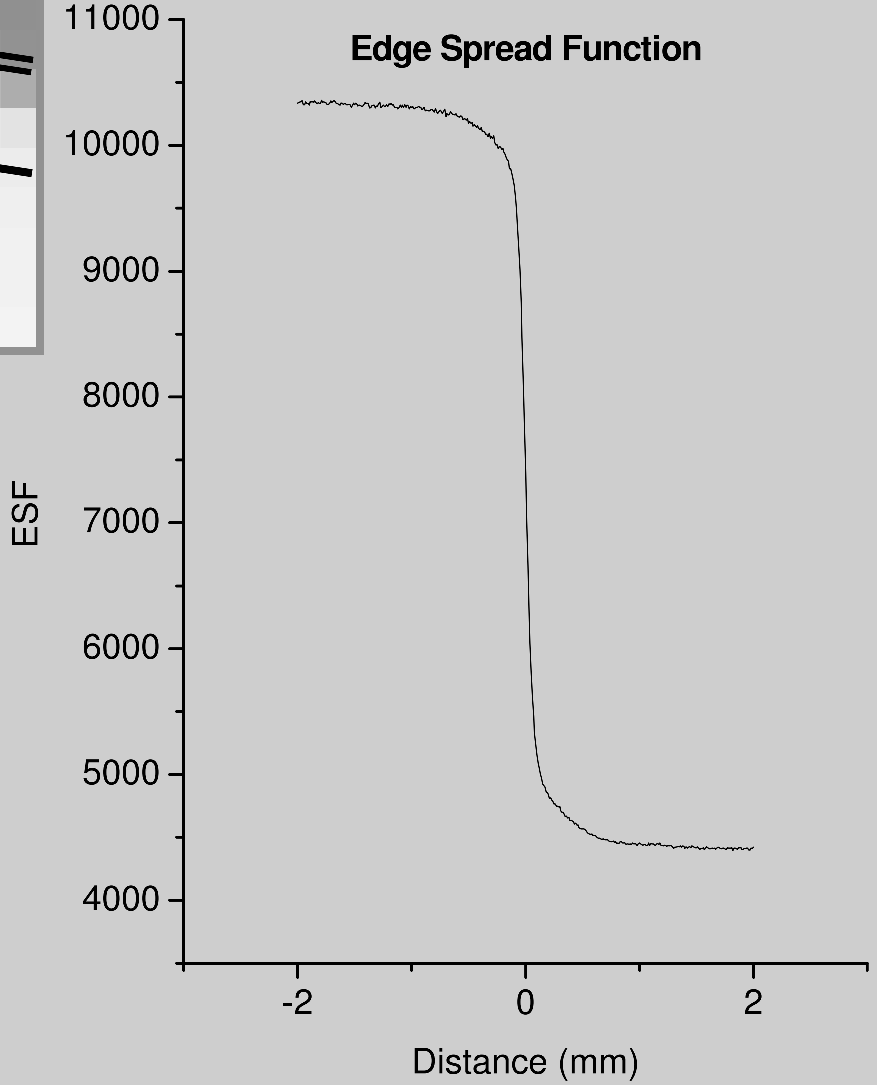
MTF:
Edge
Fit



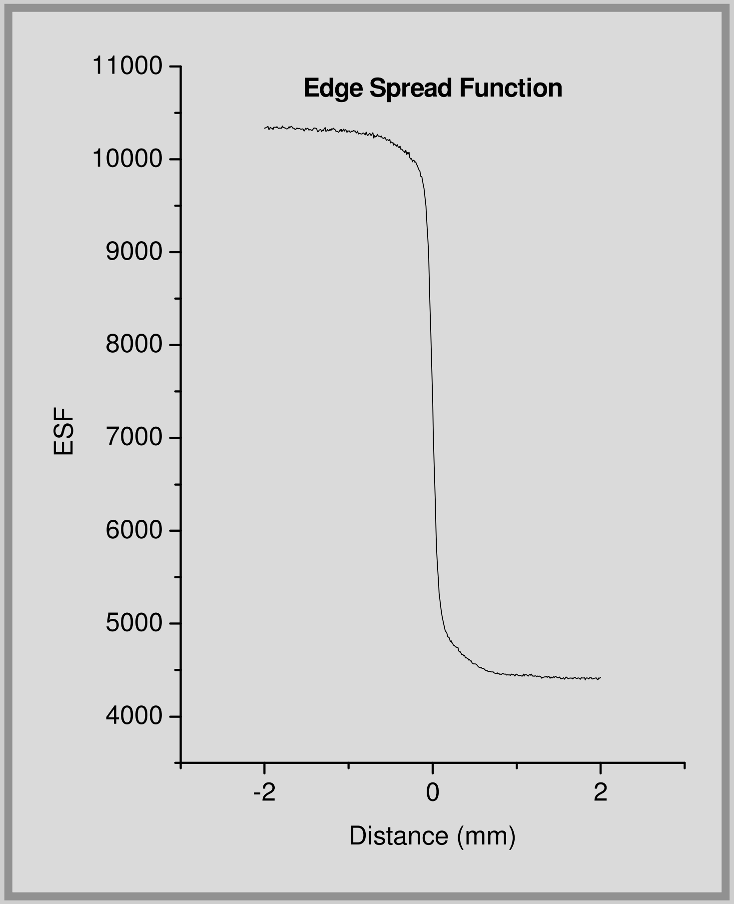
MTF: Distance Binning

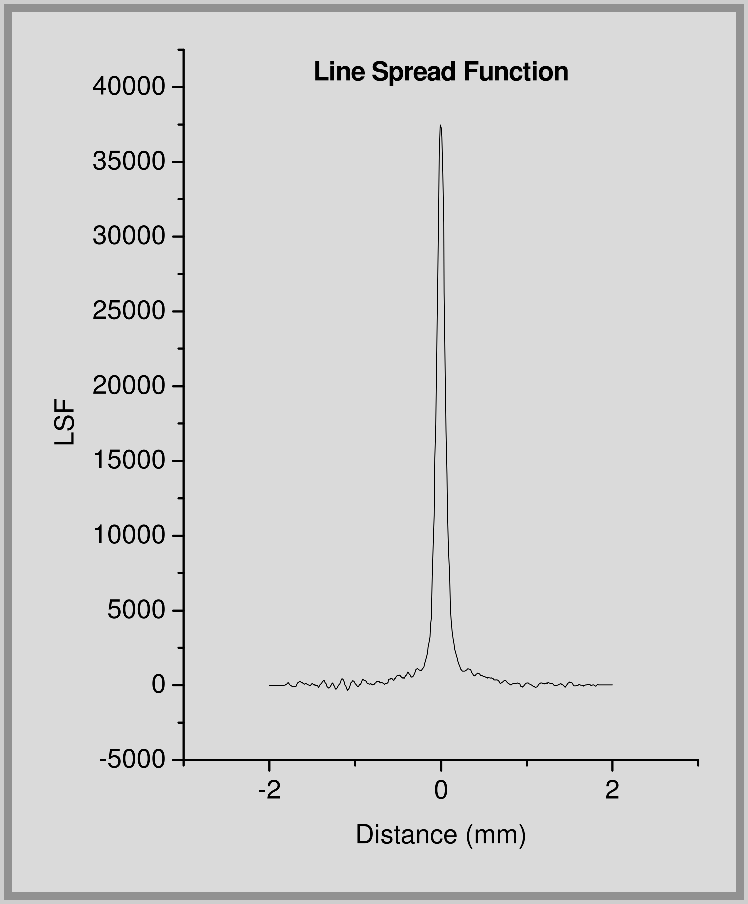


- For each pixel in the ROI;
  - The closest distance to the edge is determined.
  - This value is converted to an integer index with a unit length equal to 1/10 of the pixel size.
  - The pixel value is added to the array element of this index.
  - A counter array element of this index is incremented by 1.
- The average value of the ESF is then computed from the accumulated values and numbers in these arrays.



MTF: ESF to LSF

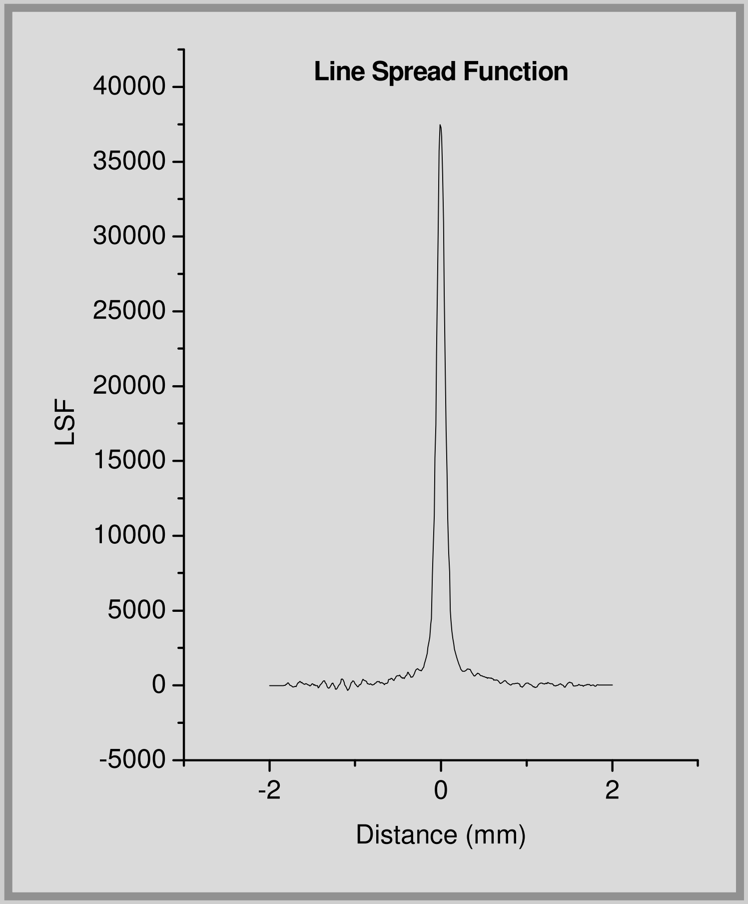


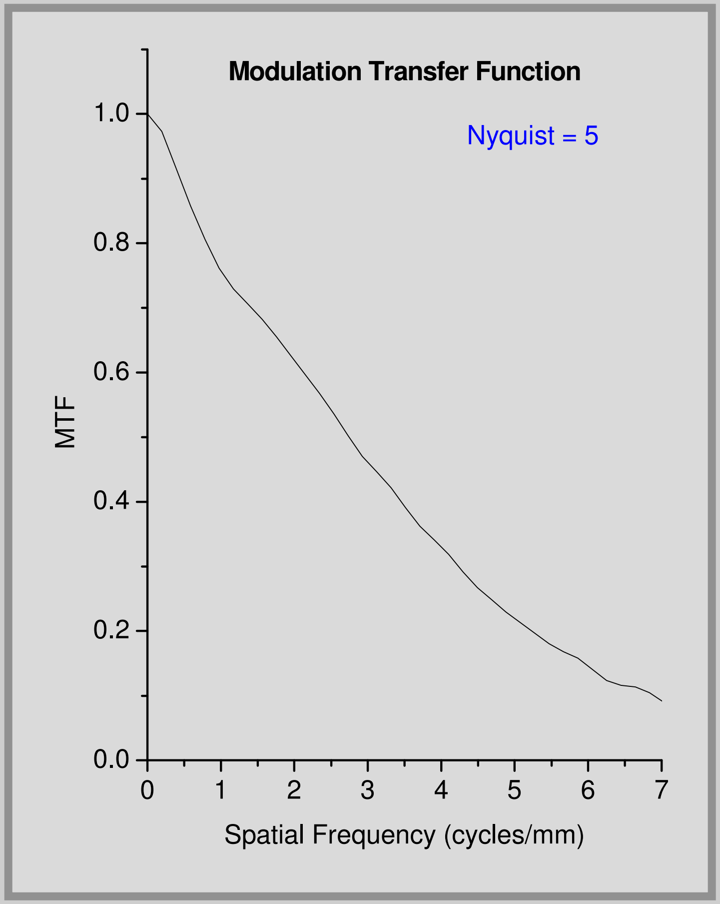


ESF to LSF transformation

- The ESF is smoothed in regions of low slope using a moving polynomial fit
- The ESF is numerically differentiated to deduce the LSF
- The baseline of the LSF is corrected using the ends of the data

MTF:
LSF
to
MTF



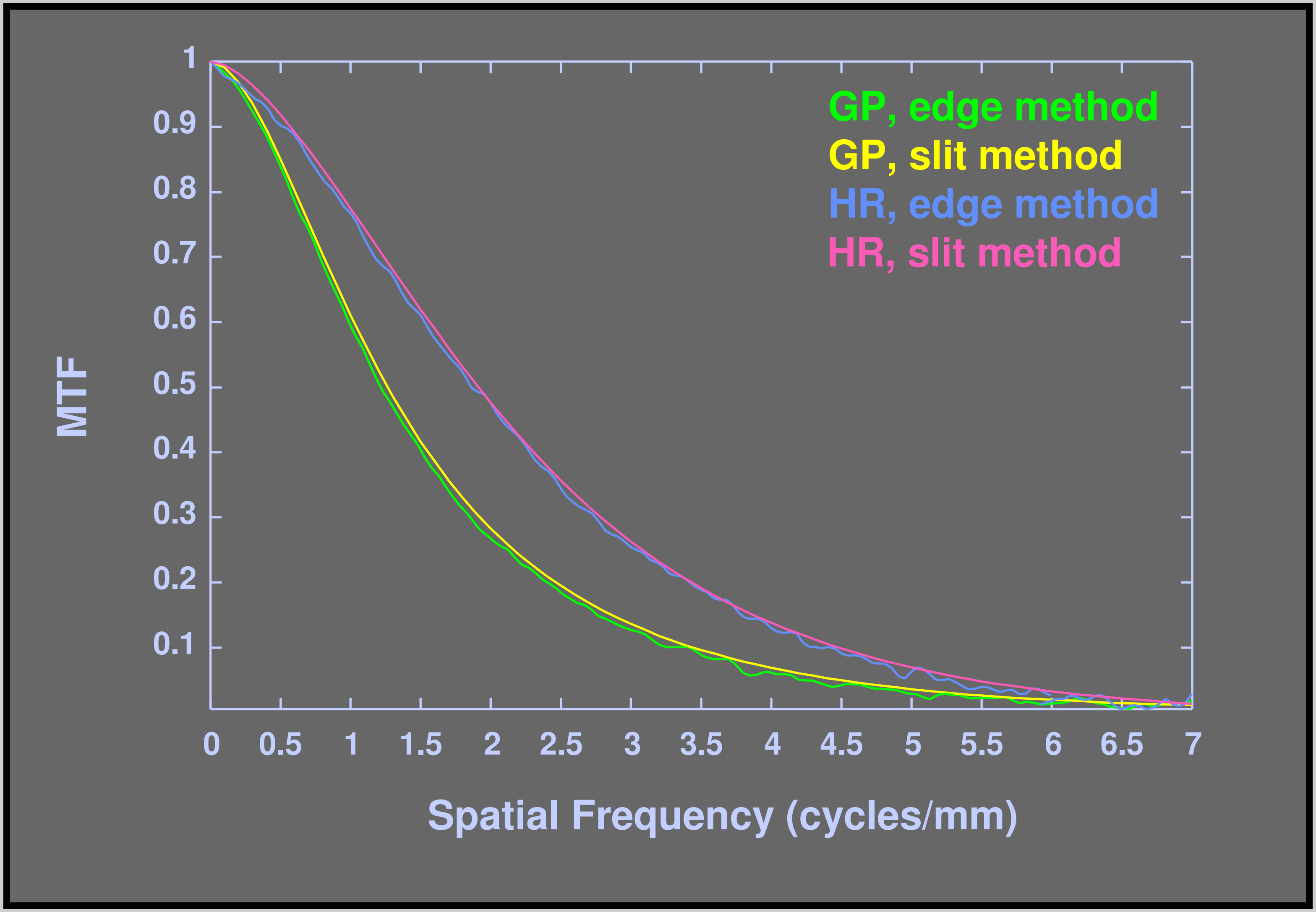


LSF to MTF transformation

- The LSF is multiplied by a Hamming window function.
- The magnitude of the Fourier transform of the LSF is computed.
- The MTF is determined by normalizing the magnitude to 1.0 at a spatial frequency of 0.0. Since the transform is symmetric, only values at positive spatial frequency are used.

- Edge and Slit Methods Comparison
- KESPR, GP-25 and HR, 0.1 mm pixel size, 90 kVp, PSC

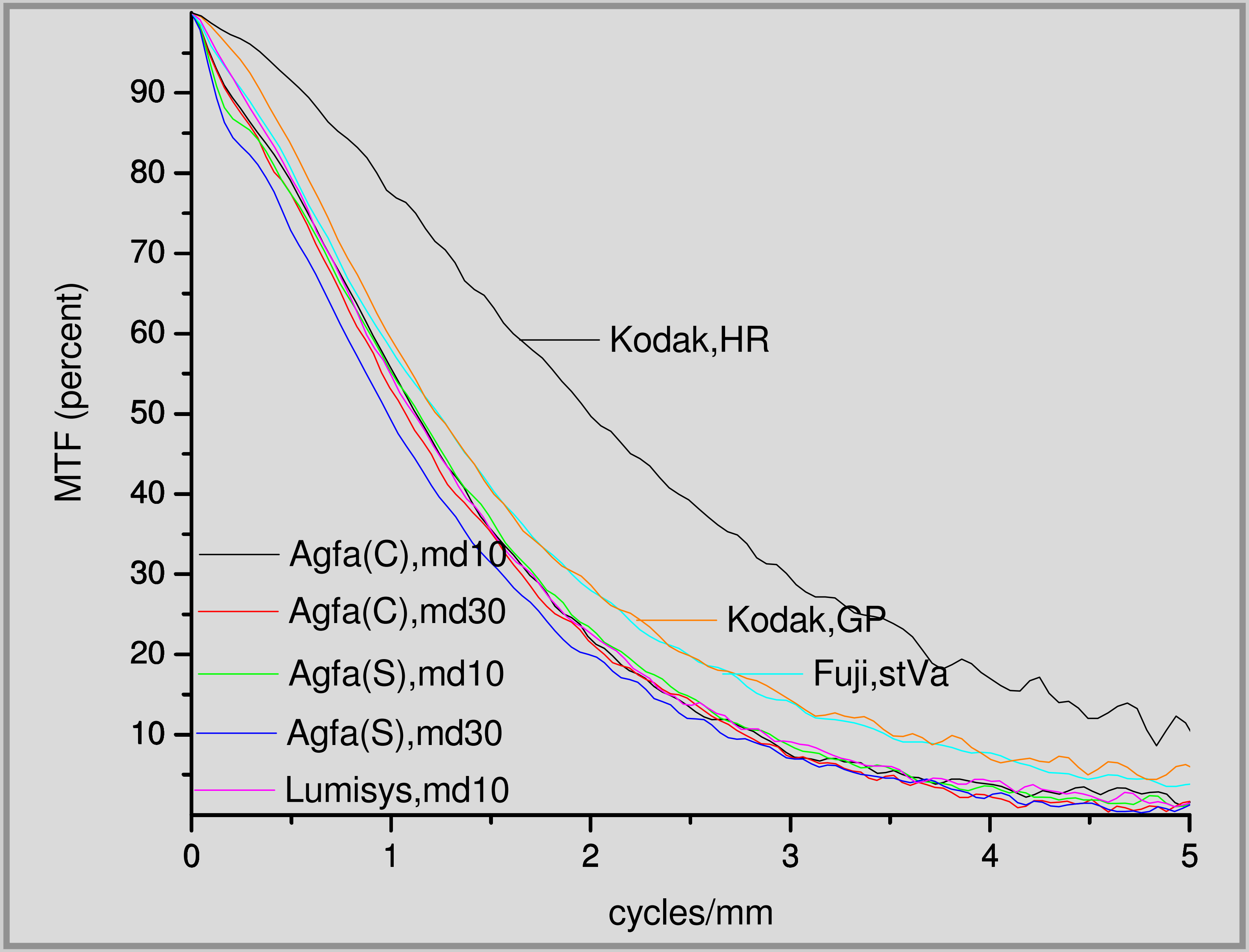
MTF:
Edge
vs
Slit



Samei
Med. Phys.
1998

MTF of 8 CR reader-screen systems

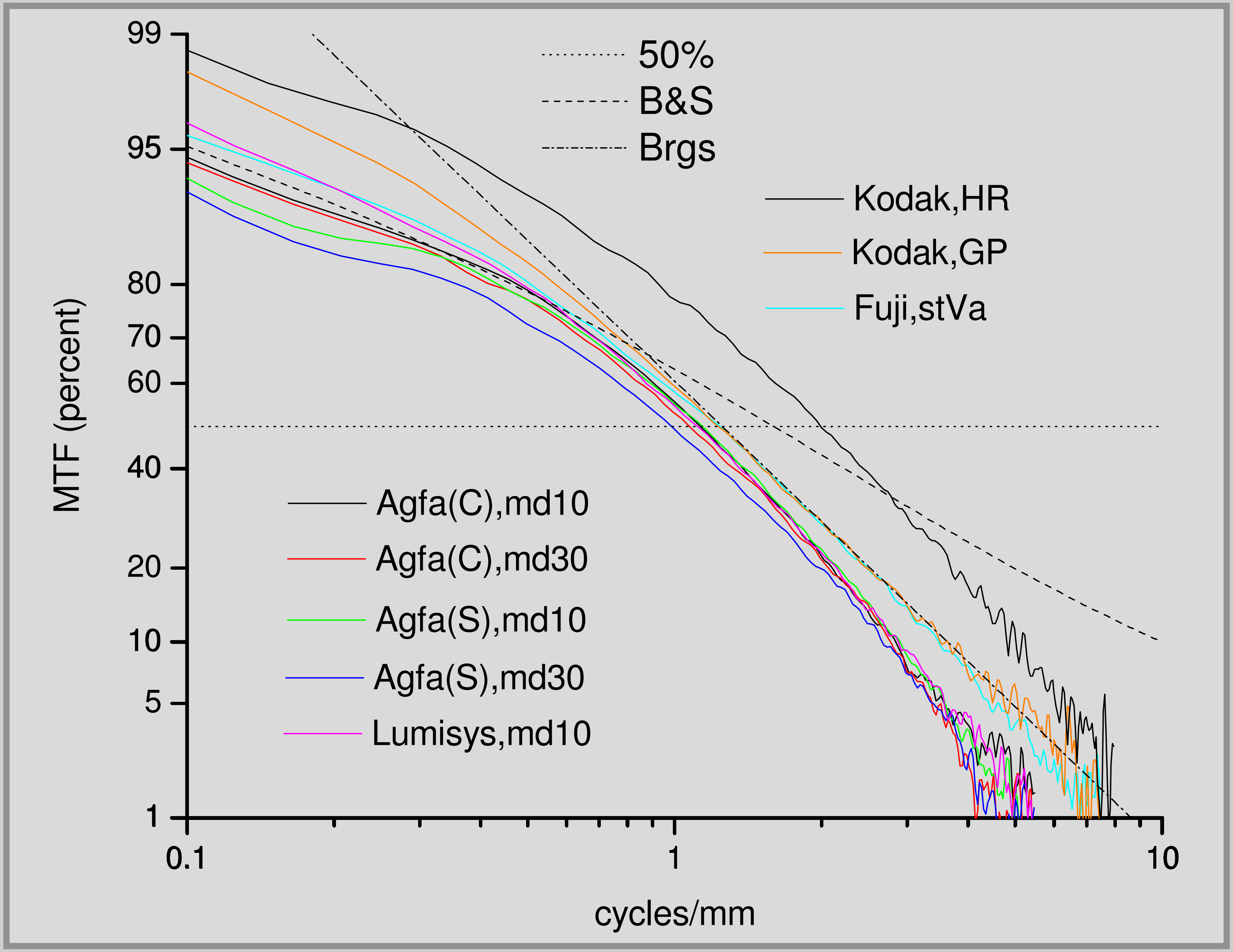
MTF:
CR
Systems



Samei & Flynn, Med.Phys. 2002

MTF of 8 CR reader-screen systems

MTF:
Probit
Plots



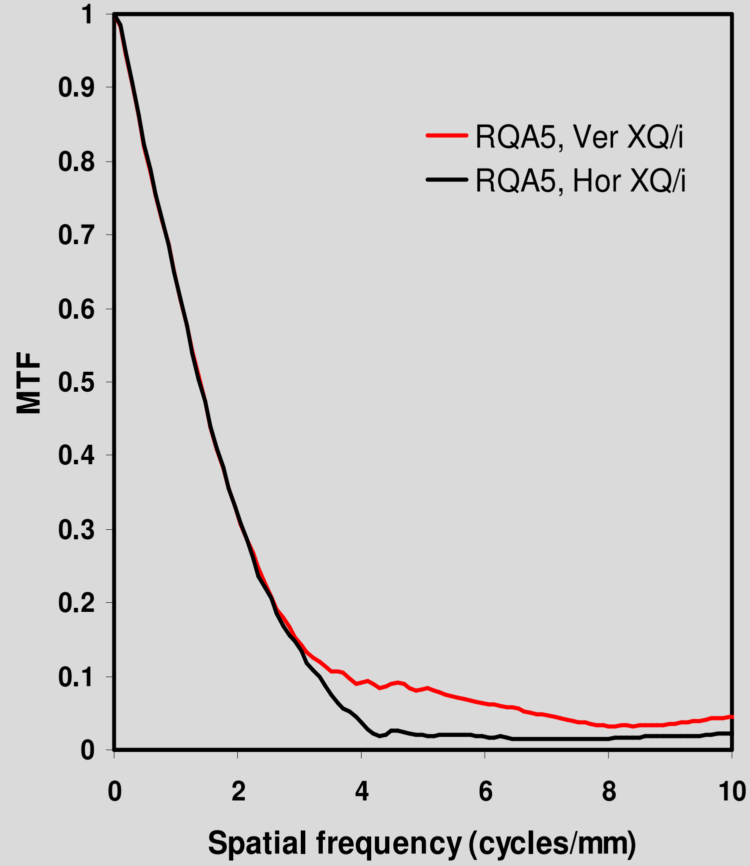
Samei & Flynn, Med.Phys. 2002

MTF: DR Direct and Inirect

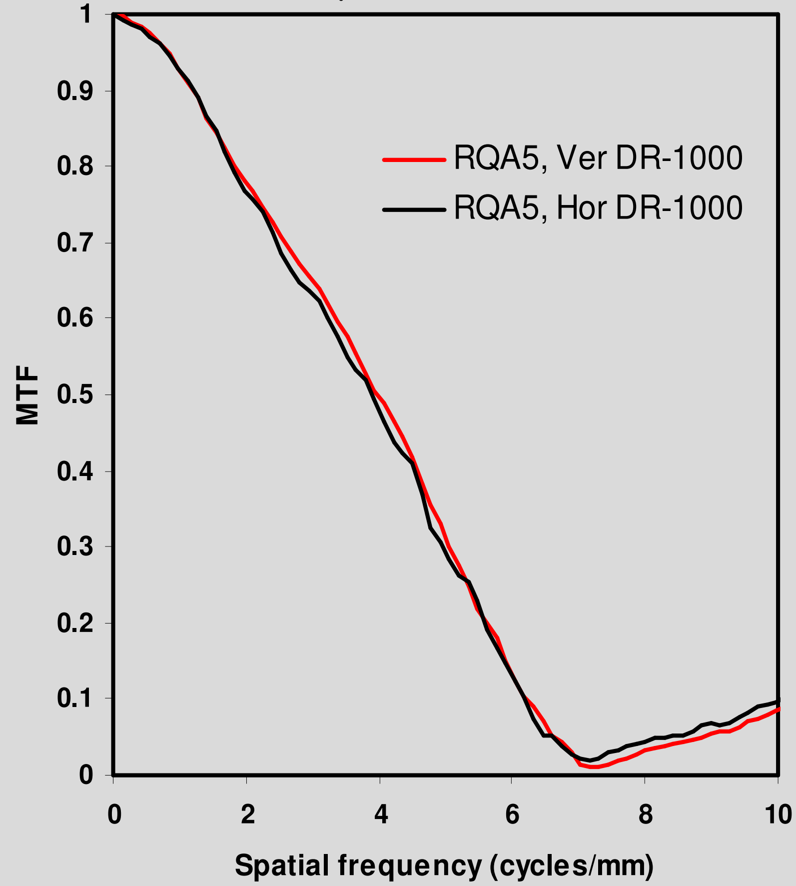
MTF for Digital Flat Panel Radiography

•Samei & Flynn, Med.Phys. 2003

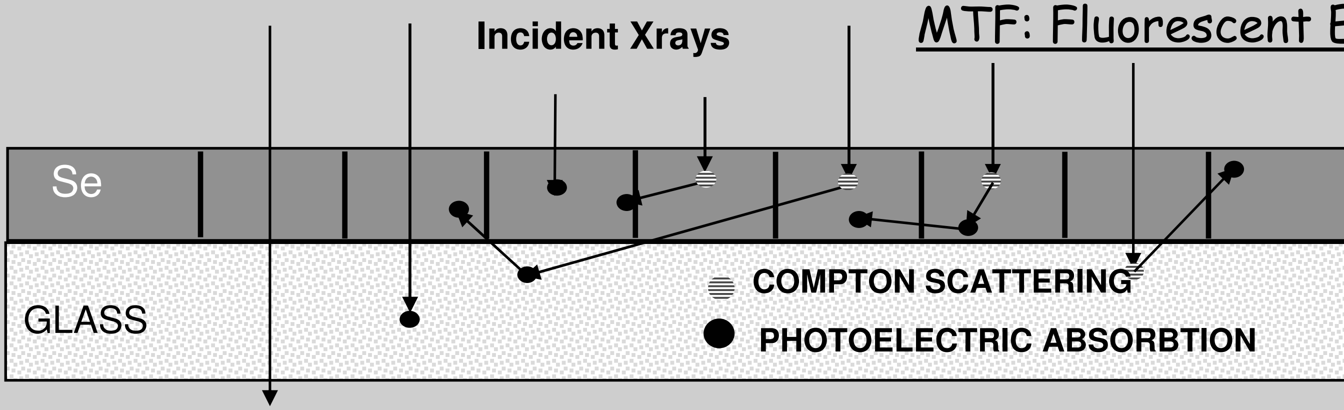
•CsI indirect detection

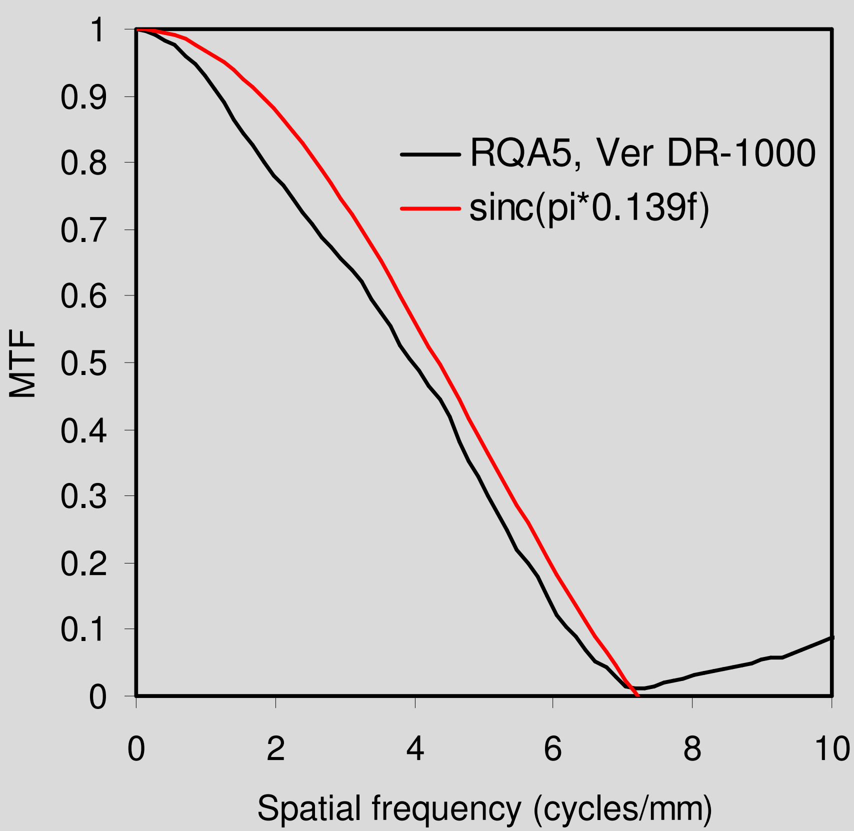


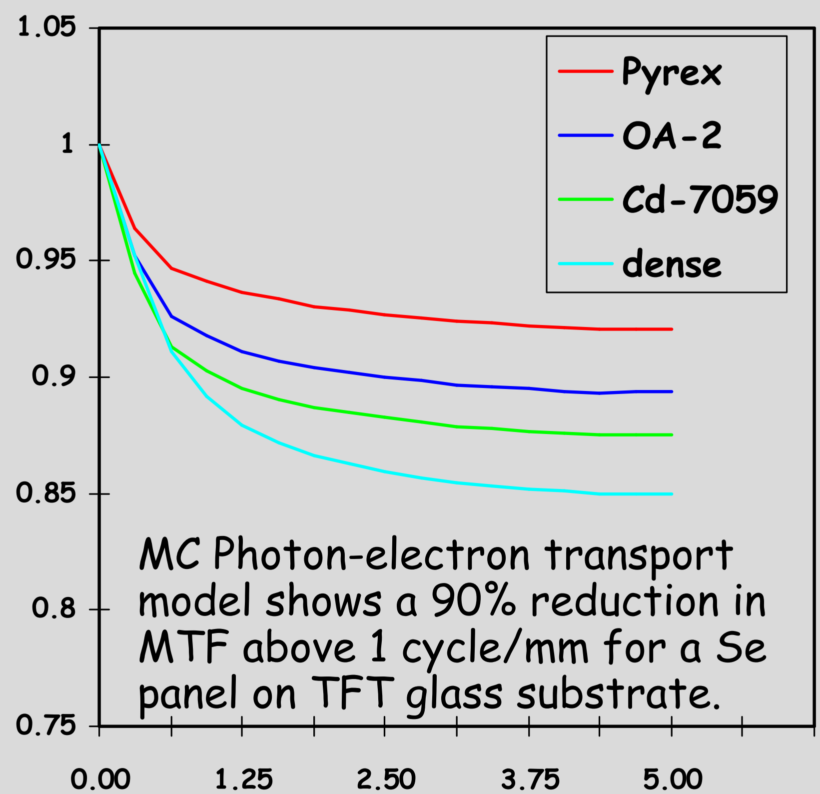
•Selenium direct detection



MTF: Fluorescent Effects







MC Photon-electron transport model shows a 90% reduction in MTF above 1 cycle/mm for a Se panel on TFT glass substrate.

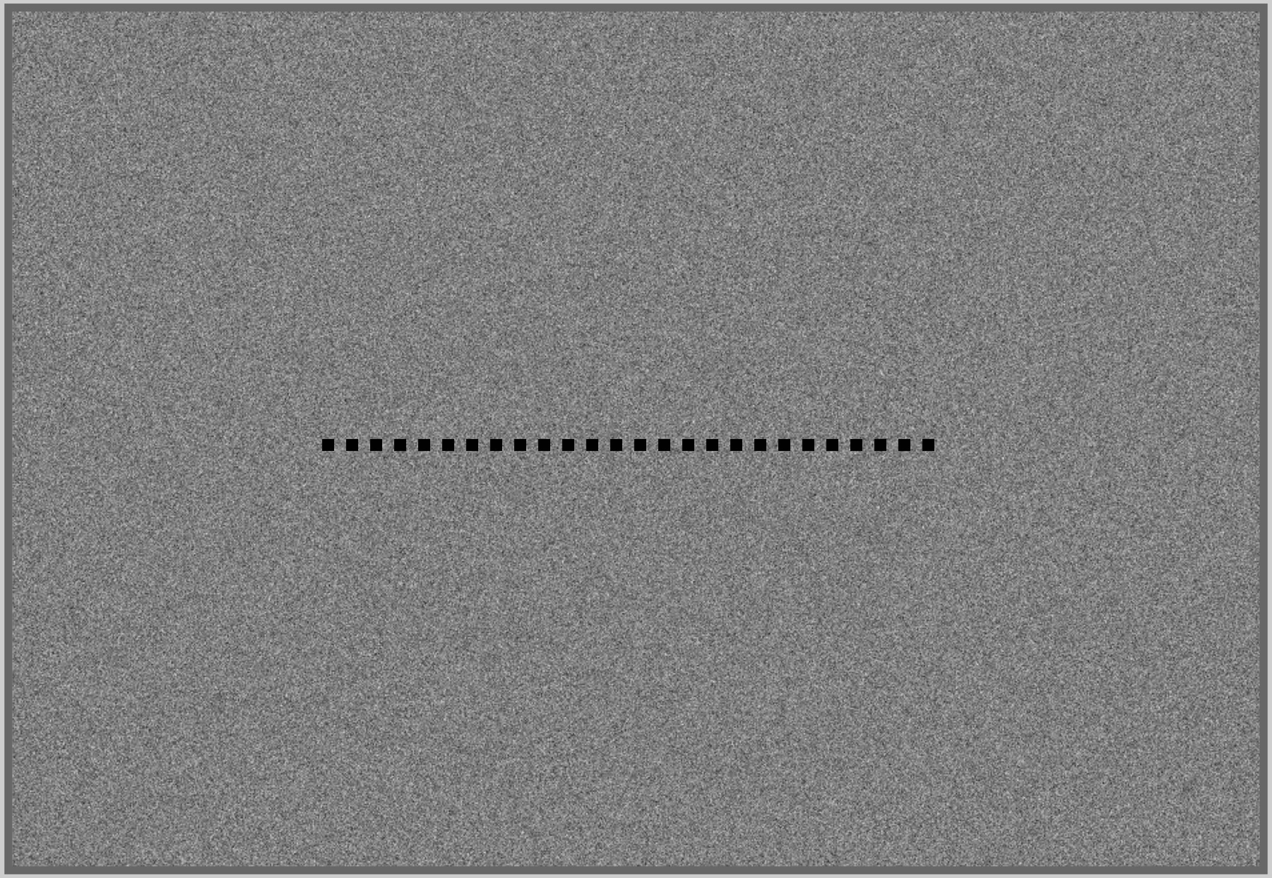
Part
3

3 – Noise Power

NPS:
Noise texture

- Statistical fluctuations in the number of x-rays detected in each pixel cause image noise.
- Correlation of the signal amongst pixels results from signal transport processes in the detector. This effects the noise texture.

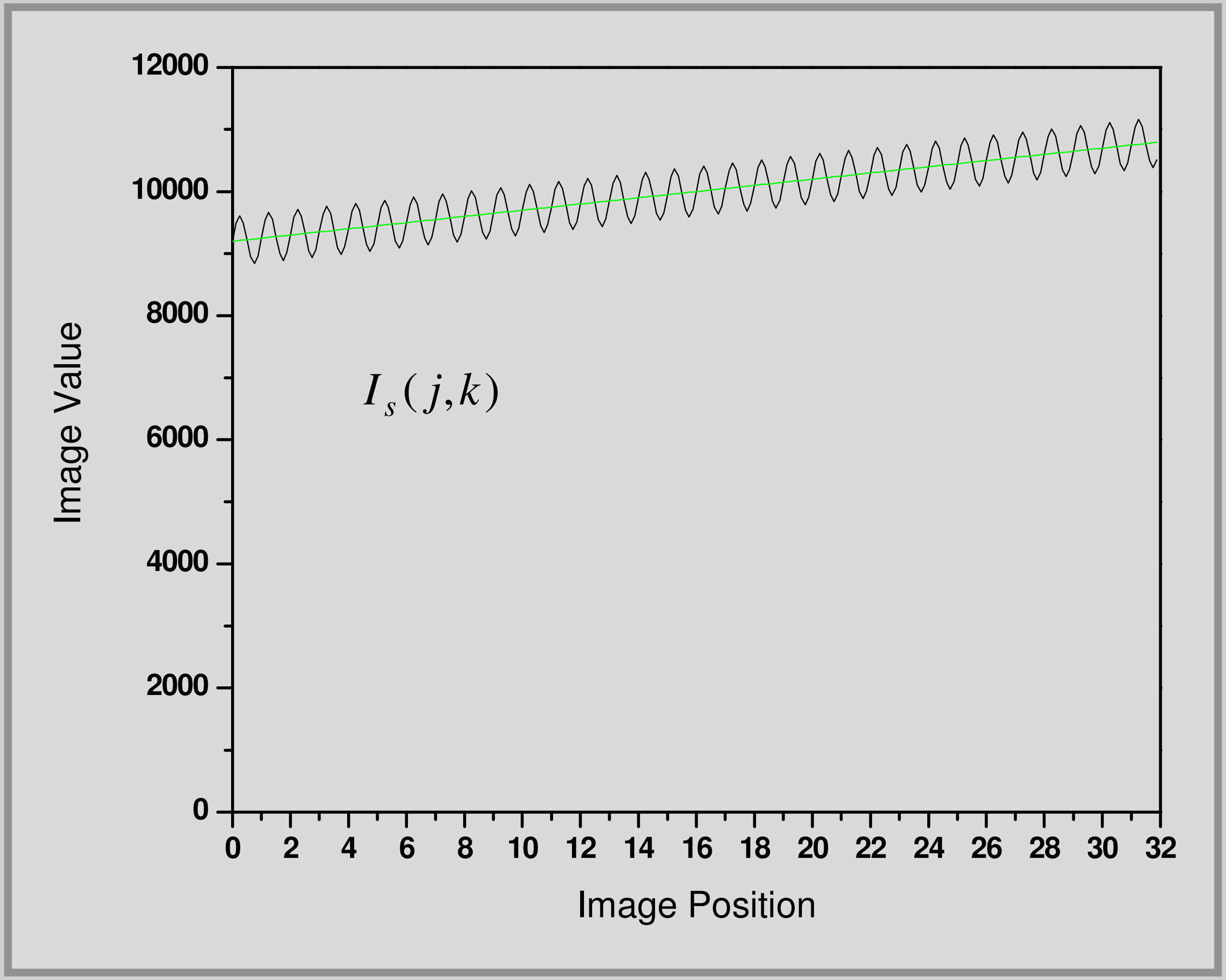
Simulated digital image with random noise having a poisson distribution



A simulated noise signal with a characteristic spatial frequency is illustrated along with a baseline trend for the image values.

NPS:
1D
example

$I_s(j,k)$



Flynn & Samei
Med. Phys.
V 28, N 8
1999

- A low frequency fit to the image data is made and subtracted to remove the effects of trends associate with the heel effect, etc.
- The relative noise is then computed as the residual noise signal divided by the mean value.
- Then the relative noise is normalized by multiplying by the square root of the pixel value.

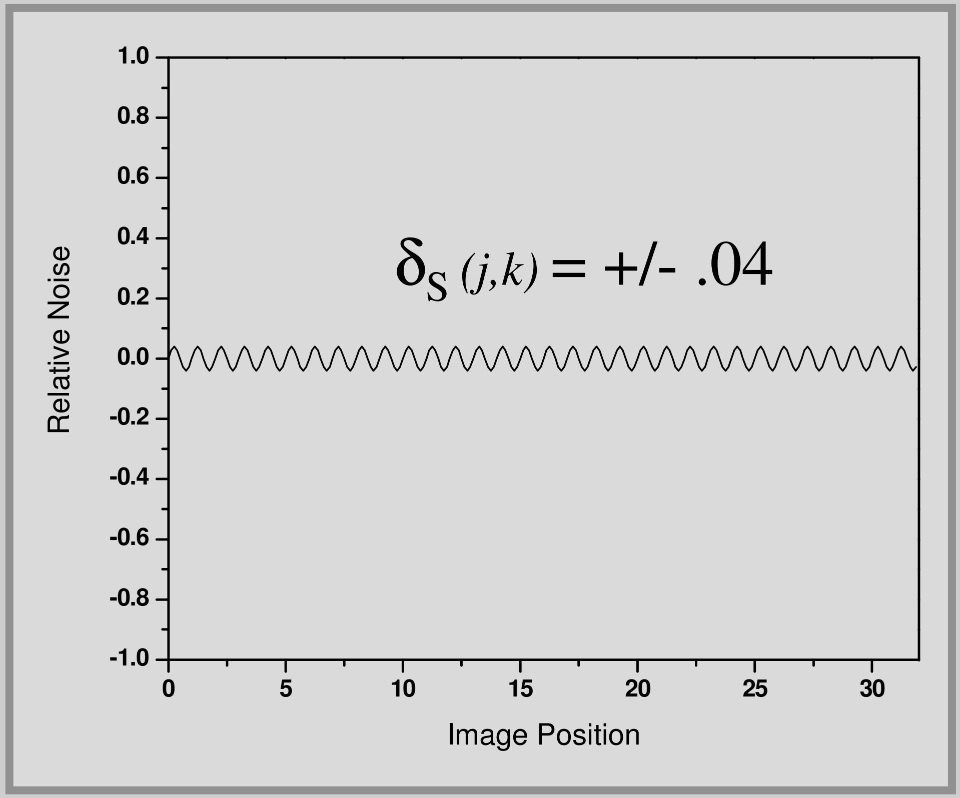
NPS:
Data
Normalization

$$\Delta_S\,(j,k) = I_s(j,k) - I_{fit}(j,k)$$

$$\delta_S\,(j,k) = \Delta_S\,(j,k)\,/\,[I_s]$$

$$\delta_S'\,(j,k) = a_p^{1/2}\,\delta_S\,(j,k)$$

Units: $\delta'^{\,2}_{s}$, 1/(quanta/mm^2) or mm^2/quanta

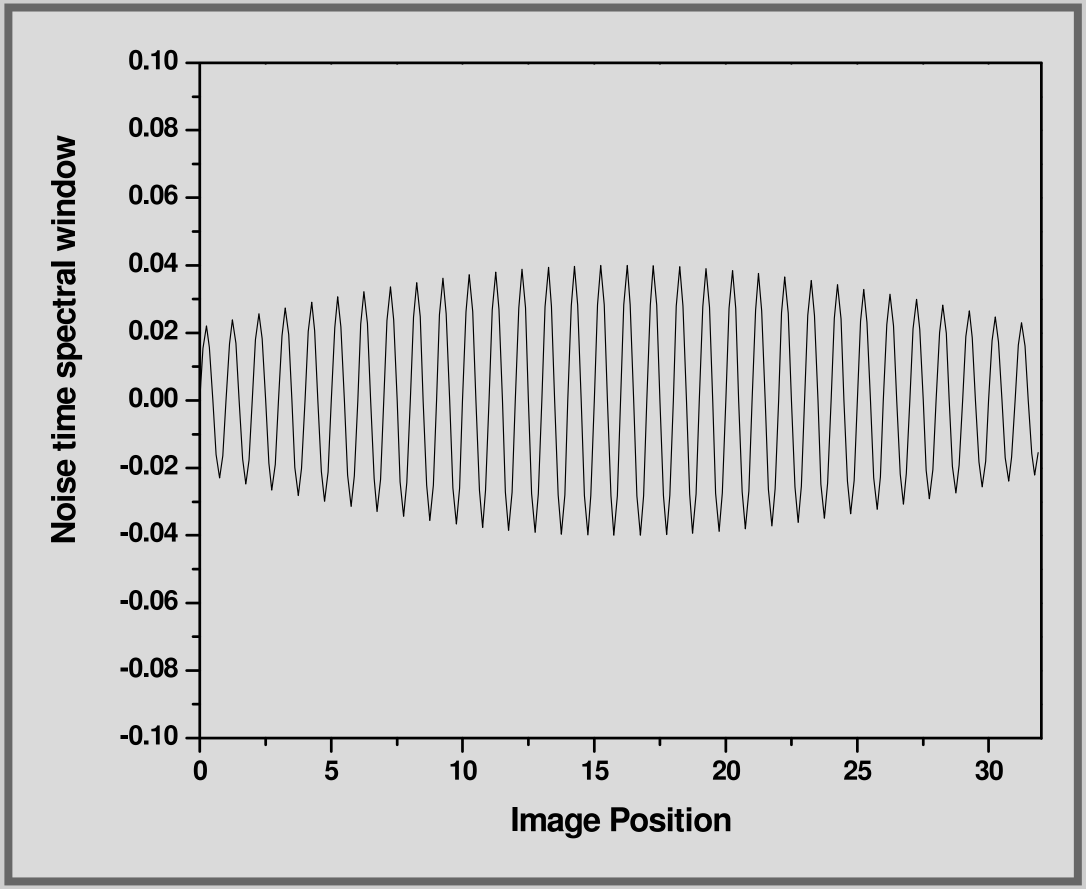


NPS:
Spectral
Estimation

- Prior to the transform, the input is multiplied by a spectral window function (Hamming) according to standard spectral estimation methods

$$\delta_S''(j,k) = H(r)\,\delta_S'(j,k)$$

$$H(r) = .54 + .46\cos(r)$$



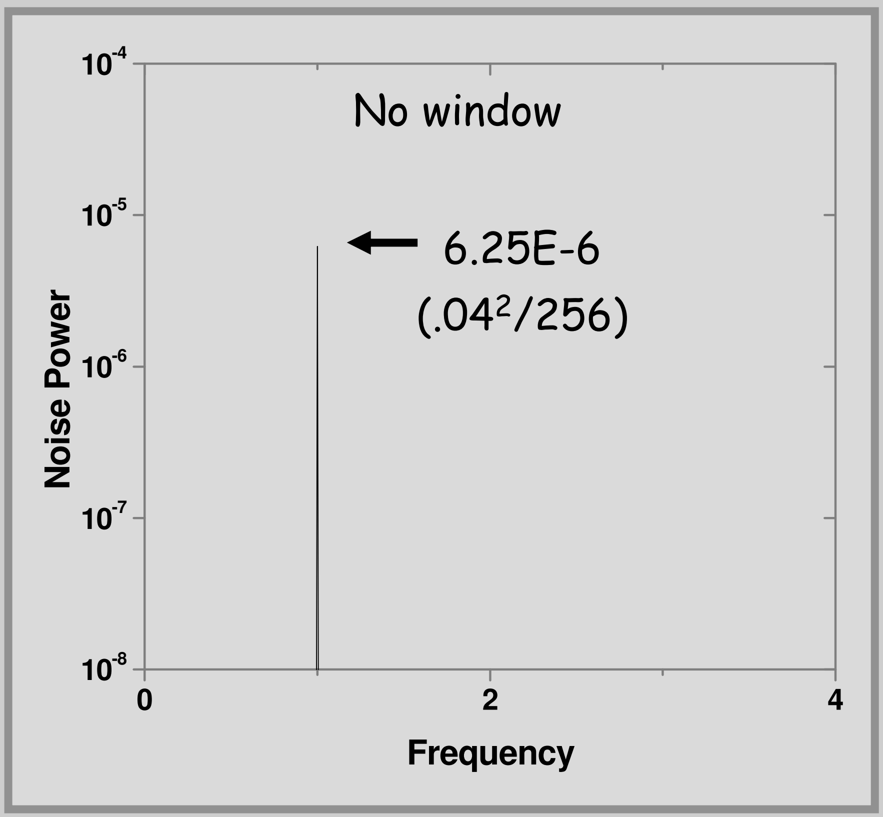
NPS:
Window
Effect

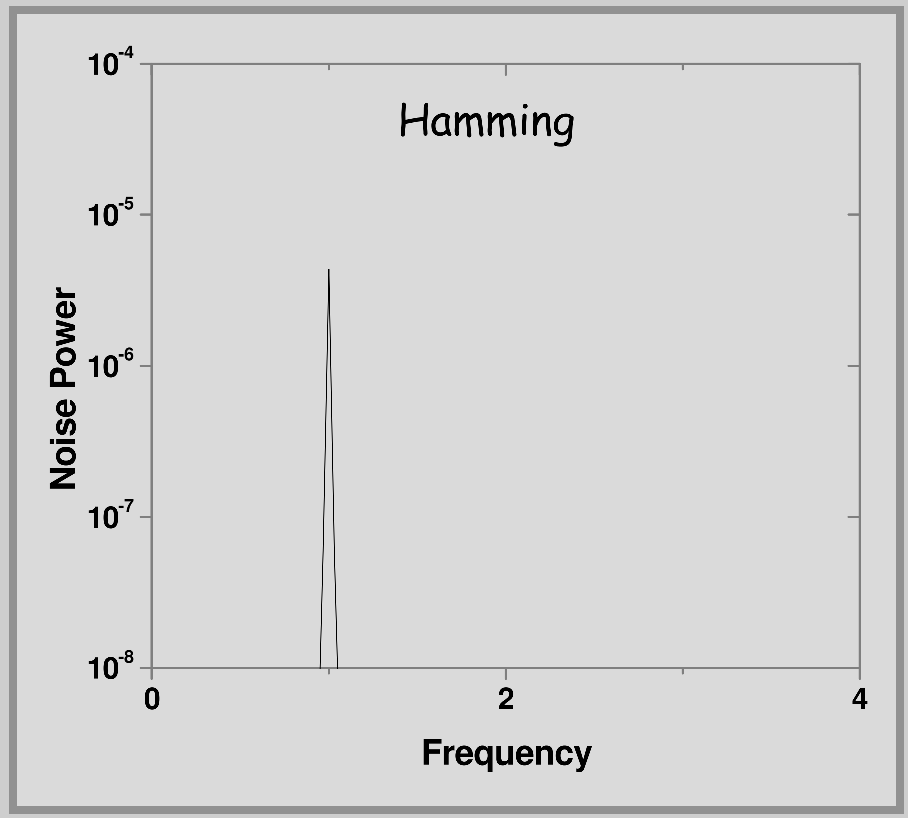
- The NPS is computed as the magnitude squared of the Fourier transform of the normalized relative noise.

$$\eta_s\,(n,m) = \mathrm{FFT}(\delta_S''\,(j,k)\,)\;/\;N$$

$$\mathrm{NPS}_s = [\eta_s\,(n,m)\,]^2$$

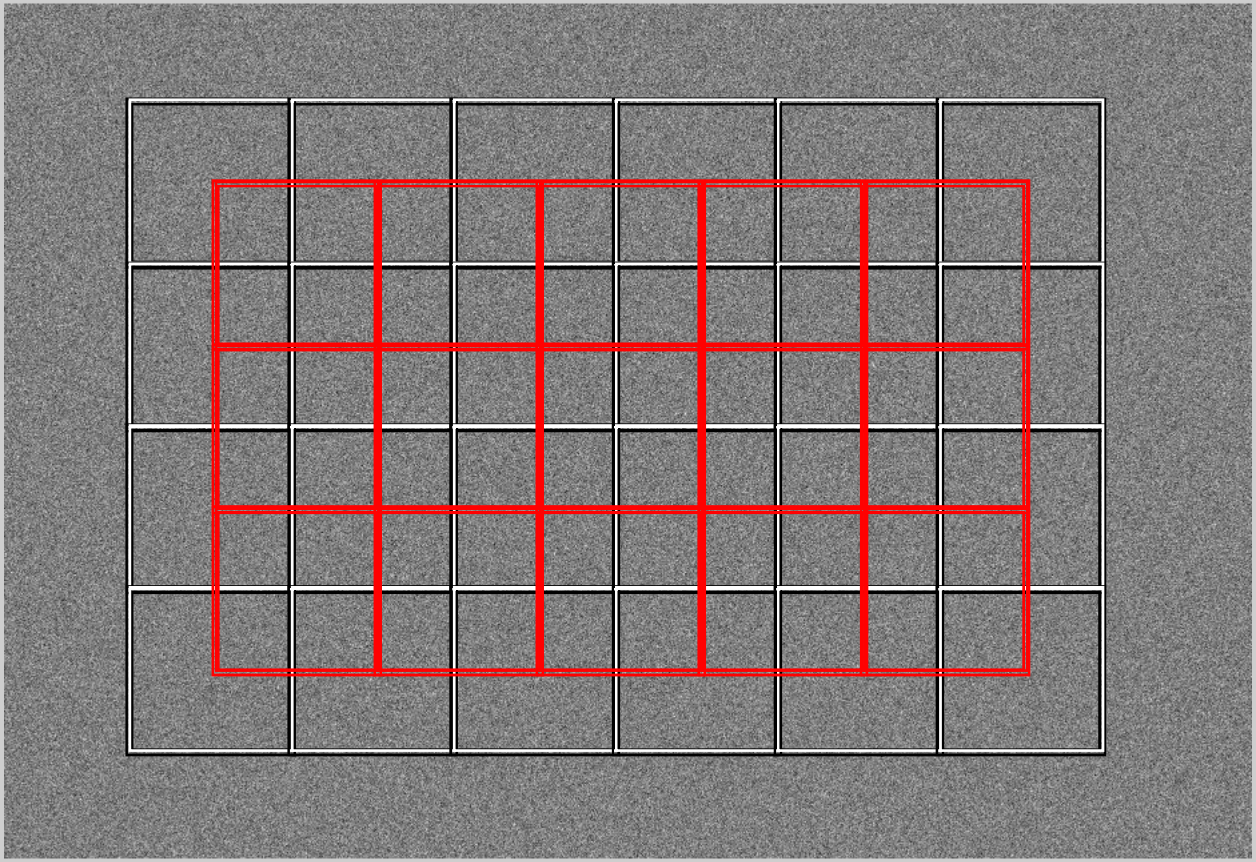
- The spectral window reduces artifacts from the edge of the window, but broadens the spectral resolution.





NPS:
2D Block Average

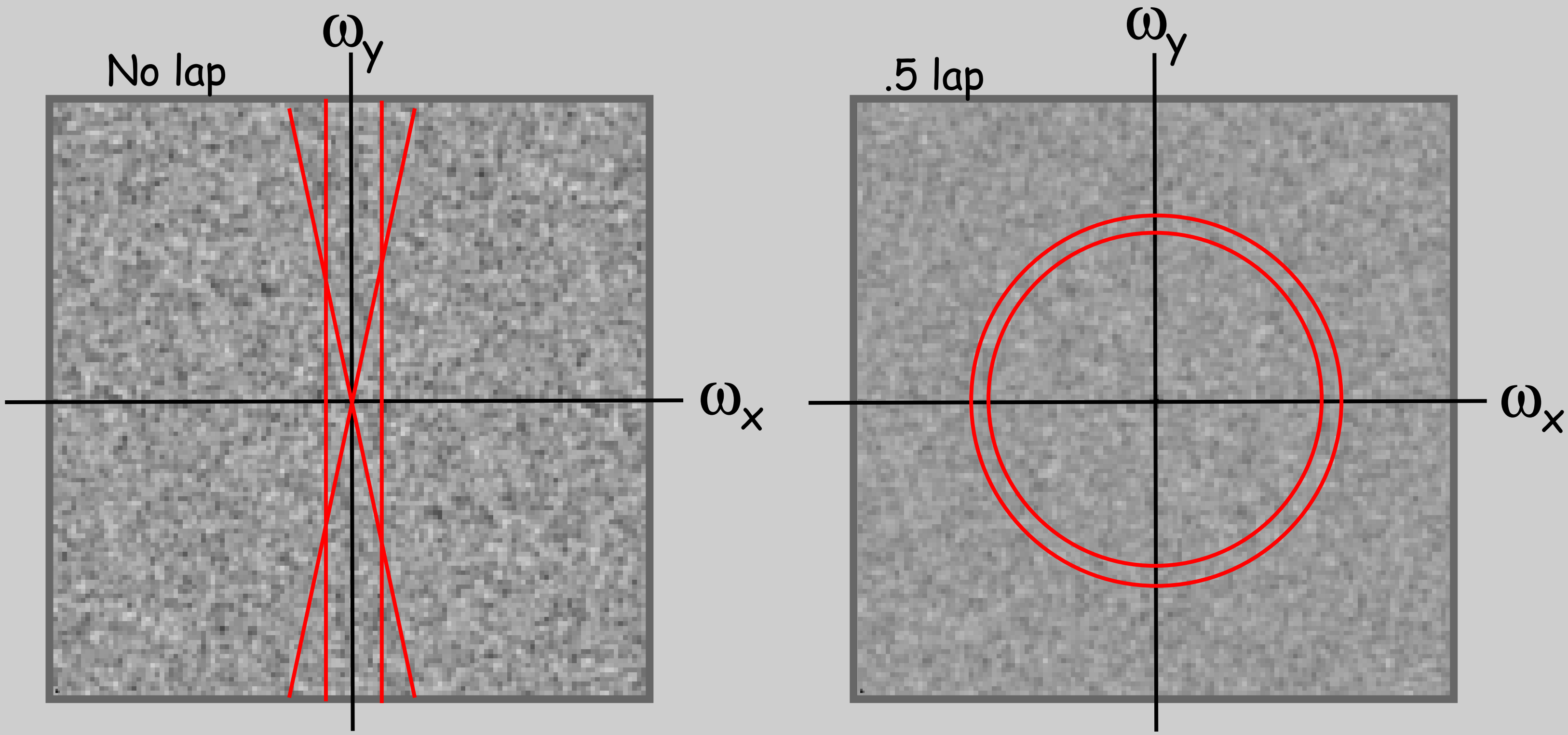
To reduce noise in the estimate of the NPS at the expense of spectral resolution, the 2D NPS is computed in many small blocks and averaged.



•Flynn, Med. Phys., V 26, N 8, 1999

NPS: 1D Results from 2D NPS

- The 2D NPS can be displayed as an image with values proportional to the log of the NPS.
- An 1D estimate can be derived from the 2D NPS by averaging all values within;
  - NPS(y) : A horizontal band about the w_x axis
  - NPS(x) : A vertical band about the w_y axis
  - NPS(r) : A circular band centered on the origin



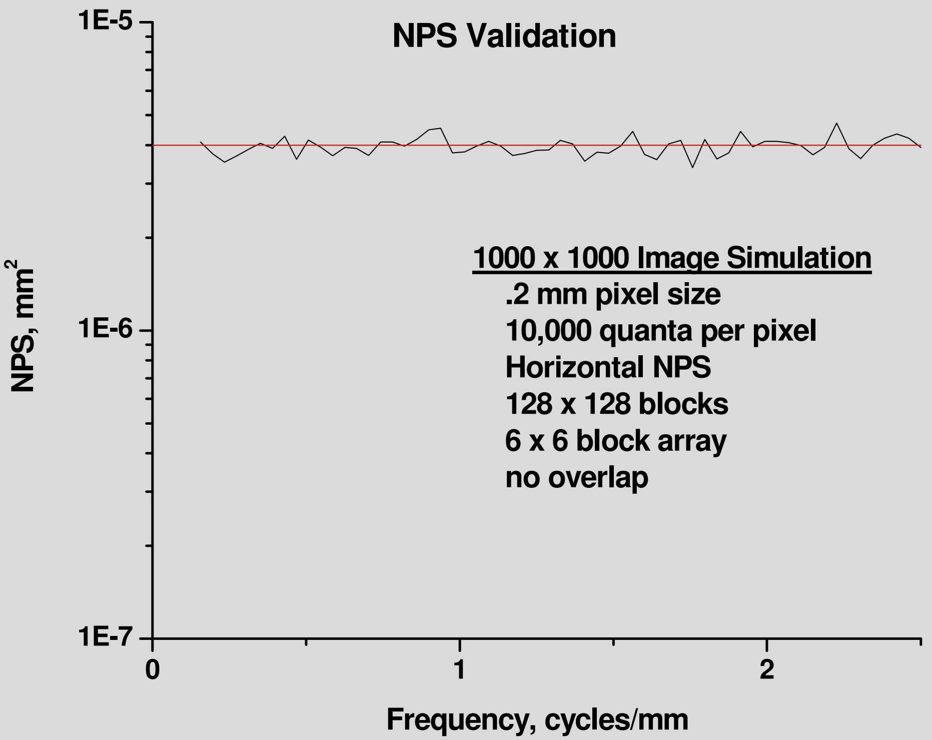
NPS:
Block
Overlap
Effect

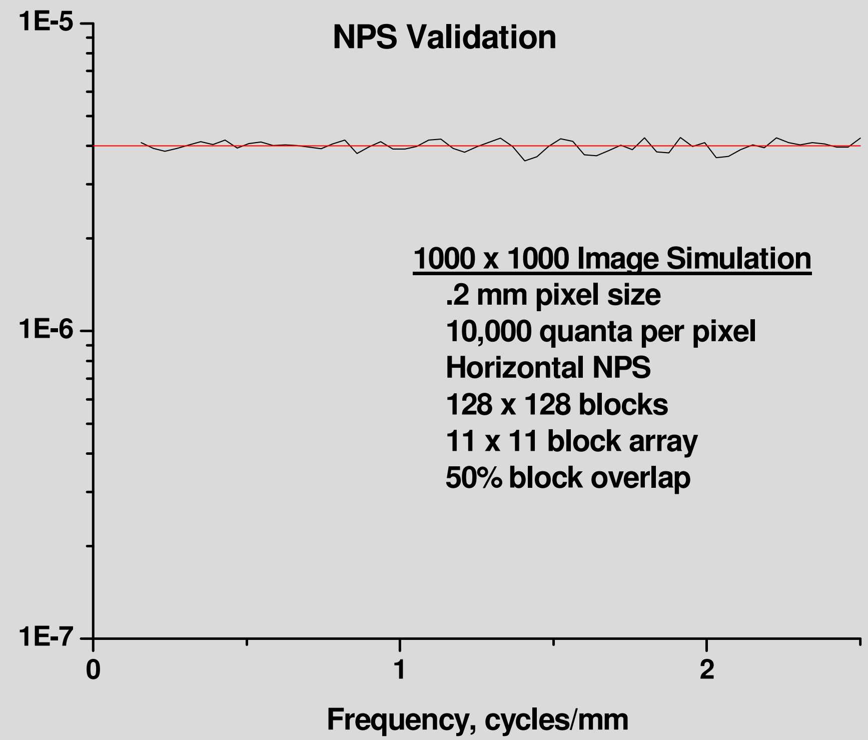
NPS for a simulated image

uncorrelated gaussian noise

10,000 #/pixel x 25 pixels/mm^2 = 250,000 #/mm2

1/250,000 = 4E-6 mm^2

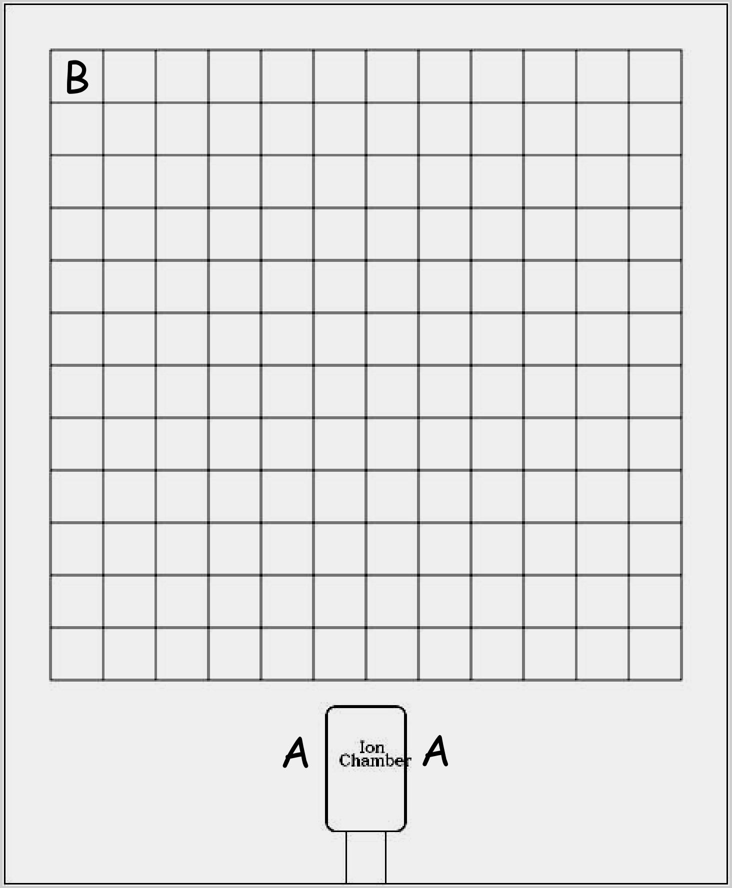




NPS;
exposure
conditions

- For all NPS measurements, the exposure to the detector for a uniform field is measured at the edge using an ionization chamber.
- The chamber is placed midway between the source and the detector and the exposure measurement corrected for distance to produce an estimate of the exposure in air at the position of the detector.

- The resulting estimate of NPS is corrected for exposure trends and related to a specific exposure to the detector.
- The exposure of a reference block is estimated as the measured exposure times the ratio of the linear signal at B and at A.
- The NPS of each block is linearly adjusted by the ratio of the linear signal in the block to the linear signal in the reference block.



IEC Standard Radiographic Conditions

NPS:
Radiographic Technique

#	added Al Filtration	HVL mm Al	nominal kVp	typical kVp
RQA5	21 mm	7.1	70	72-77
RQA9	40 mm	11.5	120	120-124

The IEC radiographis conditions provide consistent beam quality by adjusting kVp to achieve a specific measured HVL. Errors in generator calibration are therefore conveniently avoided.

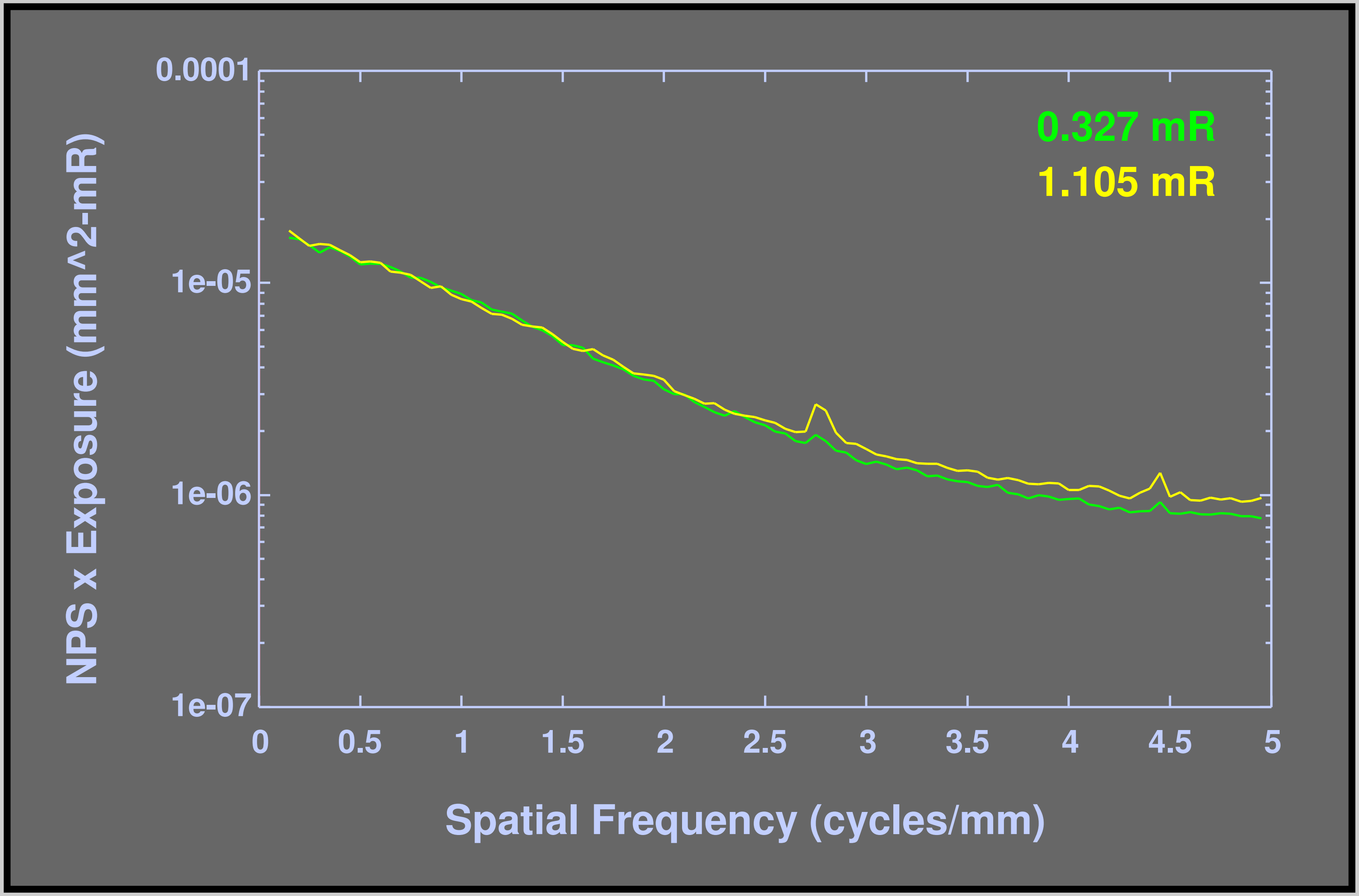
It is now common to measure NPS over a range of exposure values. For quantum mottle, the NPS will be inversely proportional to exposure. Deviations from this behavior indicate fixed pattern or electronic noise sources.

Suggested Nominal Exposures

.1 .3 .6 1.0 3.0 6.0 10.0

NPS:
NPS
exposure
product

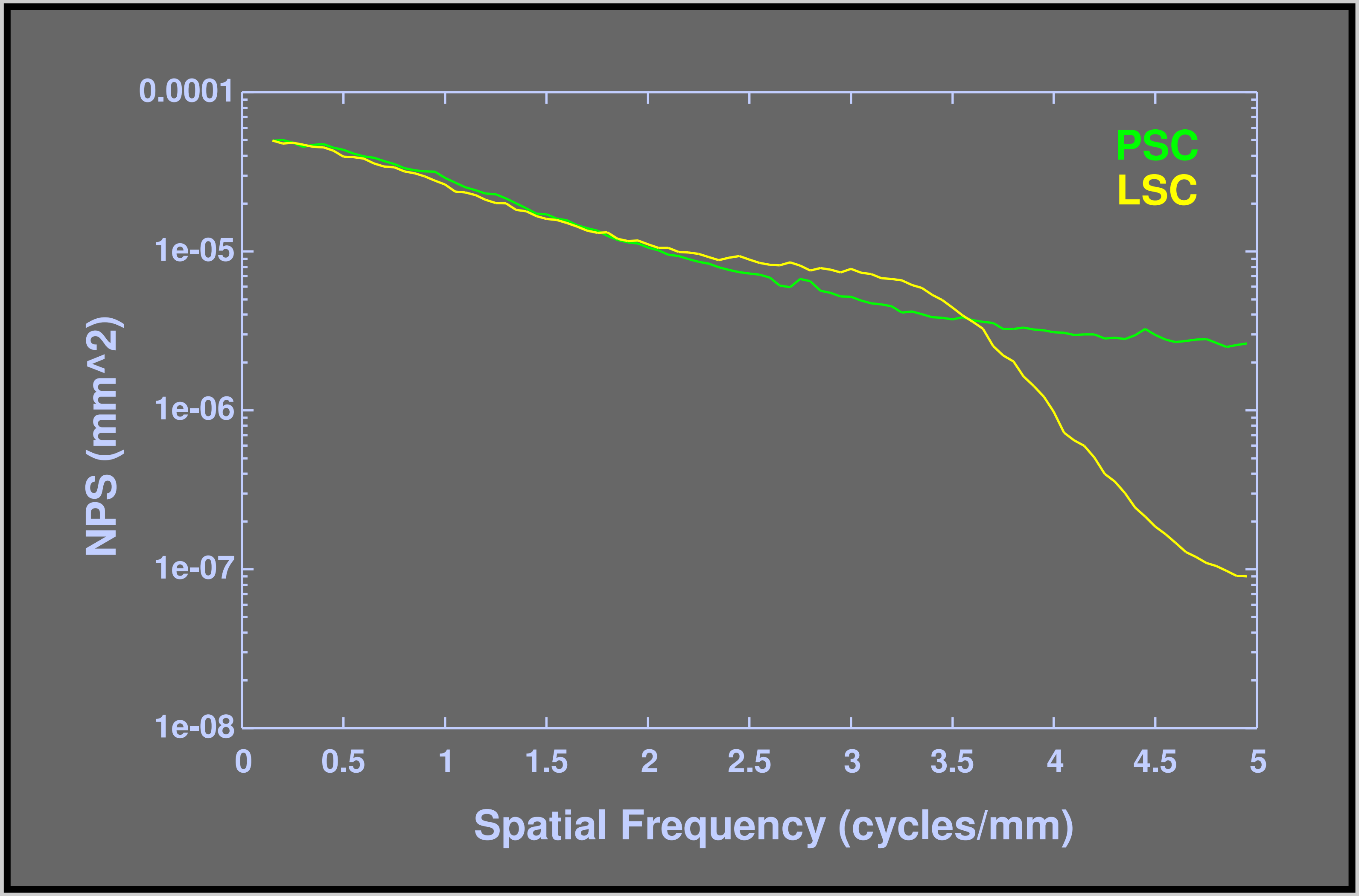
The comparison of results made at different exposures can be done by plotting the product of NPS and exposure



Samei & Flynn
SPIE
1997

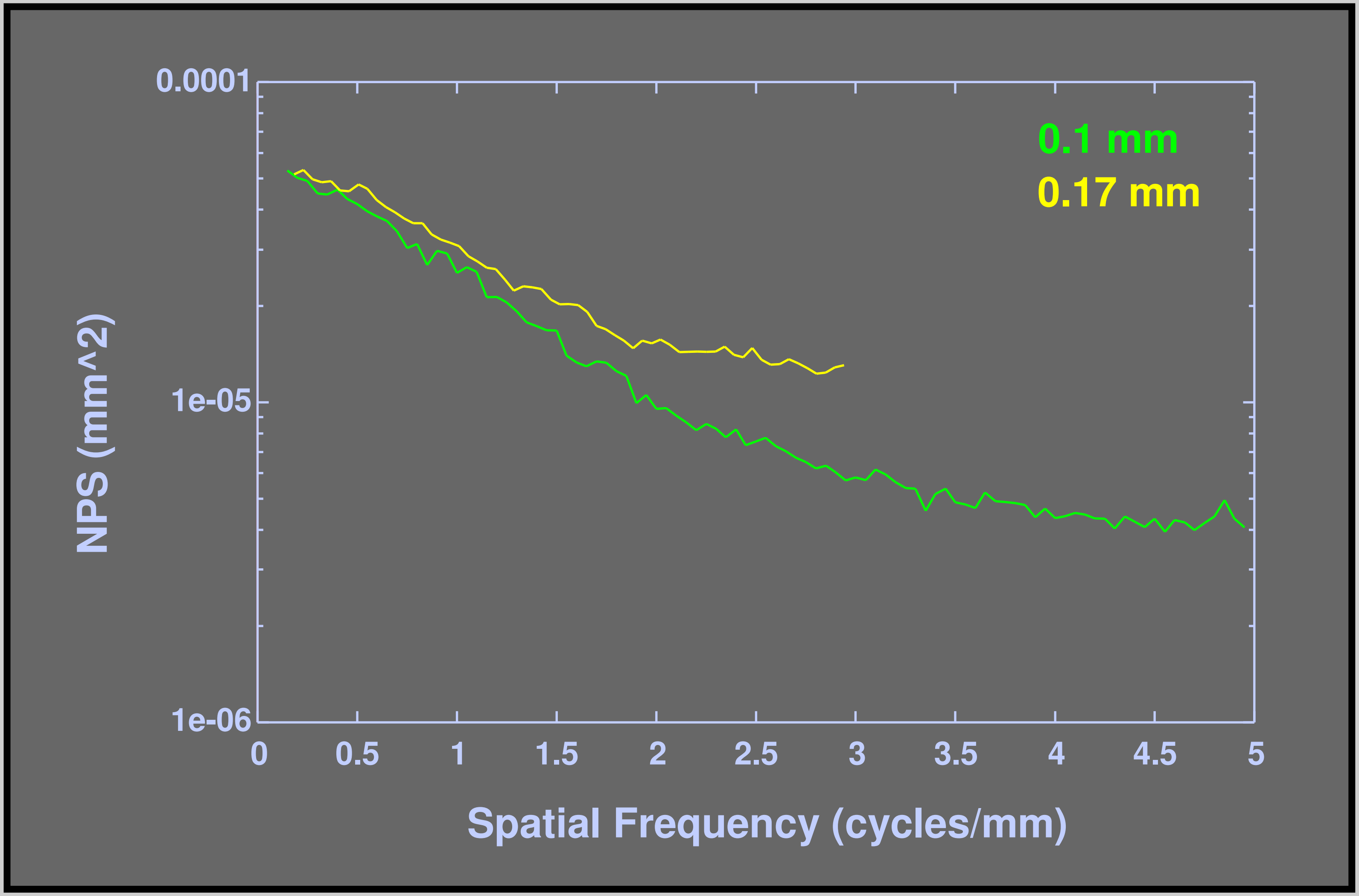
Asymetric noise power properties have been observed for some CR systems due to electronic filtering of the signal in the laser scan direction.

NPS;
vertical-horizontal difference



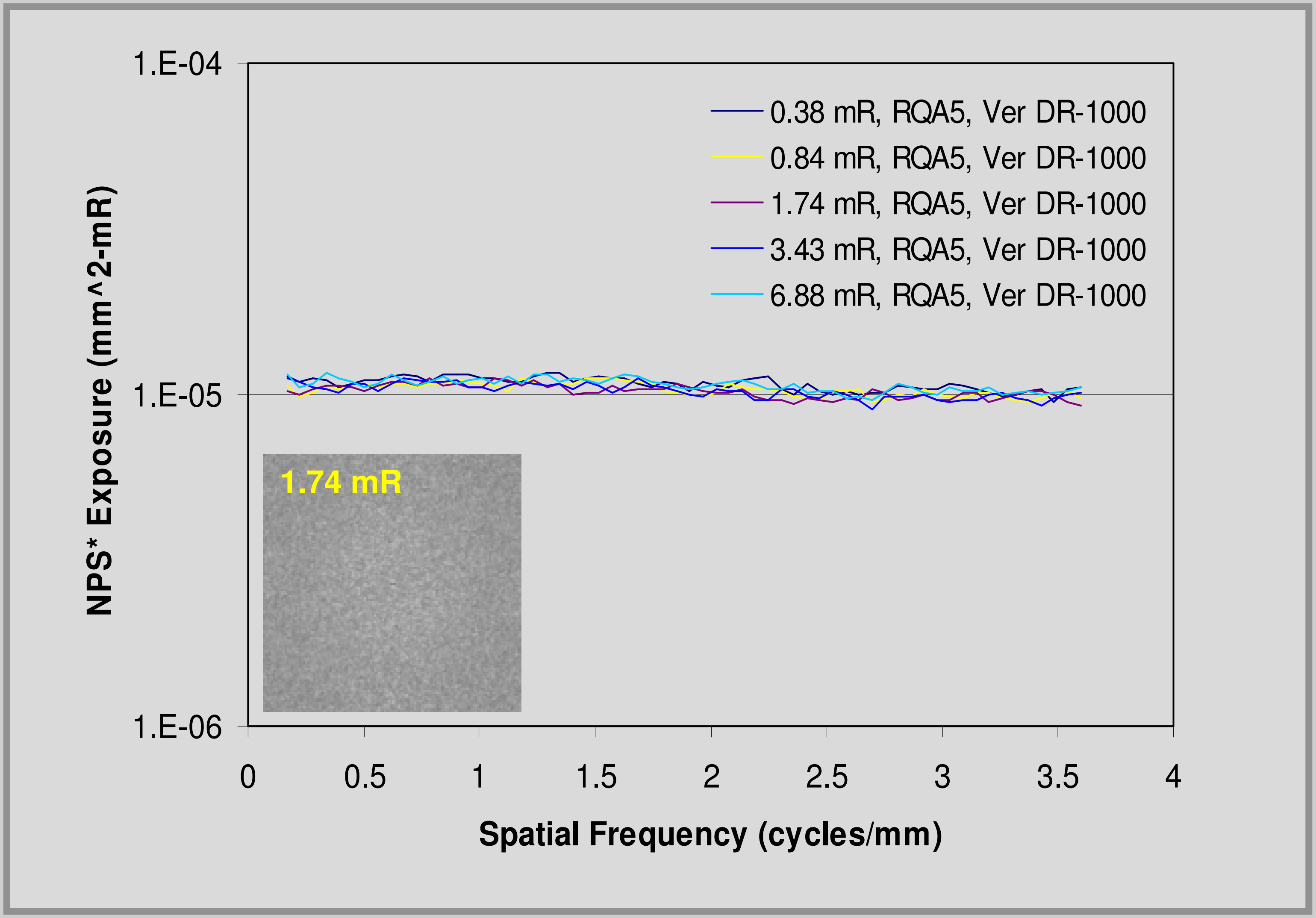
NPS: Effect of Aliasing

- Estimates of NPS are bandlimited to the limiting frequency associated with the pixel sample spacing.
- Presampled noise power will thus be aliased causing the NPS near the limiting frequency to be elevated.



NPS:
Se
Flat
Panel

- Minimal charge drift or capacitive coupling occurs between pixels in a Se flat panel detector.
- Contrary to the MTF results, MC models indicate little influence on the NPS due to fluorescent radiation.



Part
4

4 – Detective Efficiency

DQE:
Detective
Quantum
Efficiency

- DQE describes the measured SNR^2 in relation to that of an ideal detector. Interpreted as a counting detector, SNR^2 is a Noise Equivalent Quanta, NEQ, #/mm^2
- The measure SNR^2 is deduced from the ratio of the MTF squared (signal^2) to the NPS (noise^2).

  Since the NPS has units of mm^2, this ratio has the desired units of #/mm^2

$$DQE(f) = SNR^2_{meas} / SNR^2_{ideal}$$

$$DQE(f) = \frac{\left(MTF^2(f) / NPS(f)\right)}{Q \times X_{meas}}$$

$$Q = SNR^2_{ideal} / X_{meas}$$

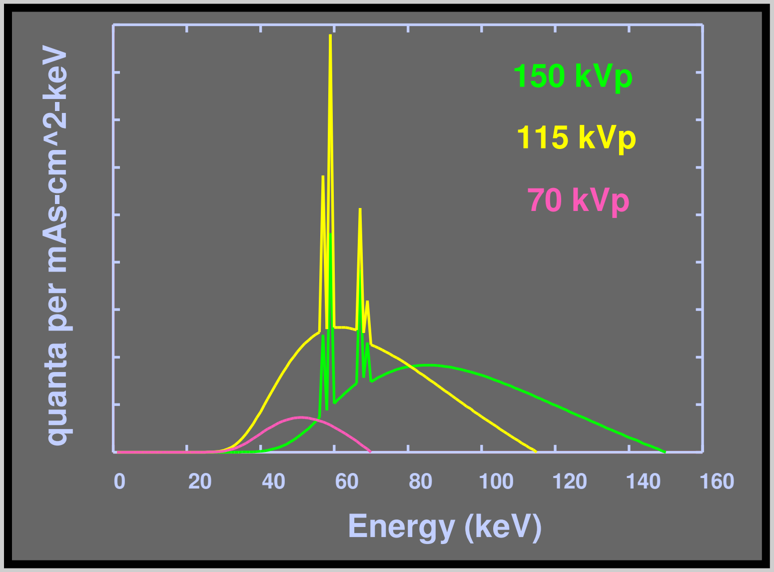
$$DQE(f) = \frac{MTF^2(f)}{q \times \left(NPS(f) \times X_{meas}\right)}$$

Note: classical expression for DQE include a gain term, G, for detectors with non linear response. Since we assumed a linear signal, the gain term is unity.

DQE: ideal SNR

- The Ideal SNR^2 is computed from an estimate of the x-ray spectrum and a model of an ideal energy integrating detector;

$$q = \frac{\left(\int_0^{kVp} \varphi(E) E dE\right)^2}{\int_0^{kVp} \varphi(E) E^2 dE} \bigg/ X_\varphi$$



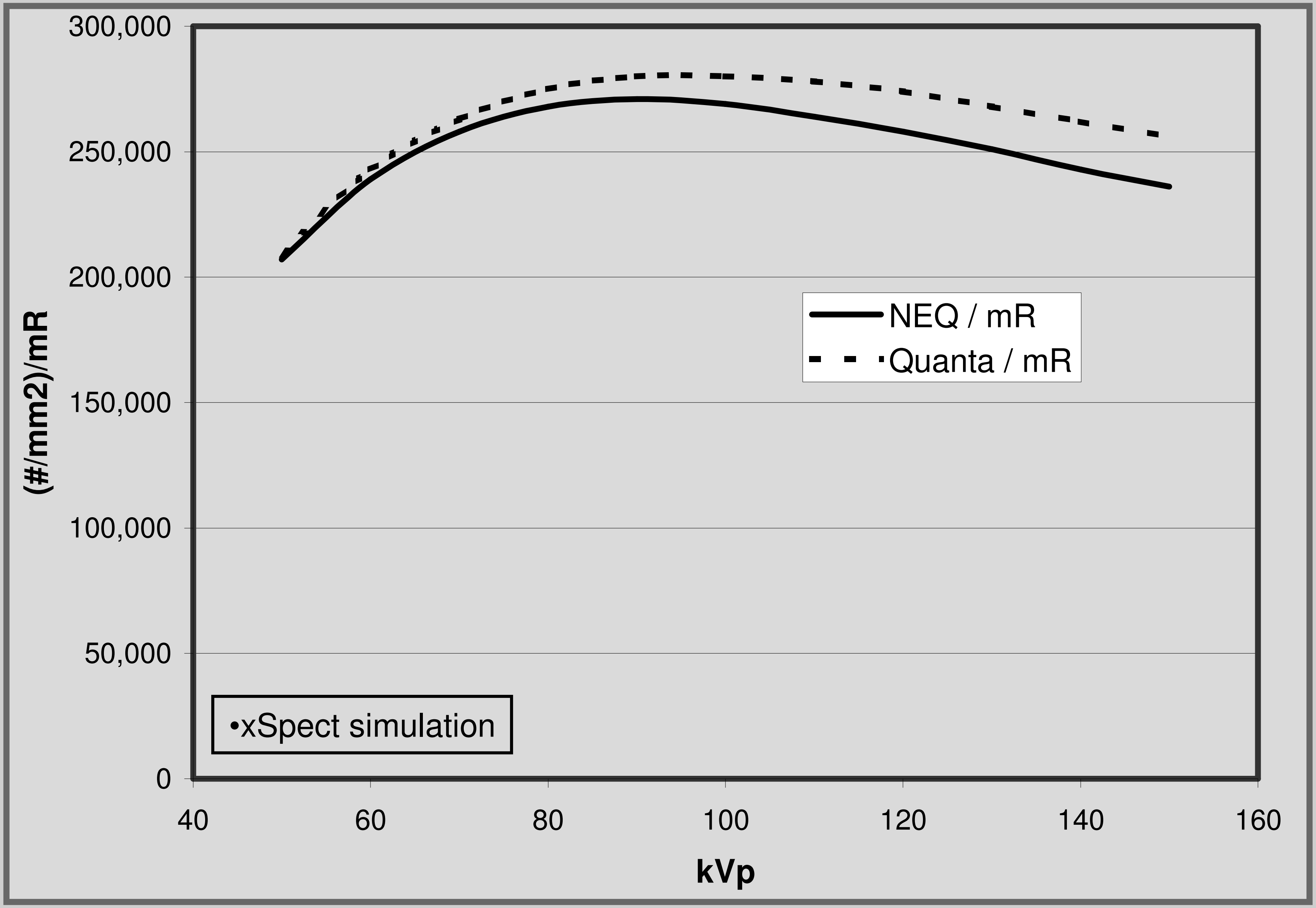
where E is the photon energy and , φ(E) is the x-ray spectrum, and X_f is the exposure in mR associated with the x-ray spectrum φ(E).

- Values for X_f are obtained by computing the energy absorbed in air for the spectrum φ(E) using mass enery-absorption coefficient data obtained from the National Institute of Standards and Technology.
- The energy absorbed in air is then converted to charge using a W value of 33.97 J/C (i.e. eV/ion pair) (Boutillon 1987).

Note: q values estimated for an ideal enery integrating detector are higher by 3-5% than for an ideal counting detector

The noise equivalent quanta per mR is a slowly varying function of kVp and filtration.

Q, quanta per mR



DQE:
CR
Systems

The DQE of CR systems is largely determined by the thickness and binder structures of the powdered phosphor screen. Screens from all manufacturers perform similarly.

Standard-resolution plates, ~ 0.1 mm pixel size, PSC

	Agfa ADC-Solo (MD-10)	Fuji FCR-9501 (ST-Va)	Kodak KESPR-400 (GP-25)	Lumisys ACR-2000 (Agfa MD-10)
		70 kVp		
DQE(0)	21.0%	28.9%	28.6%	22.0%
DQE($2mm^{-1}$)	7.5%	11.2%	10.8%	5.6%
		115 kVp		
DQE(0)	18.5%	23.0%	22.4%	15.5%
DQE($2mm^{-1}$)	7.0%	9.2%	8.5%	3.8%

- While the MTF of high resolution CR screens is marketly better the DQE at low frequency is poor.
- For small body parts with detail structures (orthopaedic extremities), the improved efficiency above 2 cycles/mm makes the HR screen preferable.

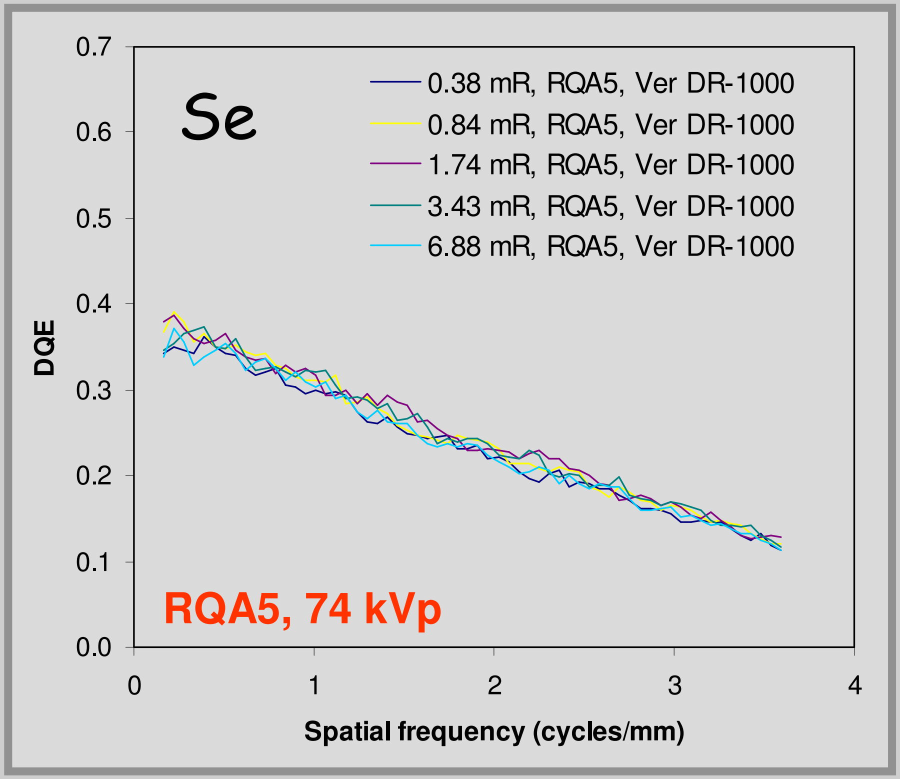
DQE:
High Res. CR Screens

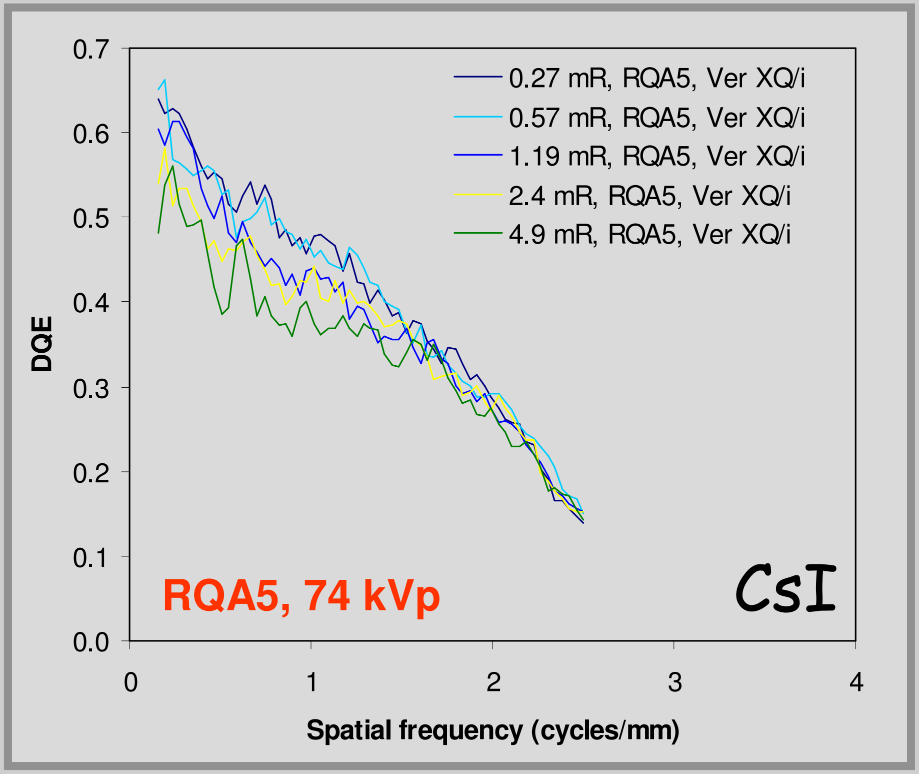
115 kVp, 0.1 mm pixel size, PSC

	Kodak KESPR-400 (GP-25)	Kodak KESPR-400 (HR)
DQE(0)	22.4%	13.9%
DQE ($2mm^{-1}$)	8.5%	8.7%
0.2 MTF	2.56 (mm^{-1})	3.81 (mm^{-1})
0.1 MTF	3.60 (mm^{-1})	4.95 (mm^{-1})

DQE:
Flat Panel Detectors

- The detective efficiency of Se (direct) and CsI (indirect) flat panel detectors exhibit a crossover at a spatial frequency of 2.2 mm^{-1}
- The excellent MTF of Se detectors and good efficiency at high frequency should make it useful for small body part with detailed structures.
- The very high low frequency efficiency of CsI detectors make it attractive for soft imaging of large body parts.



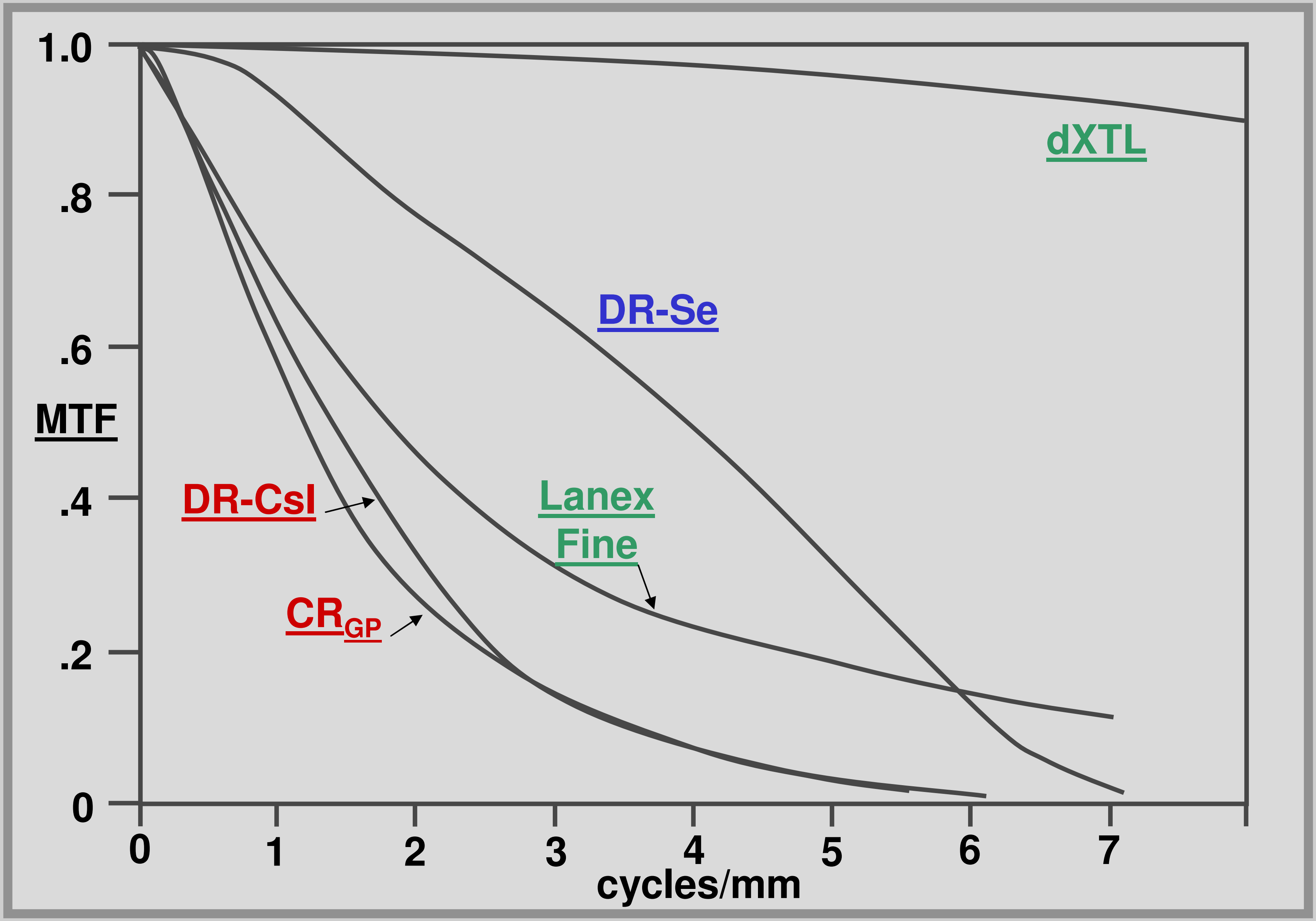


Part
5

5 – Diagnostic Value

Typical MTF

MTF – CR, DR, and film



Hand
Phantom
CR

CR

Distal
Ulna

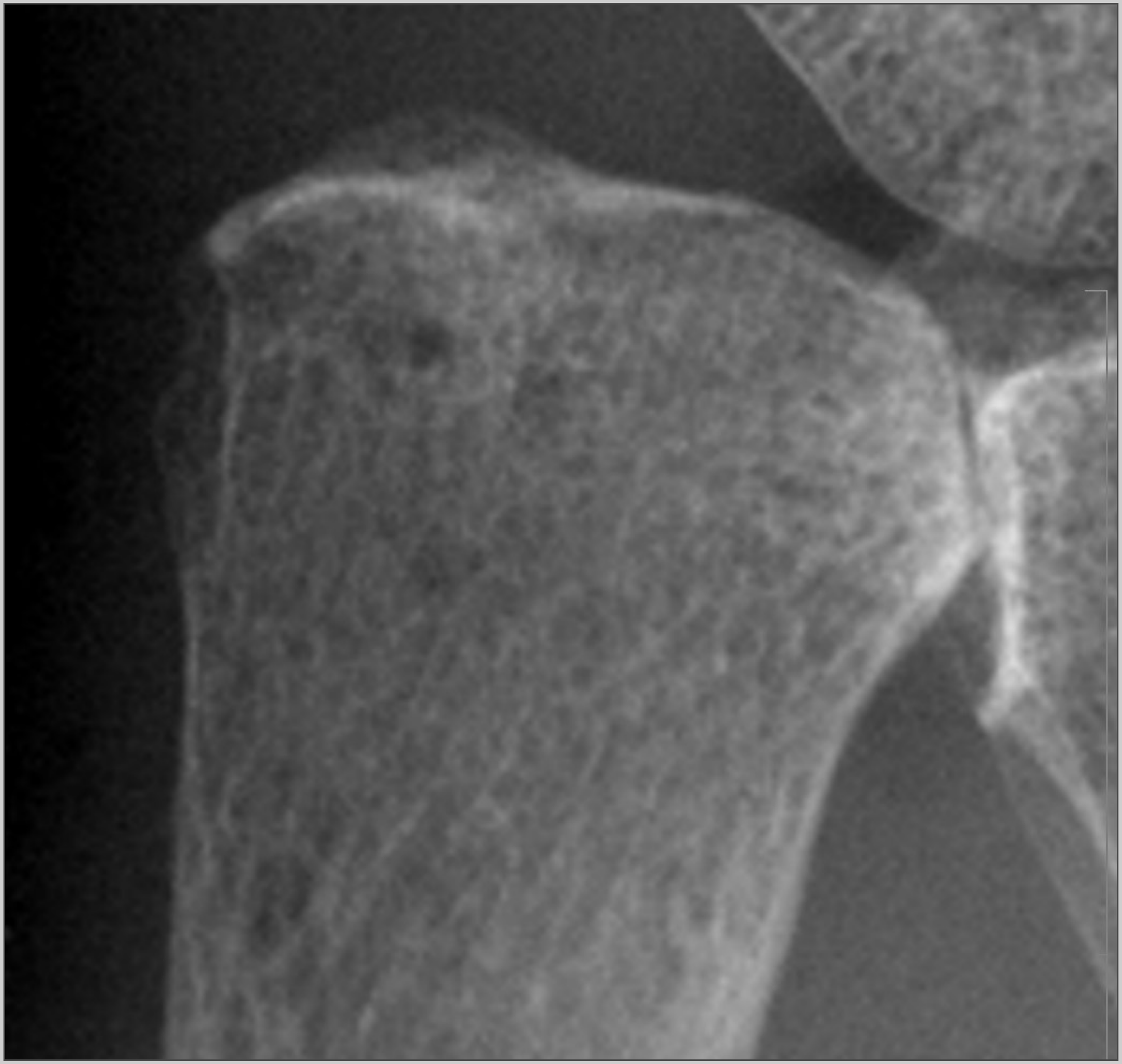
200
Speed
Exposure

Hand
Phantom
CR

DR

Distal
Ulna

200
Speed
Exposure



Hand Phantom digitized XTL film

XTL Film

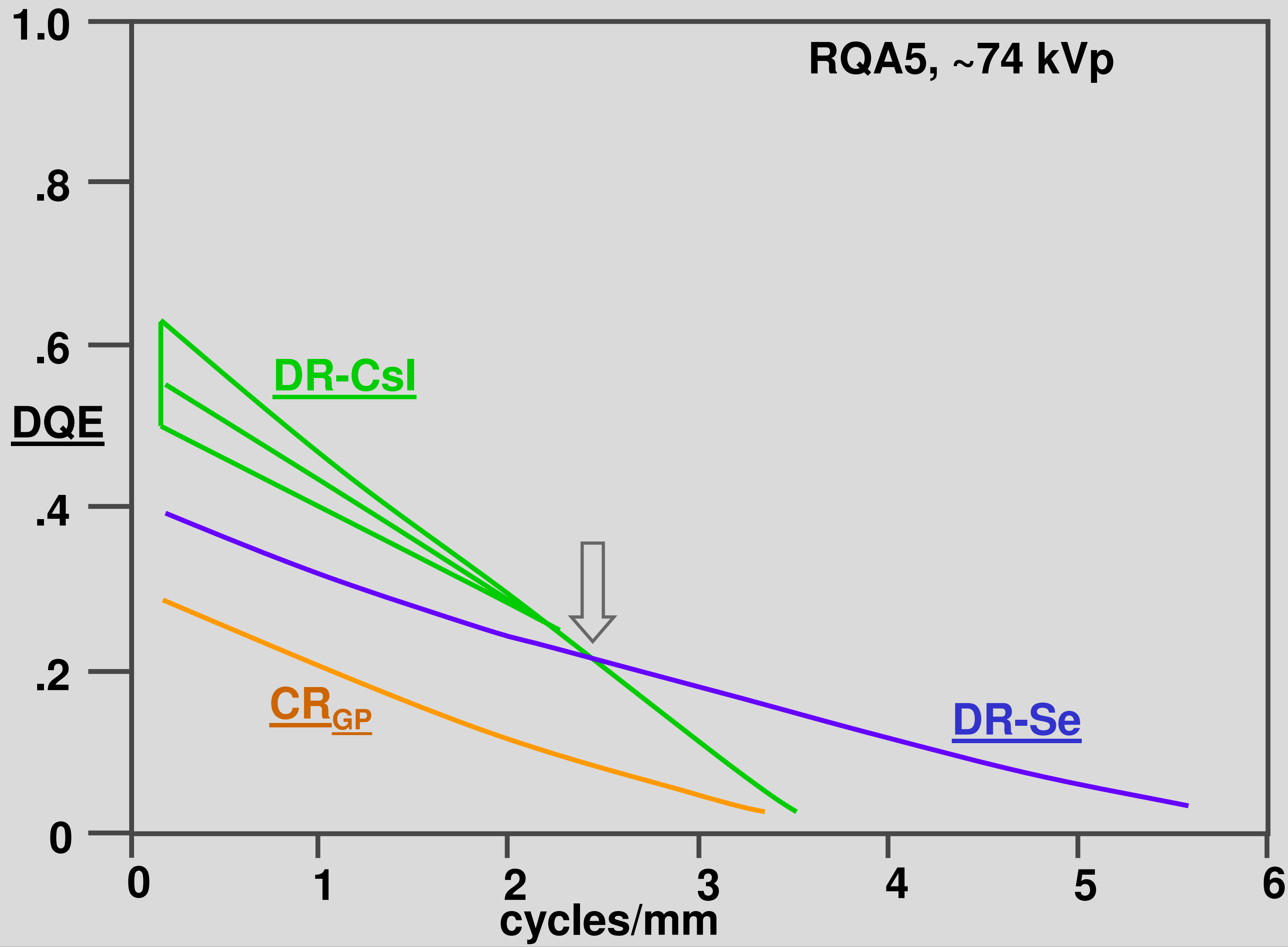
Distal Ulna

7 Speed Exposure

RADIOLOGY RESEARCH

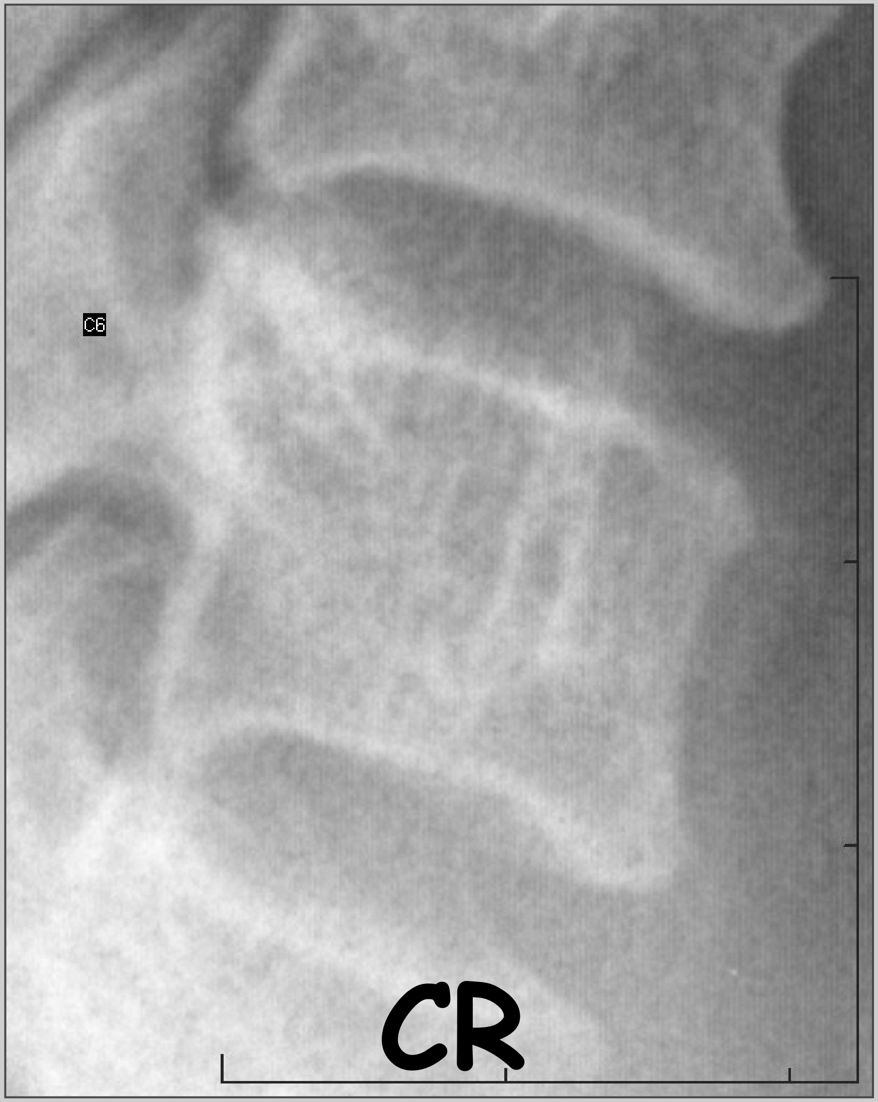
Typical DQE

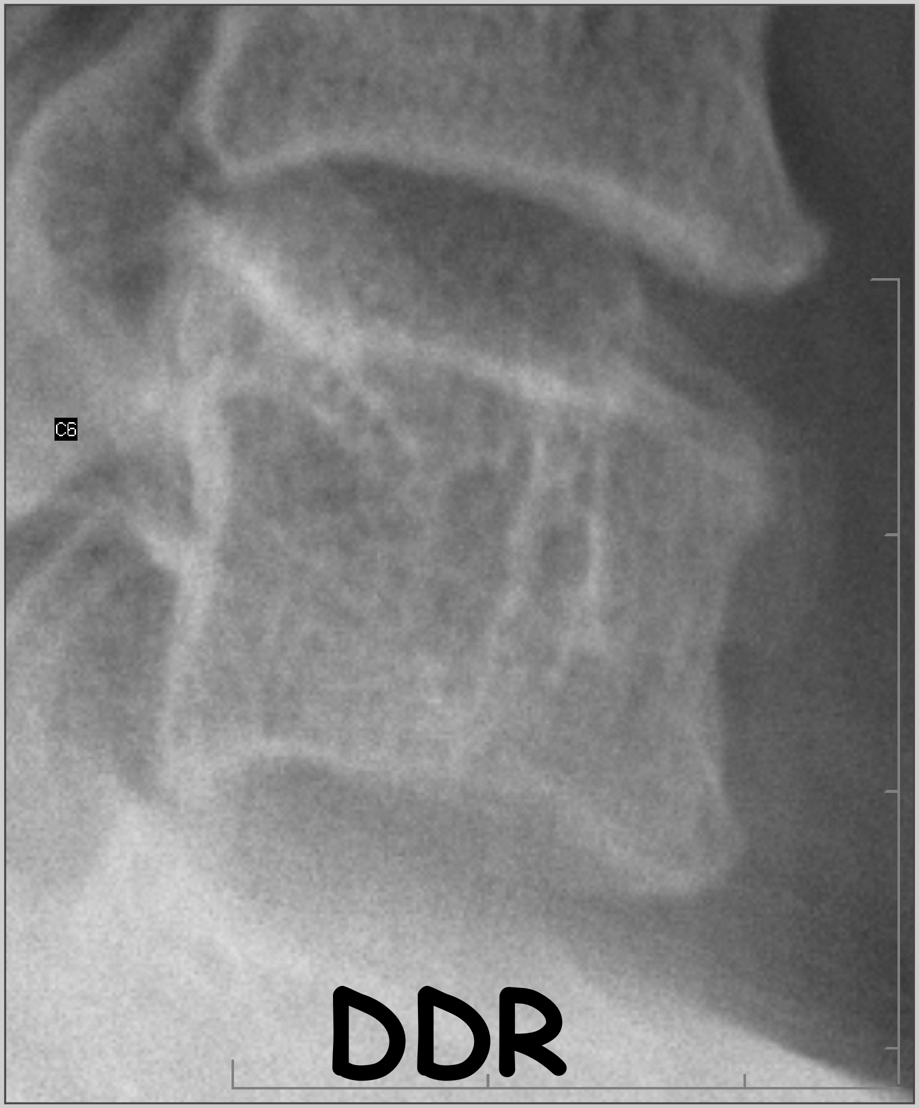
DQE - CR and DR



Spine noise

Lateral Spine - Equivalent exposure

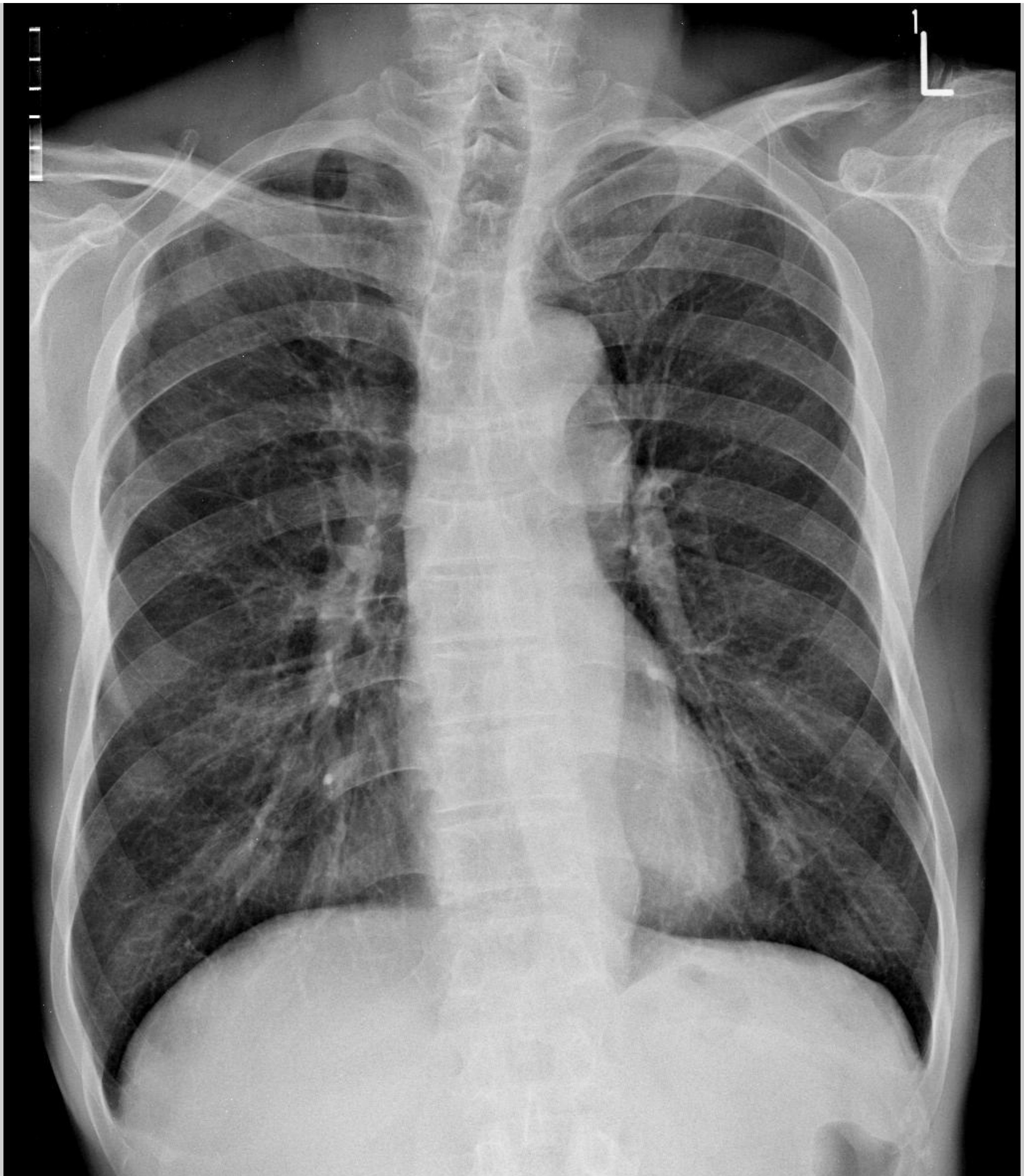





Indirect
Direct

Value in thoracic imaging

Is the diagnostic value of reduced noise (dose) in mediastinal regions more important than the value of improved detail in lung regions where quantum noise is minimal?



EKC CR 900

Musculo-Skeletal Value

Musculo Skeletal specialist place high value on detail relative to the relative noise(dose) in soft tissue regions

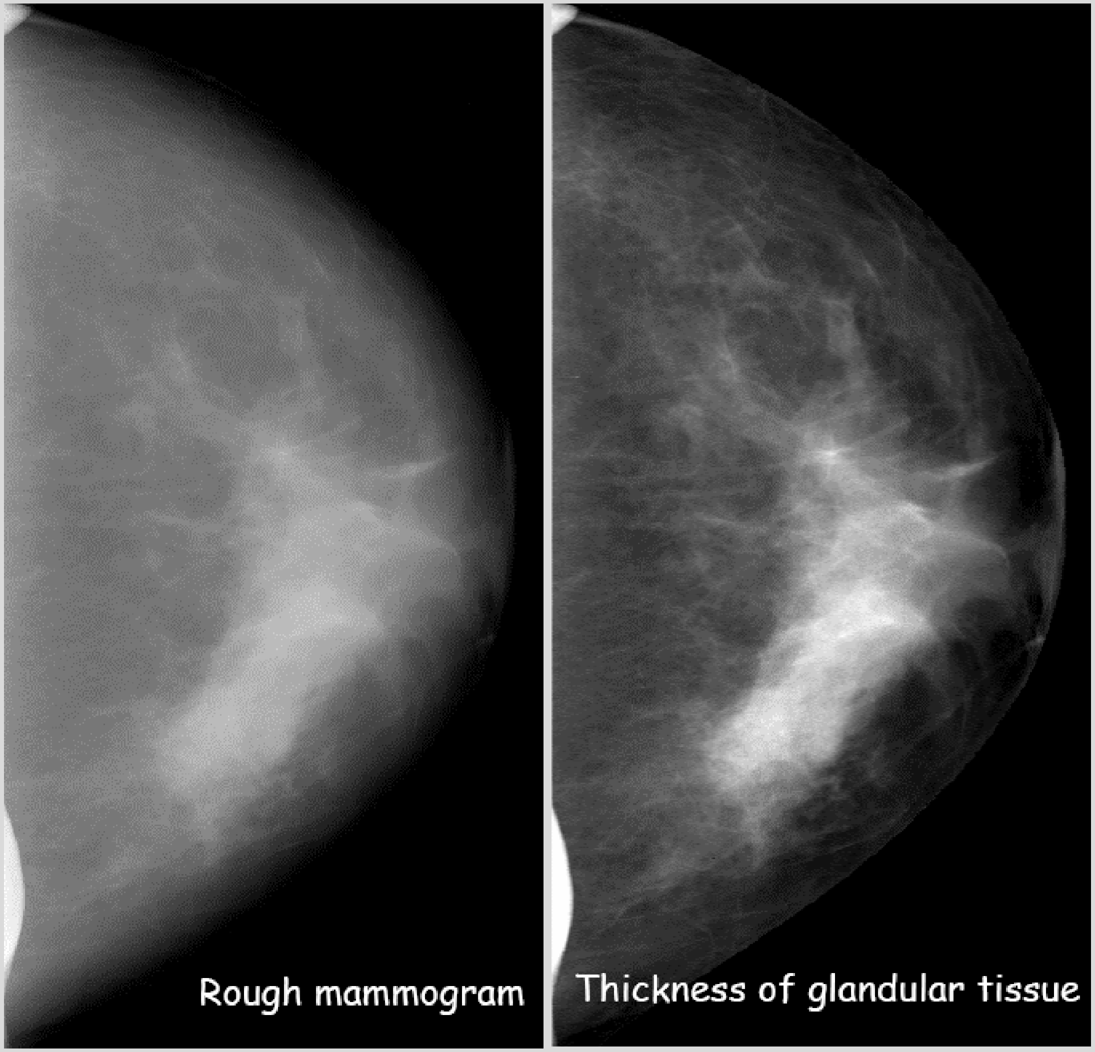
Dual Energy Chest

- Dual Energy digital radiography adds value by separating tissue types.
- Key to the method is the ability to obtain two images very rapidly.

Images from
GE Medical Systems (Web)

Dual Energy Mammography

- Dual Energy imaging can be used to separate glandular tissue and fat in the breast.
- This improves contrast for the tissue structures of interest.



Jean-Louis AMANS, Web 2002

LETI, Systems Imaging Laboratory

Diagnostic Value

- Systems using digital radiography detectors have numerous value attributes
  - Image Quality – Detail and Noise(dose)
  - Processing - Equalization, Restoration, …
  - Speed – Static vs dynamic
  - Geometry – gantry, scatter rejection
  - Advanced - Dual Energy, Tomosynthesis

- The relative diagnostic value of these attributes is not well understood.
- DQE describes only a particular property of an imaging detector. It does not address the issue of diagnostic value.