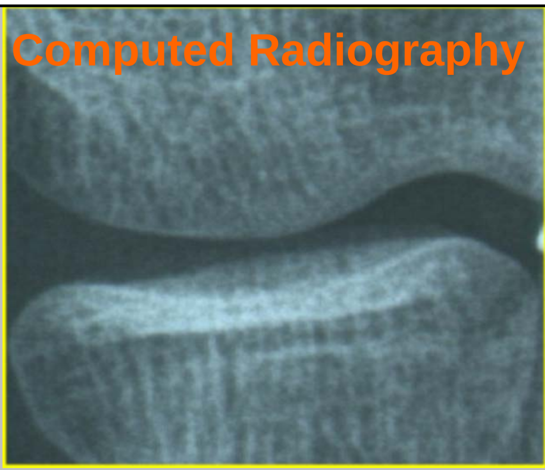
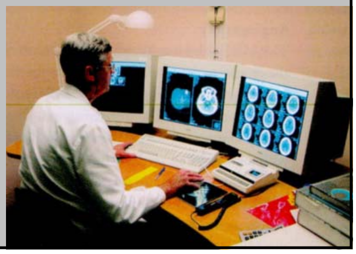
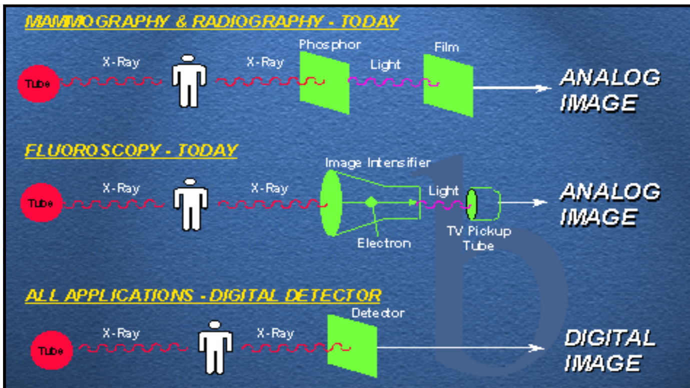
Basis of Computed Radiography & PACS



Slavik Tabakov

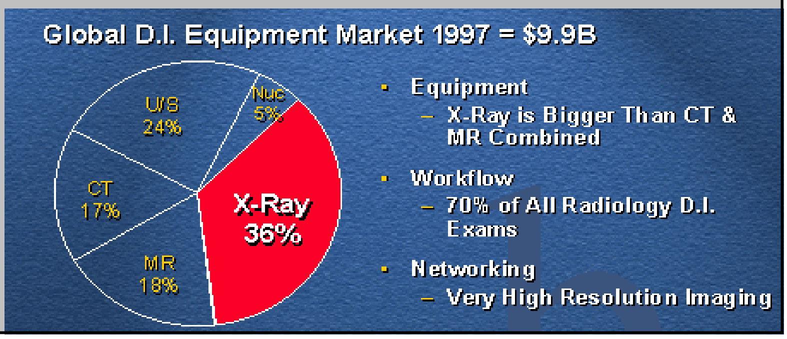




Computed Radiography (CR) refers to new types of X-ray detectors (i.e. replaces the X-ray Film)

The CR output media is a digital image, which can be processed, printed and stored in PACS

Most of the X-ray examinations are radiography based



Comparison

Digital

Film-screen

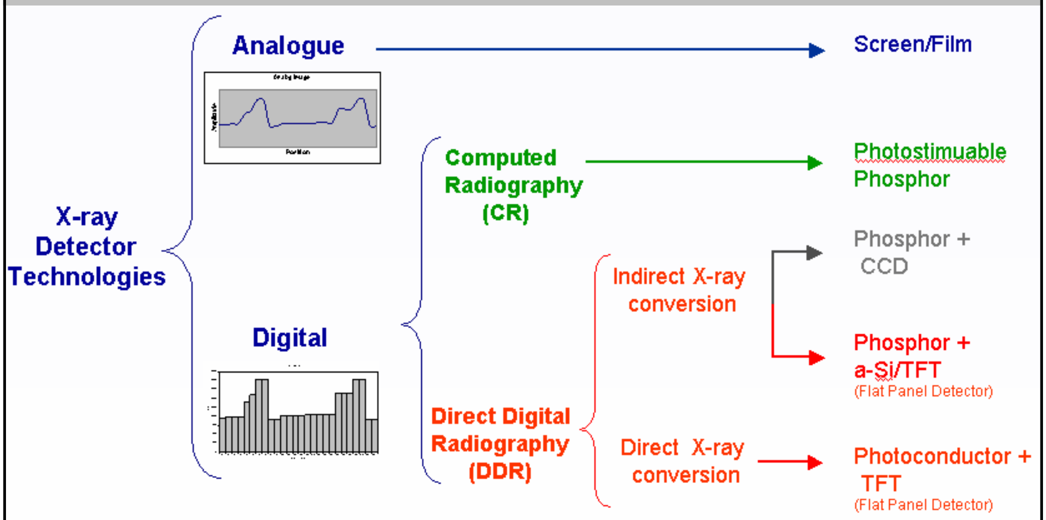


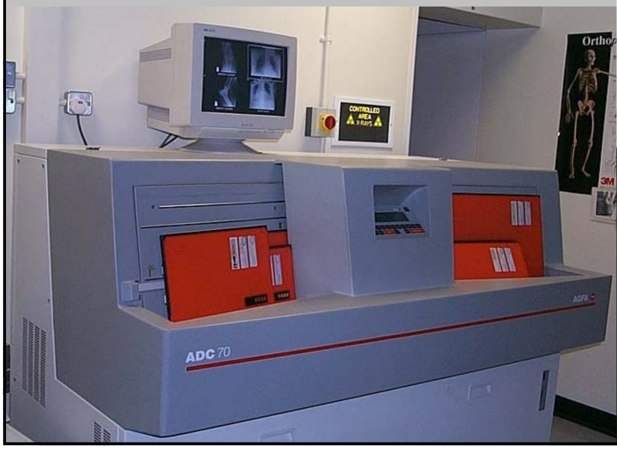
Fig.3 – Current X-Ray detector technologies for x-ray diagnostic imaging

Source: A. Pascoal

CR system using laser stimulated ***storage phosphor screens***

Very similar radiographic usage:

X-tube>patient>cassette>Reader> >re-use



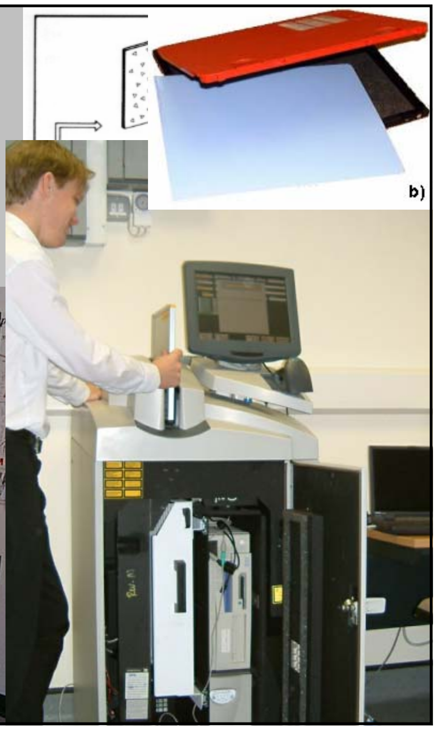
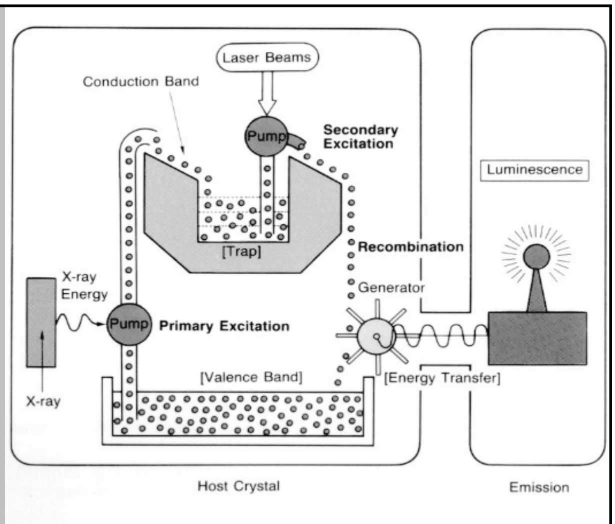
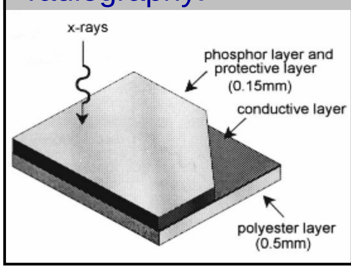


Photo-stimulated luminescence mechanism

The storage phosphor, usually made from **BaFX:Eu^{2+} (X=Cl, Br, I)** is contained within a cassette, similar in appearance to those used in film-screen radiography.





Eu^{2+} >> x-ray >> Eu^{3+} + *free e*

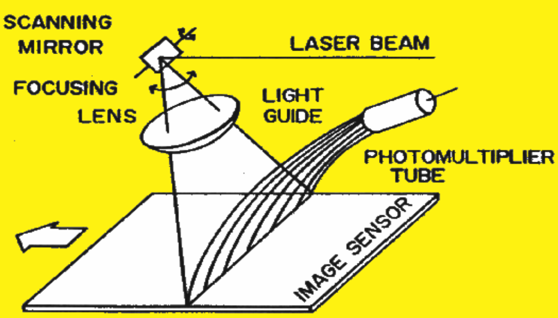
free e >> into bromine energy traps

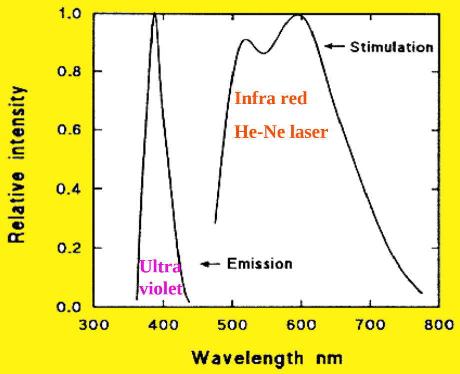
Eu^{3+} + *free e* >> IR laser >> Eu^{2+} + PSL (390 nm)

He-Ne laser stimulus infra-red (632 nm)

Eu characteristic radiation (PSL) – 390 nm (ultra-violet)

Fast scanning (PSL~0.8 ms)

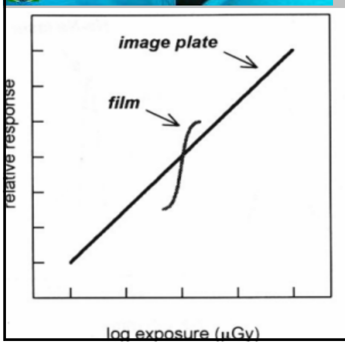




Commercial plates matrix:

1760x2140 (standard resolution):
2000X2510 (high resolution)

Resolution ~ 3 - 5 lp/mm (12 bits)



Storage-Phosphor (CR) against Film-Screen

-Much higher dynamics of CR (1:10000)

-Virtually no bad CR exposures (repetition)

-Very good contrast of CR

- Image processing in CR plus edge enhance

- Digital storage and retrieval of CR images

- Patient dose reduction

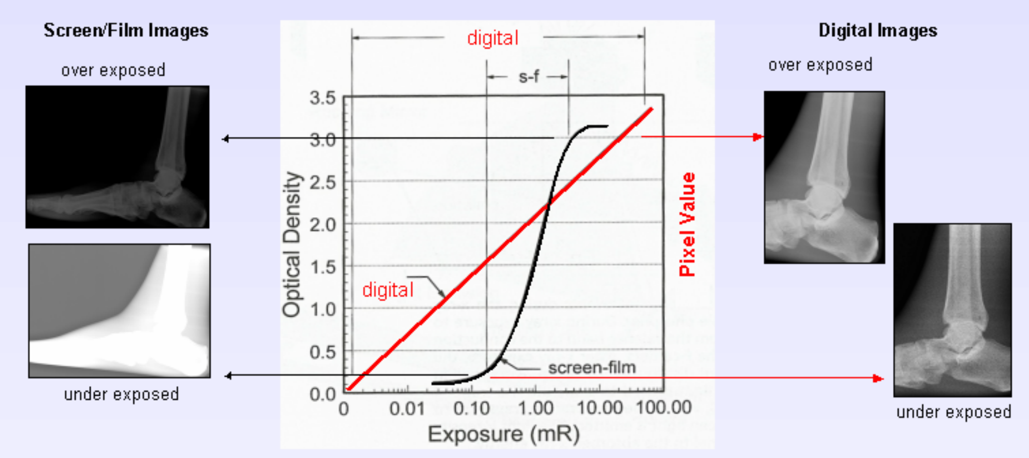
- Radiographic techniques preserved

- Film still with better resolution (mammo)

- Often CR images printed with laser imager

The wide dynamic range of CR systems is an advantage, but could easily lead to overexposure of patient

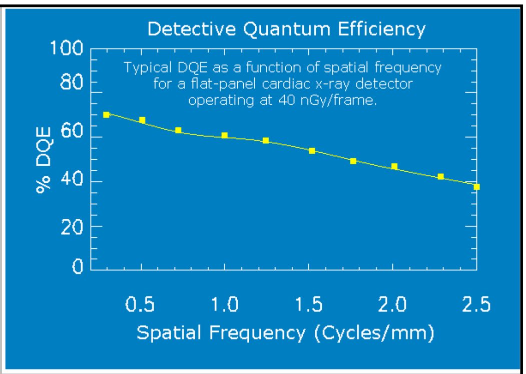
Optimization of CR procedures !



Source: A. Pascoal

The simplest definition of detective quantum efficiency can be stated in the formula. It shows that the DQE is the ratio of the output SNR squared to the input SNR squared.

$$DQE = \frac{SNR_{OUT}^2}{SNR_{IN}^2}$$

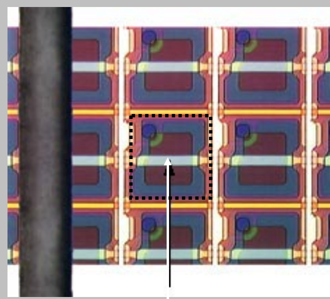


$$DQE_{PSP} = \frac{X_{abs}}{[1+CV(E)][1+CV(el)][1+CV(S)] \; + \; < g >^{-1}}$$

where: Xabs = fraction of incident x-ray photons absorbed in the phosphor layer
CV(E) = coefficient of variation of the x-ray energy absorbed in the phosphor layer
CV(el) = coefficient of variation in the number of trapped electrons for a given absorbed energy
CV(S) = coefficient of variation of the light signal emerging from the phosphor for a given number of trapped electrons
<g> = the average number of photoelectrons detected per absorbed x-ray

Direct Radiography– Flat Panel Detector (FPD)

- Indirect Conversion
  - Detector: Scintillator + a-Si diode (ex: CsI)
  - Readout: Thin-Film-Transistor
- Direct Conversion
  - Detector: Photoconductor (ex: a-Se)
  - Readout: Thin-Film-Transistor



hair

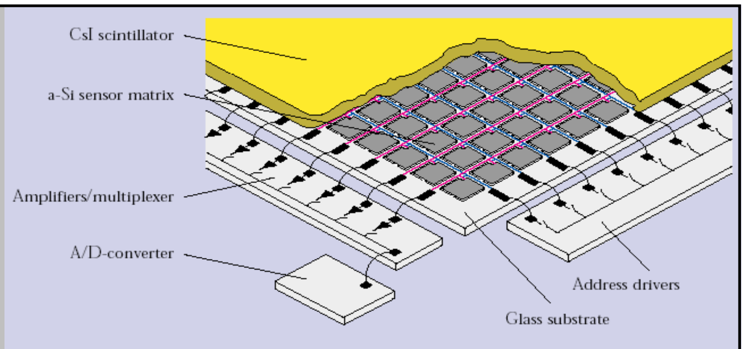
pixel
(0,10-0,20 mm)

Direct Digital Radiography with

Flat Panel Detectors

INDIRECT

(a-Si)



Amorphous Silicon matrix with array of sensors, each with own switching element – the readout is line-by-line (through address drivers), followed by amplification and A/D converter.

The X-ray sensitive converter is normally the needle-shaped CsI phosphor (used also in Image Intensifiers)

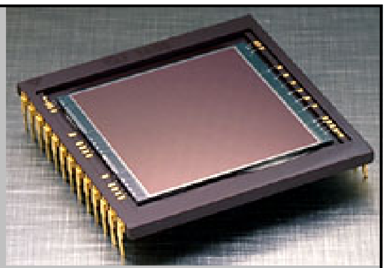
Detector size 43x43 cm, matrix 3000x3000 (pixel size 0.14 mm) > Resolution ~3 Lp/mm

DQE ~ 60% (twice the conventional film/screen)

Allows integration with Bucky table (anti-scatter)

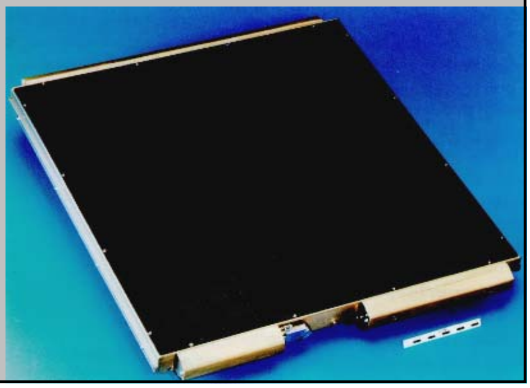
Very high workflow (patient flow)

Still quite heavy detector



Similar to CCD (mono-crystalline), but much larger due to a-Si.

Due to the rapid-sequence imaging, it is expected that in future the flat detector will replace the Image Intensifier TV systems in real-time examinations (fluoroscopy)

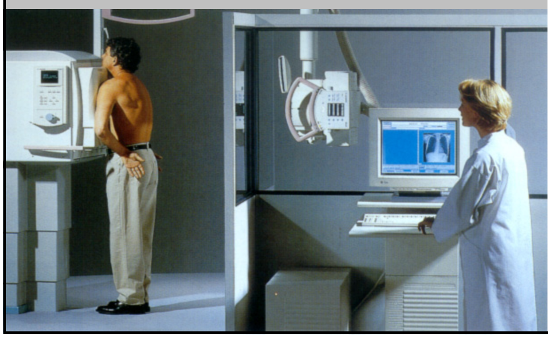


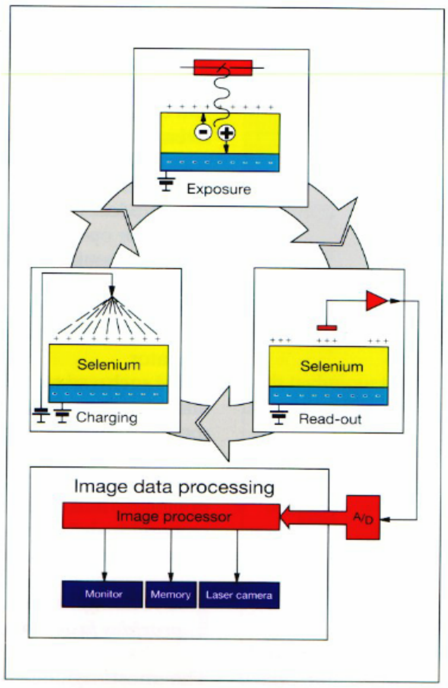
Flat Panel Detectors – **DIRECT (a-Se)**

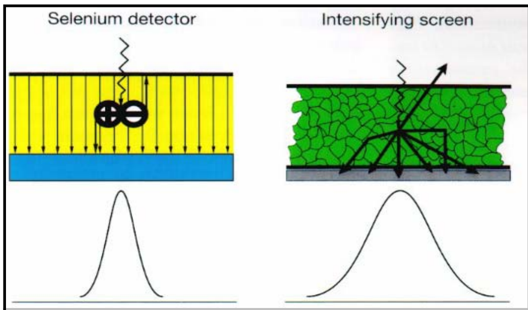
Direct Digital Radiography with **Selenium** Philips Thoravision

Uses amorphous Selenium (similar to xeroradiography)

Direct conversion of X-ray quanta into electrical charge – avoids noise from conversion





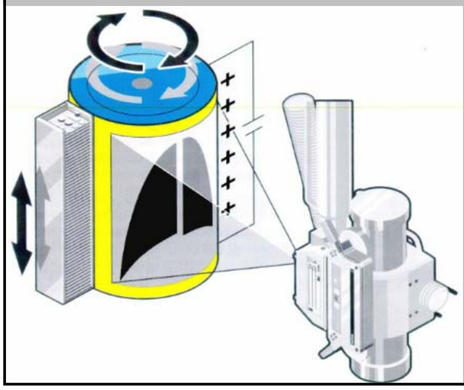


No intermediate light – the signal is transferred through electrical charge.

very good Signal/Noise Ratio (**$SNR \sim n^{1/2}$**)

Noise Equiv. Quanta **$NEQ = SNR^2$**

Detective Quantum Efficiency (DQE) - ideal DQE= 100% (the detector absorbs all impinging quanta)



Drum with 50 cm diameter

0.5 mm Selenium (43x49cm)

Read-our array of 36 probes

2000x2000 pixels (each 0.2mm, 14 bits)

Excellent contrast (wide dynamic)

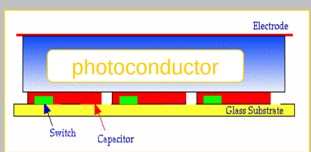
No transport of cassettes (fast radiography)

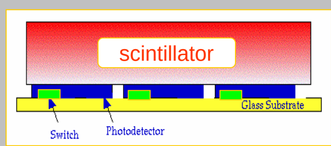
Directly linked to PACS

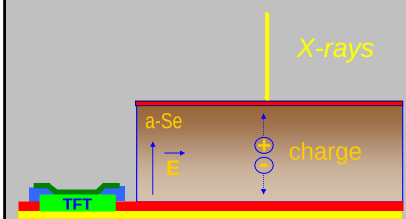
Direct Radiography - FPD

➔ Direct Conversion (photoconductor + TFT)

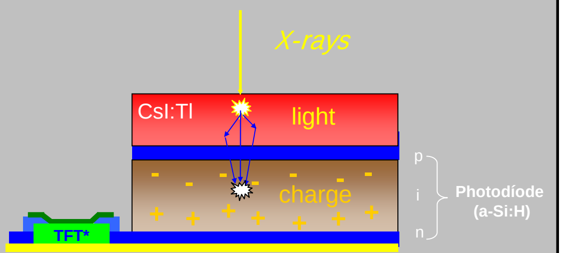
• Indirect Conversion (scintilator + a-Si/TFT)







*Thin-Film Transistor



*Thin-Film Transistor

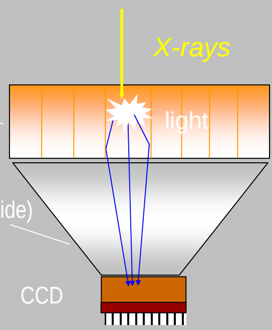
Direct Radiography– CCD

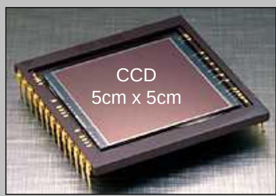
➔ Detector:

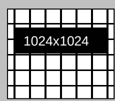
Scintillator

➔ Read-out:

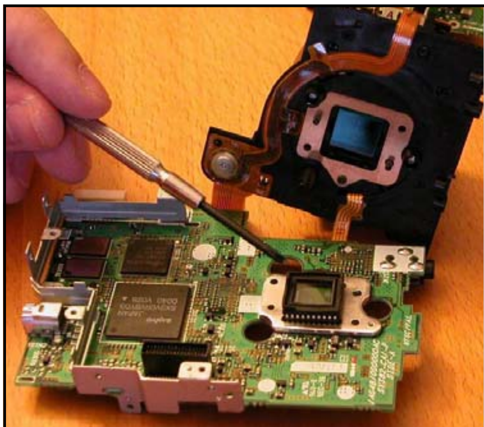
Charge-Coupled Device







pixel size
(50-100 µm)



01-start.jpg

Each pixels in a Digital camera includes a photo sensor (photosite) which collects and stores photons, and a CCD which transfers the signal to a readout register. The relative quantity of photons in each photosite cavity are sorted into various intensity levels.

Micro-lens between photosites collects max number of photons

The final image is processed by special imager to finalise the resolution, contrast and colour.

<u>The imaging chain includes:</u>

-Photo optics (+colour filters)

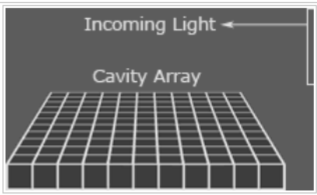
-Photo detectors (photosites)

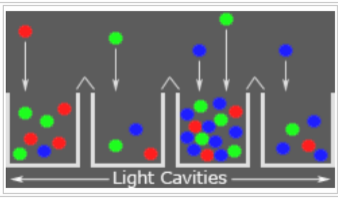
-Charge-coupled device

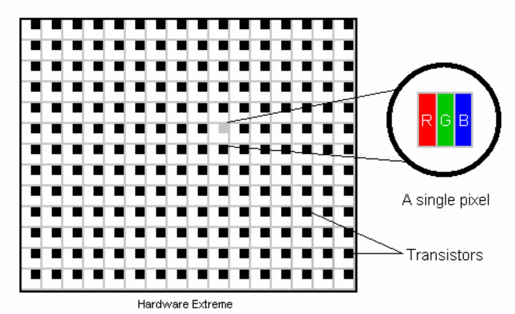
-Readout register + Imager

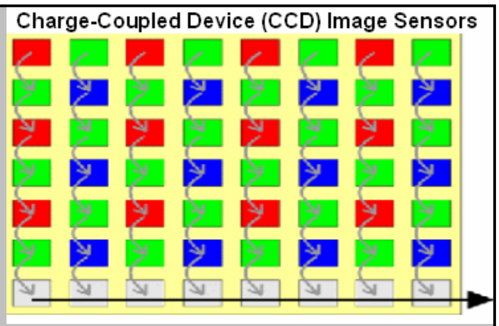
-Software (algorithm)

CCD basic principle

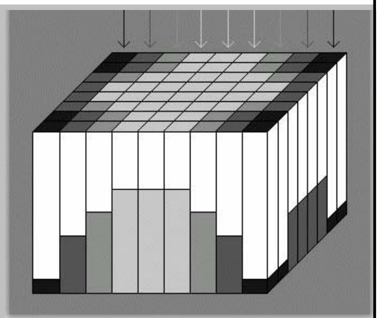


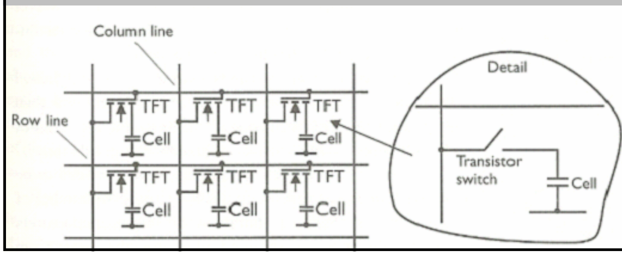






Fill factor = [light sensitive area]/[area of detector]

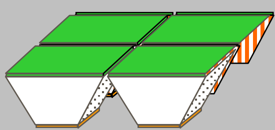




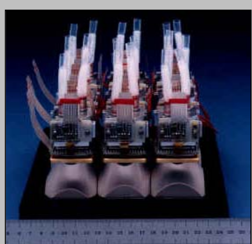
Software (algorithm)

Multiple CCDs

- "Mosaic" formation
- Requires optical link



Thoravision (Swissray)



Lorad

Imaging Dynamics

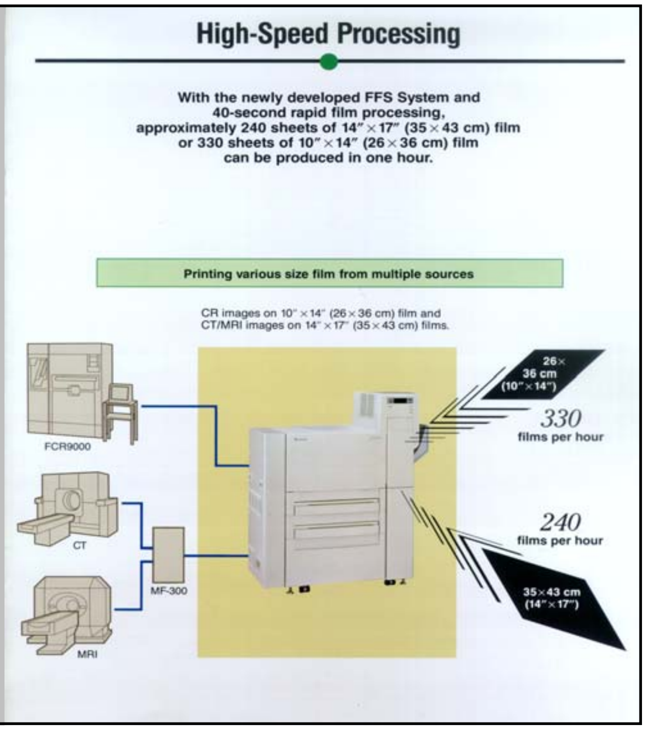
Ex. applications:

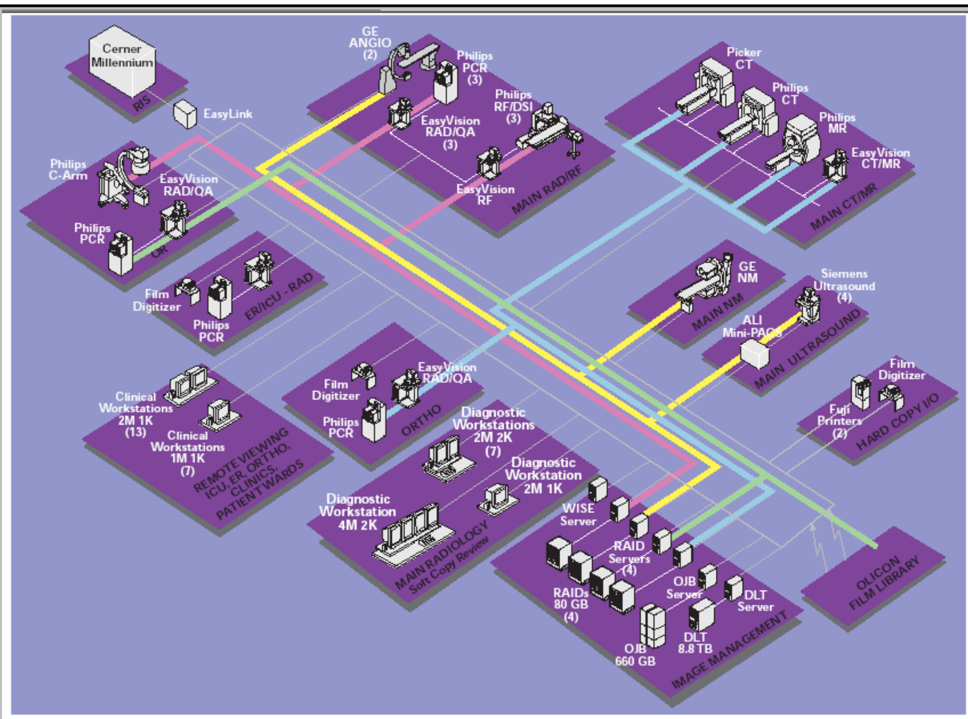
thorax (0,15 mm)
mamography (0,10 mm)
(full-field)

The digital image of CR allows archiving and share of images through PACS.

The hard-copy image of all these devices is still made on film (exposed with Laser Imager).

As in many places the diagnosis is still made from film, the final image quality will still depend on the film and imager….





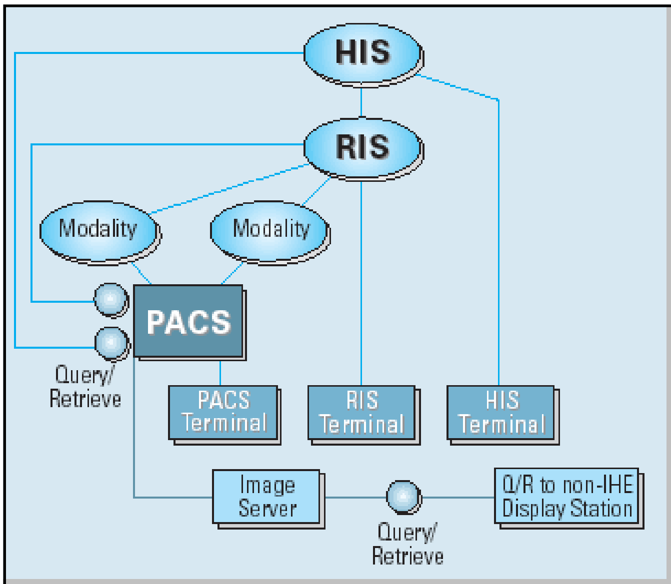
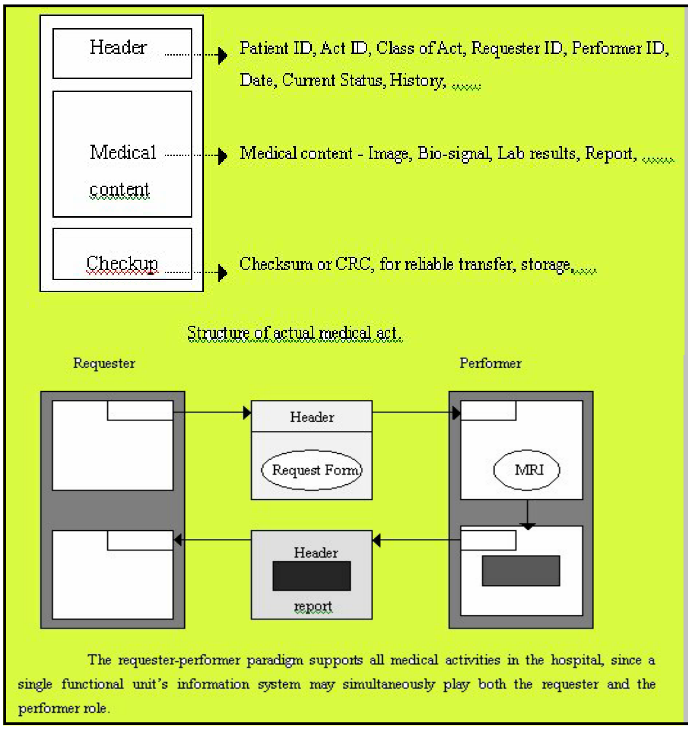


Image move (ATN): min 150 Mbits/sec;

Fibre opt. 600 Mbits/sec

Storage capacity: average 1800 Gbytes for 1 million images (based on 600 beds hospital)

Simple PACS architecture



DICOM standard facilitates interoperability of devices claiming conformance,...but does not guarantee, by itself, interoperability

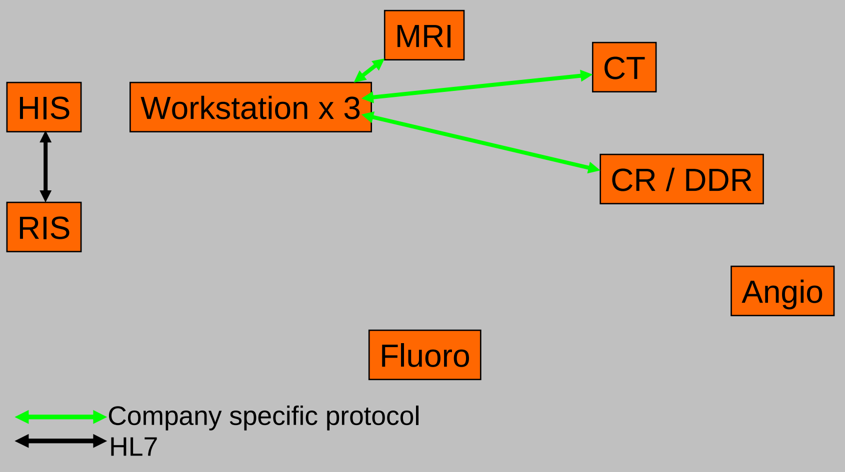
Promote communication of digital images;

Includes protocols, syntax and semantics;

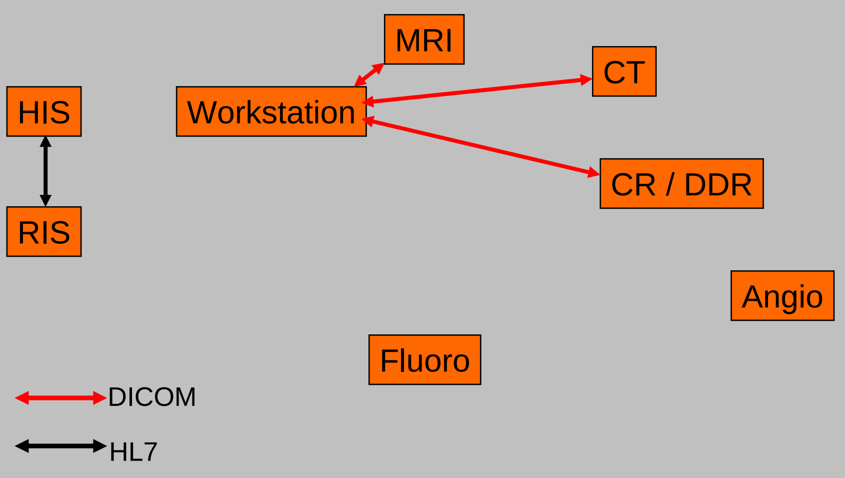
Provides a common format

DICOM: Digital Imaging and Communication in Medicine

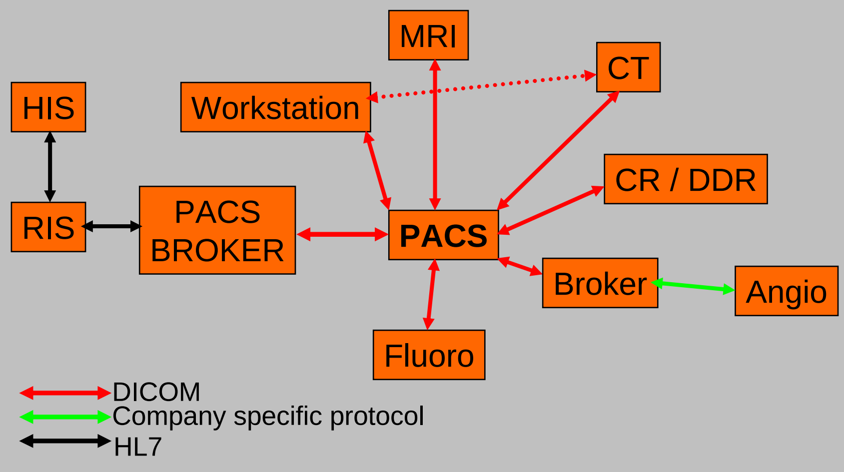
Where does DICOM fit in?

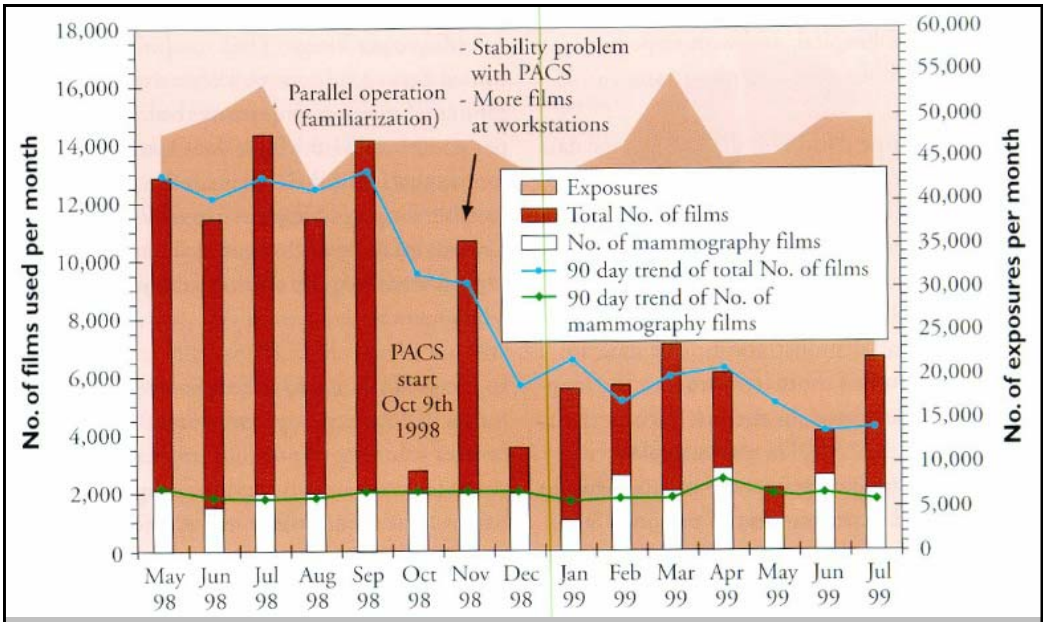


Where does DICOM fit in?

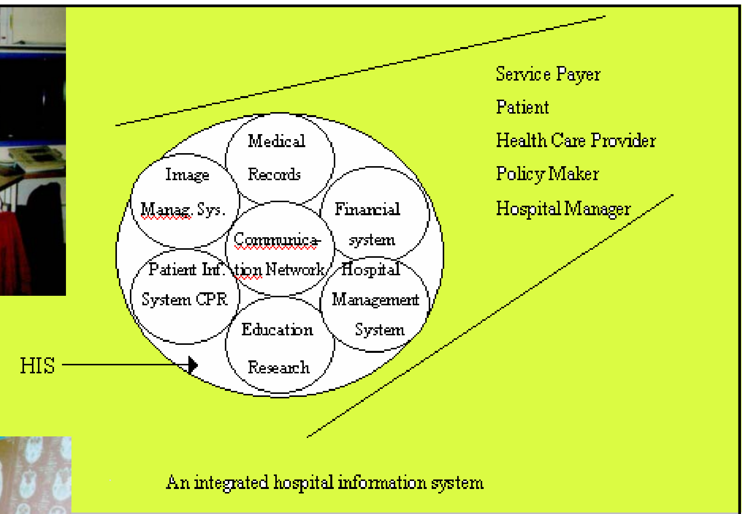


Where does DICOM fit in?





Usage of films – statistics pre and after PACS (Wehrle et al, Medica Mundi)



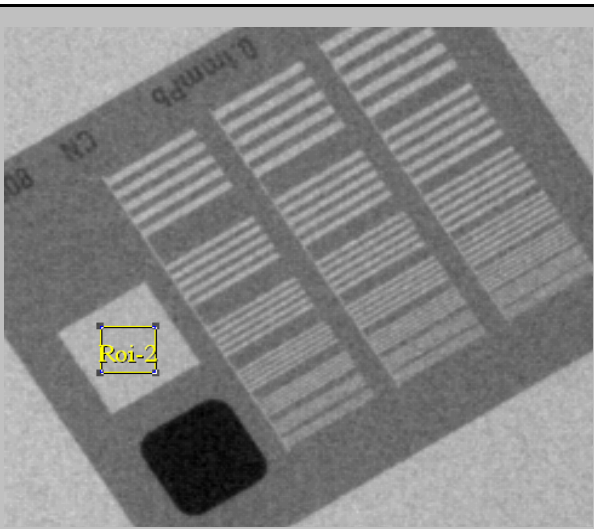
Integrated Hospital Information System

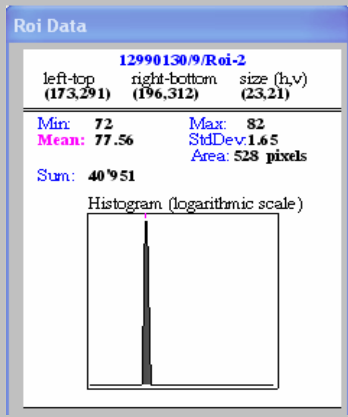
Physical aspects of image quality

- SNR - Signal-to-noise ratio. The ratio of noise to picture signal information (ICRP 93 Glossary).
- In the context of the signal detection theory, the SNR is proportional to a ratio of the magnitude of the difference between the mean values of some quantity under two conditions that are to be distinguished, to a measure of the magnitude of statistical variation in that difference.

$$SNR= [mean(background)-mean(ROI)] / \{1/2[std^2(ROI)+std^2(background)]\}^{1/2}$$

ROI = Region of interest

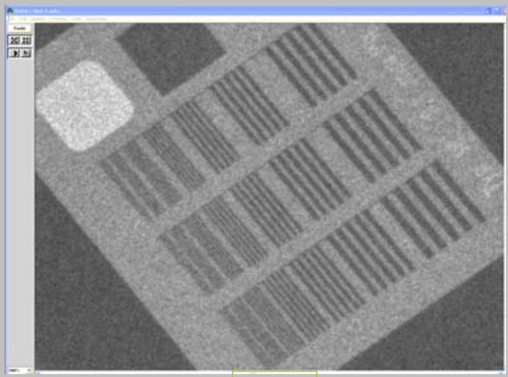




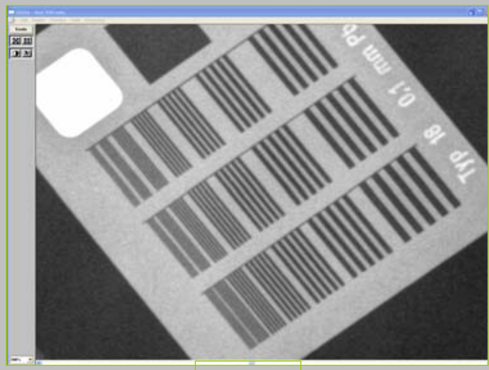
White = more radiation to the detector
Black = less radiation to the detector
Noise (absolute): 1.65
Noise (relative): 1.65 x 100 / 77.56 = 2.1%

Correlation of image parameters with dose

- ➔ The noise typically decreases when radiation dose increases.
- ➔ SNR is proportional to the square root of the average number of x-ray quanta and typically improves when increasing dose.
- ➔ Contrast improves for low kVp X ray beams (low energy photons).



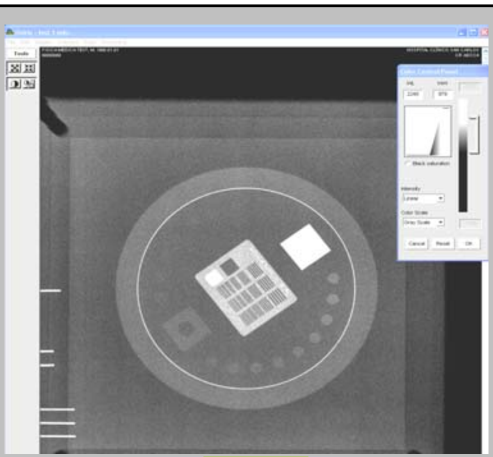
1 mAs



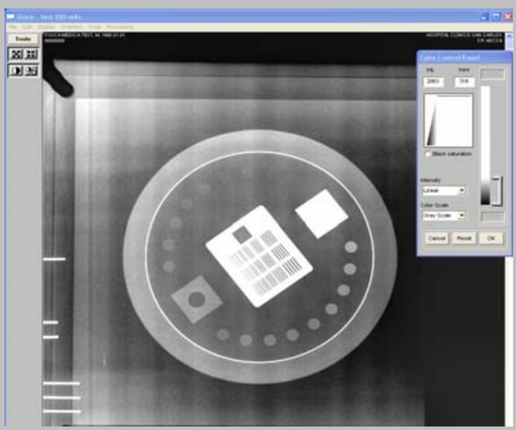
100mAs

CR Agfa system:
Left: 1mAs   - 14 spatial resolution groups
Right: 100 mAs - 16 spatial resolution groups

Noise in a digital image produces poor spatial resolution and reduces contrast



1mAs



100 mAs

CR Agfa system:
Left: 1mAs - 11 circles low contrast
Right: 100 mAs -16 circles low contrast

Example of clinical images obtained with two different levels of dose and noise

Relative dose level (Agfa system) 1.15 (image too noisy)

Relative dose level (Agfa system) 1.87 (image with enough quality) (with approx. 5 times more dose at the entrance)

Effect of the post-processing

- The standard post-processing parameters offered in some CR workstations includes the noise reduction and the edge enhancement.
- Some examples are shown for the Agfa post-processing called "MUSICA" (Multi Scale Image Contrast Enhancement). This is the basic principle of MUSICA:
  - contrast enhancement irrespective of feature size.
  - difference with respect to spatial frequency band filtering.

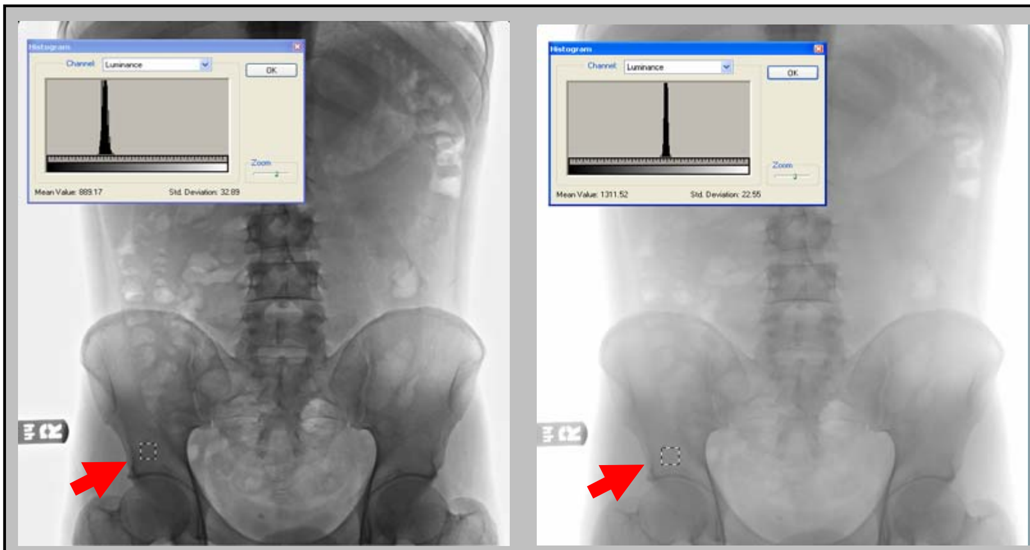
Standard image

Noise reduction

Edge enhancement

Examples of different post-processing using Agfa CR software (MUSICA)

Same image, two different postprocessings?

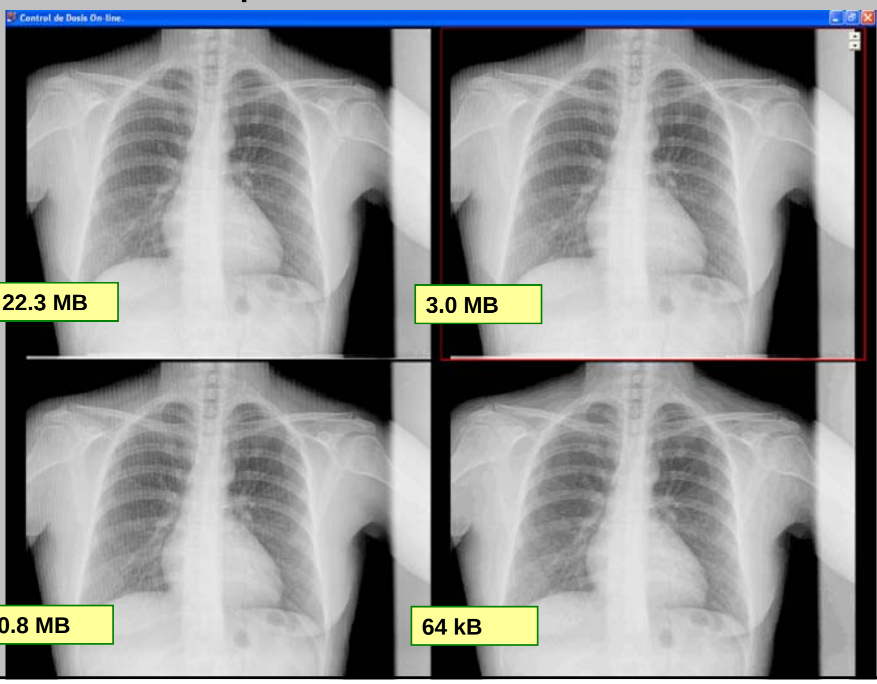


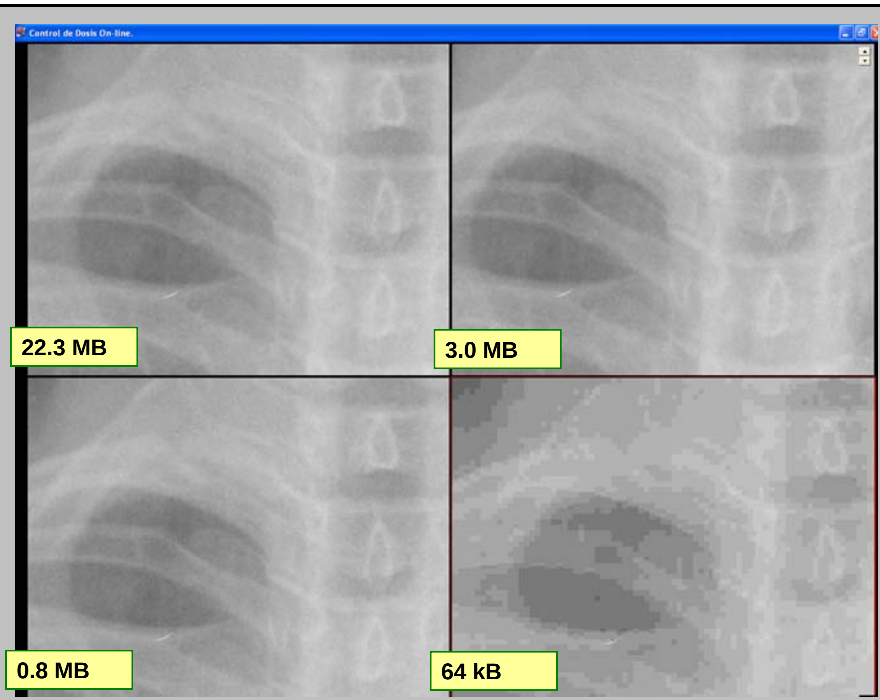
Same image, two different postprocessings?

Noise = 33/889 = 3.7%

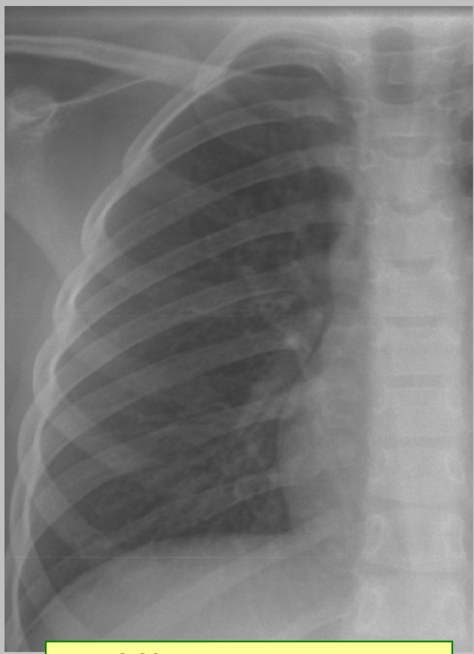
Noise = 23/1312 = 1.8%

Different compression levels

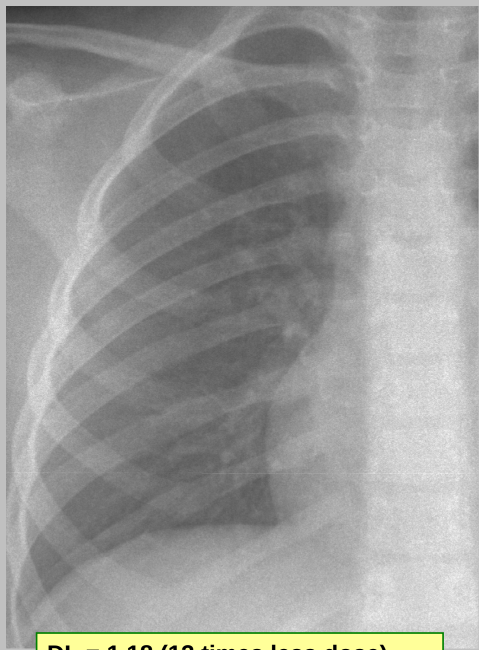




Patient Dose
More dose better image quality



DL = 2.30



DL = 1.18 (13 times less dose)

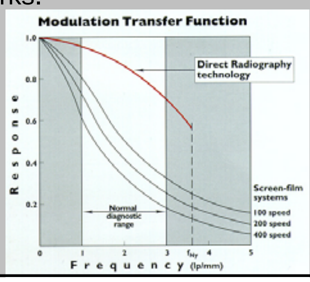
Digital radiography and digital fluoroscopy. Differences with conventional

Advantages

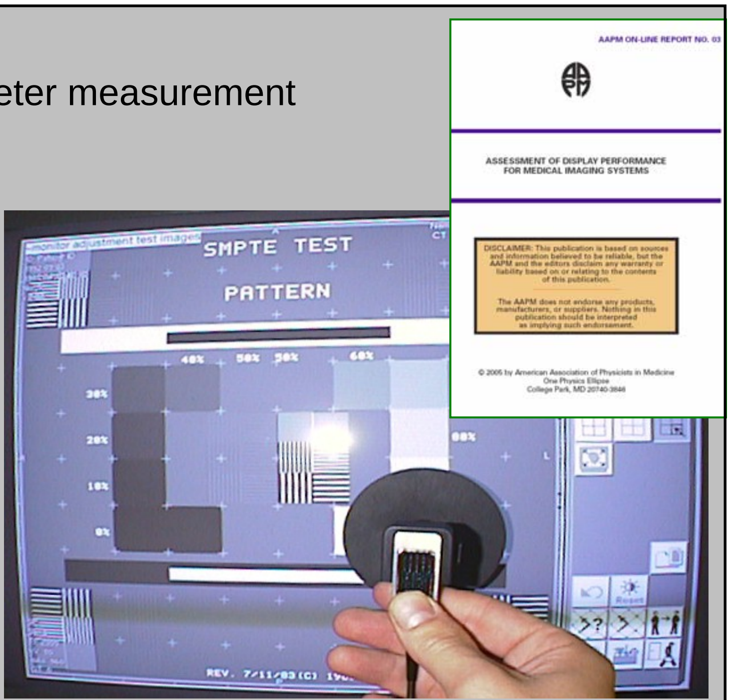
- More information can be obtained from the image (change of window and level, magnification, etc).
- Wide dynamic range (more tolerance to different dose values).
- Easy archive and transmission by networks.

Disadvantages

- Over exposures could not be noticed.
- Very easy to delete the files of the bad quality images.
- A tendency to obtain more images than necessary could occur.
- Audit of relevant radiation protection parameters can sometimes be difficult.



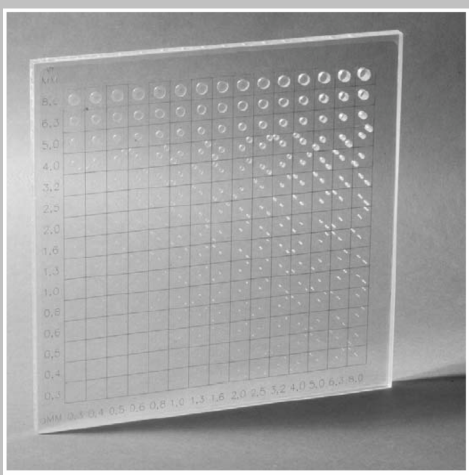
Monitors: photometer measurement

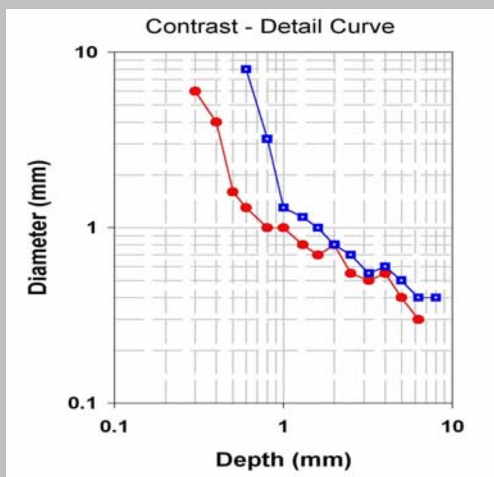


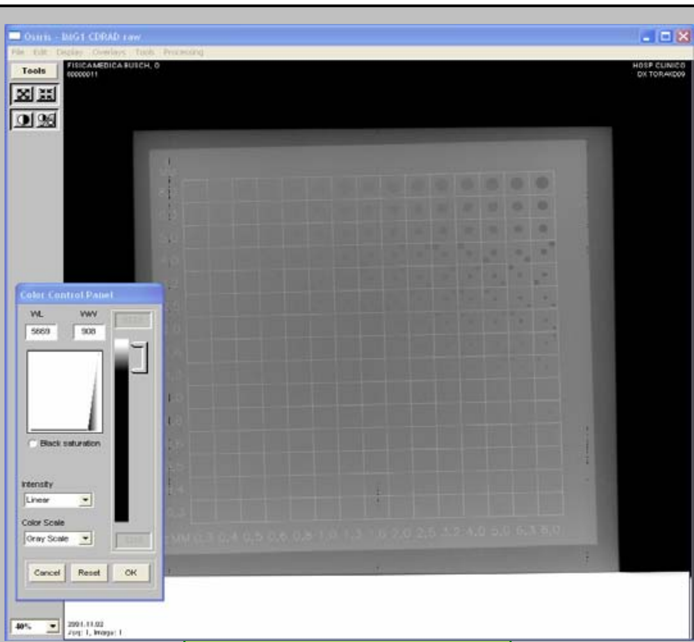
Poor conditions of the visualization monitor (e.g. lack of enough brightness or contrast, poor spatial resolution, etc) can require repetitions of exposures.

Such a visualization monitor produces sub-standard image quality.

CDRAD phantom





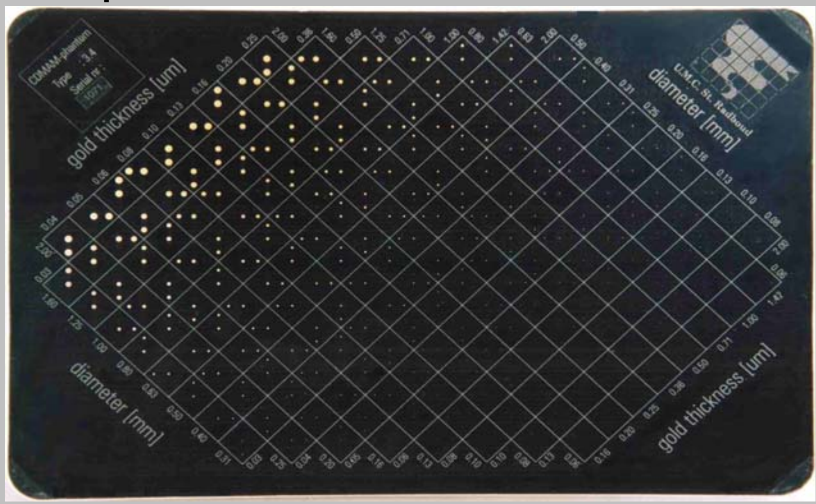


IMG1 CDRAD raw;

CDRAD phantom

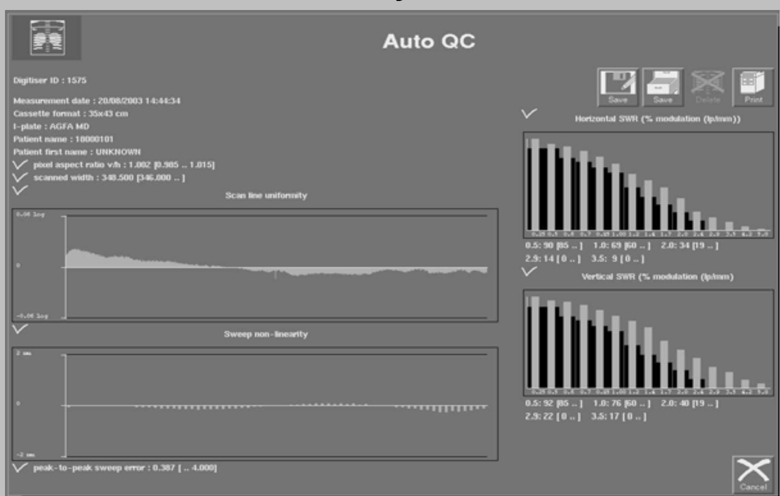
- The image shows 225 squares, 15 rows and 15 columns. In each square either one or two spots are present, being the images of the holes.
- The first three rows show only one spot, while the other rows have two identical spots, one in the middle and one in a randomly chosen corner.

CDMAM phantom



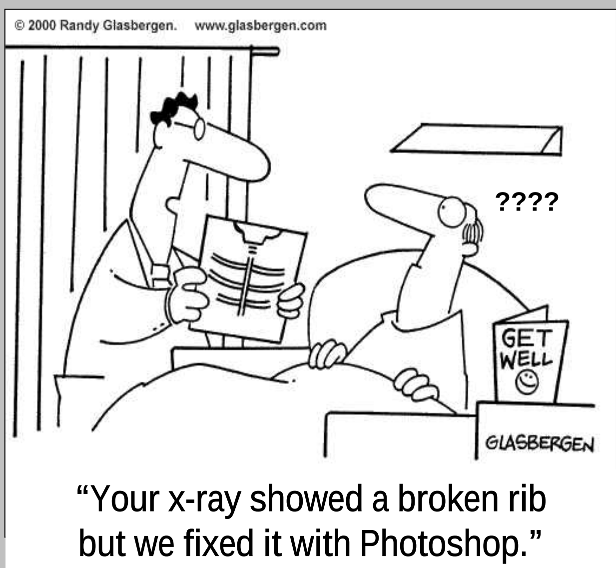
AutoQC: resolution, uniformity, linearity

➔ Automatic evaluation made by the software



CR image with artifacts. Two different linear artifacts. One derived from a fault in the digitiser (upper one), the other from the PSP

The future…



Acknowledgments

- Figures from Agfa, Siemens, Philips, GE, Fuji and Toshiba systems have been used.
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- Images from Prof. Perry Sprawls, Dr. Ramon Sanchez-Jacob, Dr. Eliseo Vano-Galvan, Anchali Krisanachinda,Ph.D, Petcharleeya Suwanpradit, and Ana Pascoal have been used
- Images from EMERALD materials and Dr S Tabakov have been used

Physical aspects of image quality