The Abdus Salam International Centre for Theoretical Physics

United Nations Educational, Scientific and Cultural Organization

IAEA International Atomic Energy Agency

2166-Handout

College on Medical Physics. Digital Imaging Science and Technology to Enhance Healthcare in the Developing Countries

13 September - 1 October, 2010

Basis of Computed Radiography & PACS

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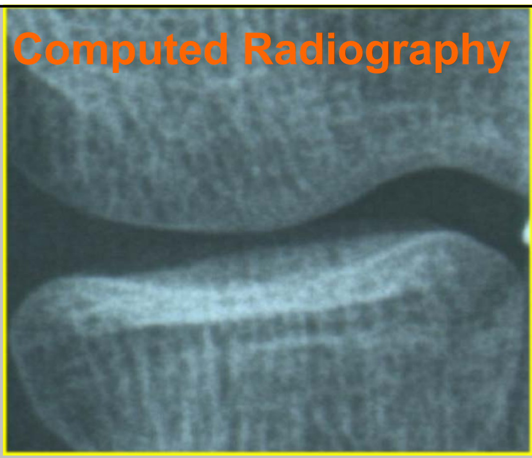
Basis of Computed Radiography & PACS



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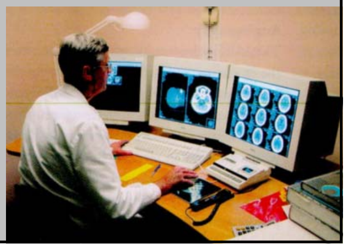


Digital	Film-screen

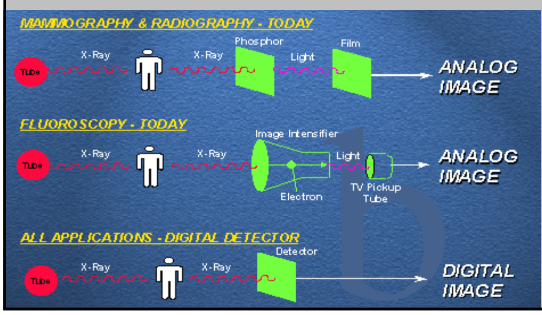


Image comparison and image transfer through various systems

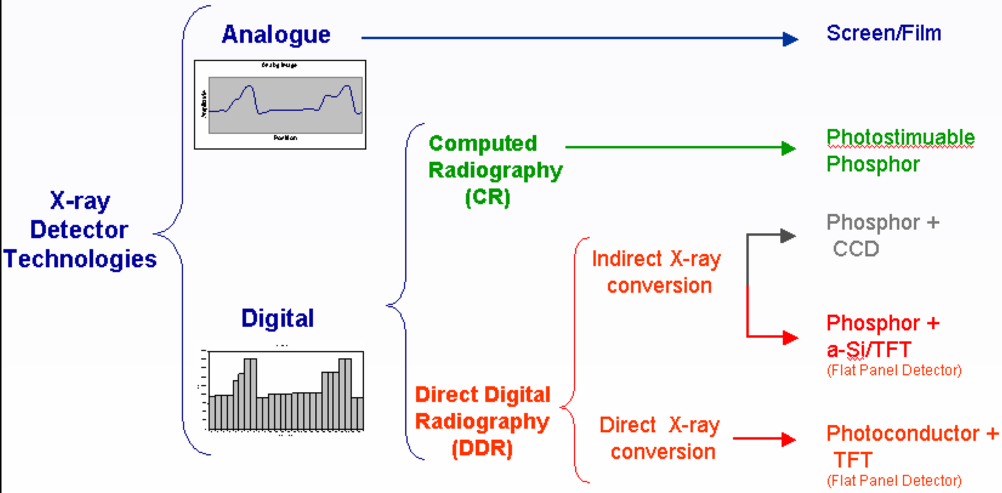


Fig.3 – Current X-Ray detector technologies for x-ray diagnostic imaging

Source: A. Pascoal

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

Very similar radiographic usage:

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

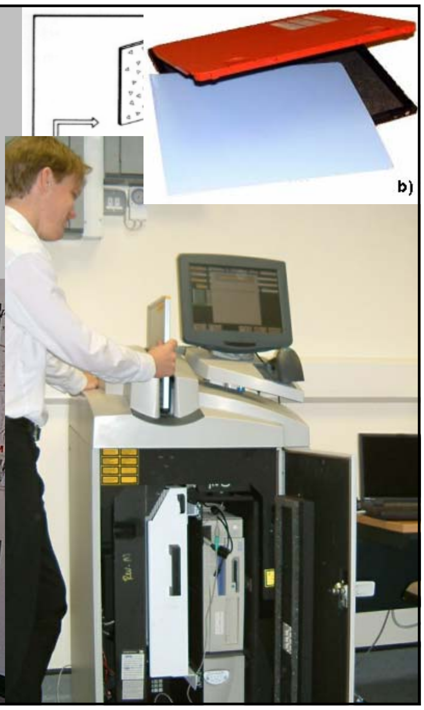
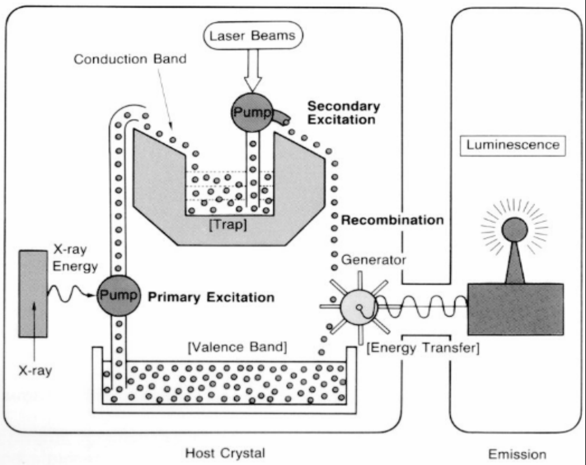
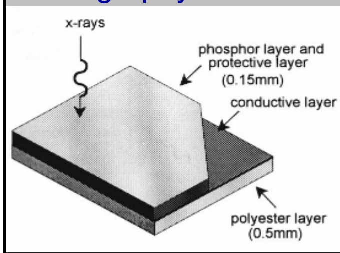


Photo-stimulated luminescence mechanism

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*

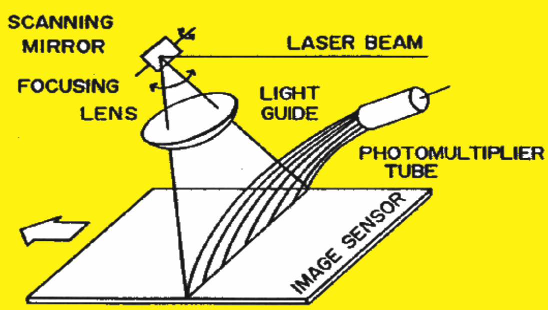
free e >> into bromine energy traps

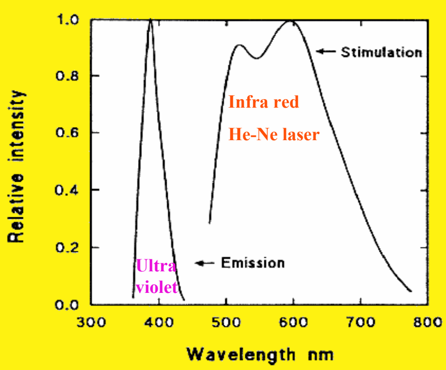
Eu^{3+} + *free e* >> IR laser >> Eu^{2+} + PSL (390 nm)

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

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

Fast scanning (PSL~0.8 ms)

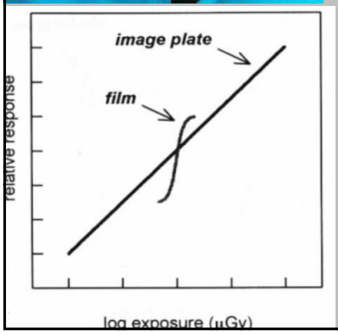




Commercial plates matrix:

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)

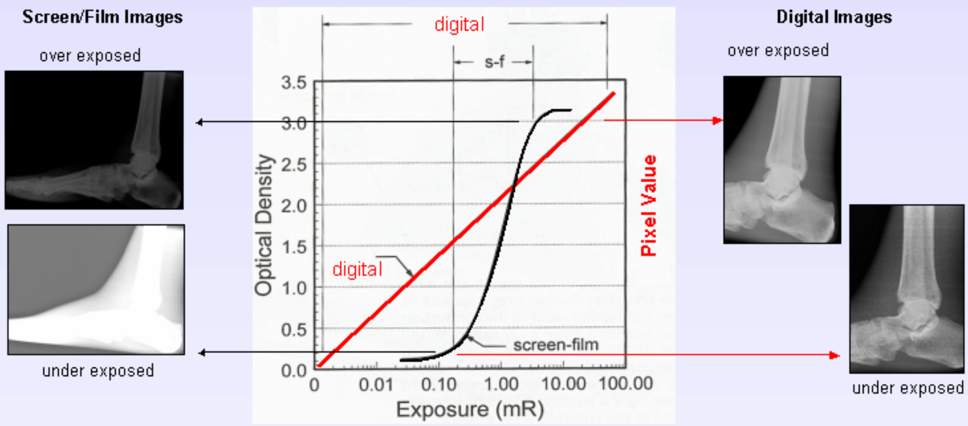
-Virtually no bad CR exposures (repetition)

-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

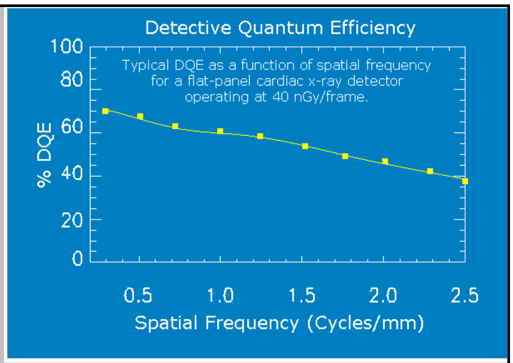
Optimization of CR procedures !



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.

$$DQE = \frac{SNR^2_{OUT}}{SNR^2_{IN}}$$

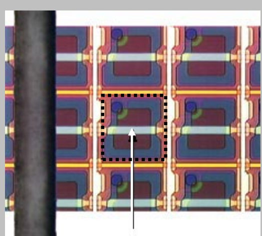


$$DQE_{PSP} = \frac{X_{abs}}{[1+CV(E)][1+CV(el)][1+CV(S)] \; + \; <g>^{-1}}$$

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

DIGITAL Radiography (DDR) – 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)

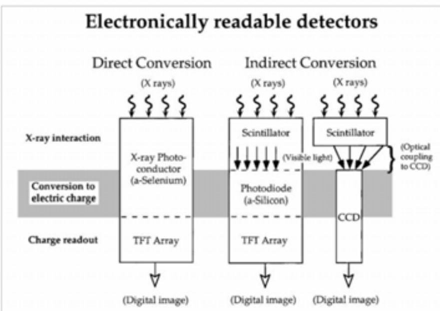


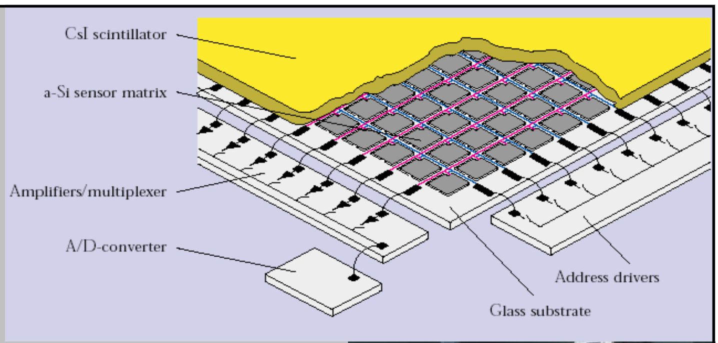
Figure 1. Direct and Indirect Conversion of the x-rays to electric charge (source: Chotas H.G. et al, Principles of Digital Radiography with Large-Area, Electronically Readable Detectors: A Review of the Basics [22])

Digital Radiography with

Flat Panel Detectors

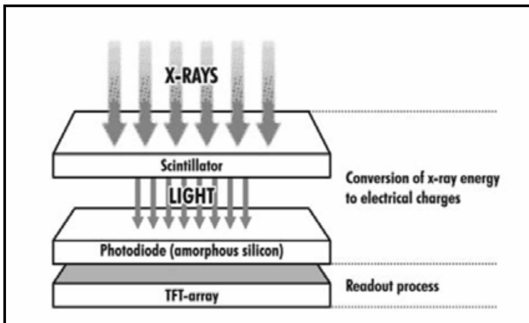
INDIRECT

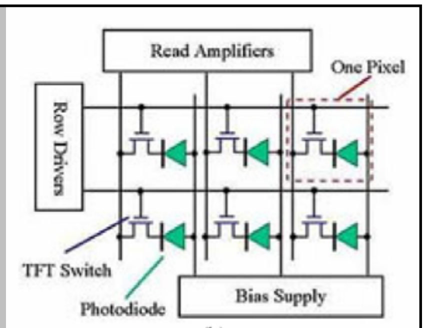
(a-Si diode – only for the electronics)



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

The light is detected by a matrix with array of sensors (a-Silicone diodes), each with own switching element – the readout is line-by-line (through address drivers), followed by amplification and A/D converter.





Indirect detector principle; sensitivity and spatial resolution of phosphor

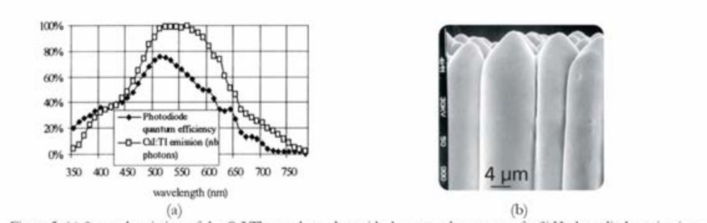
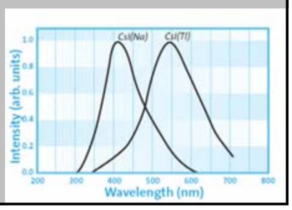


Figure 5: (a) Spectral emission of the CsI:Tl crystal matches with the spectral response of a-Si:H photodiodes using in x-ray indirect detection systems [9]. (b) electron micrograph of the needle shaped CsI:Tl scintillation crystal [10].



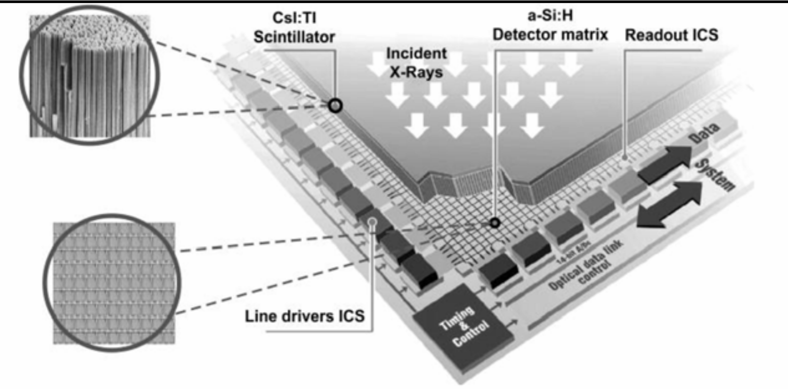


Figure 2: Schematic cross-section of an a-Si digital detector[2] (ICS = integrated circuits)

Indirect detector – (CsI Phosphor + Si Detector)

- Spatial resolution depends on pixel size and phosphor blur

-Due to the blur (slight) the influence of high freq. signals is reduced (hence less noise) and better contrast

- best DQE for all Digital detectors

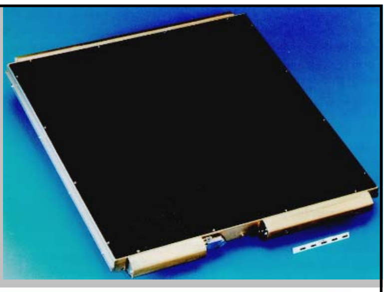
Detector size 43x43 cm, matrix 3000x3000 (pixel size 0.14 mm) > Resolution >3 Lp/mm

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

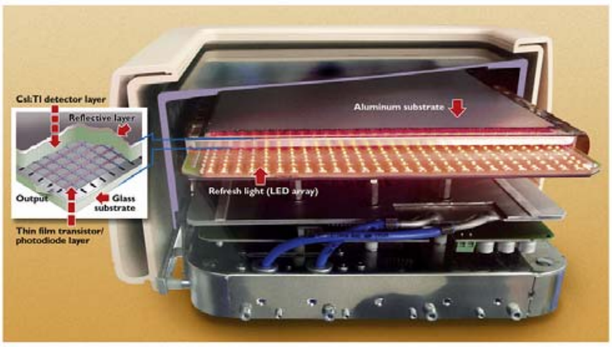
Allows integration with Bucky table (anti-scatter)

Very high workflow (patient flow)

Still quite heavy detector



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)



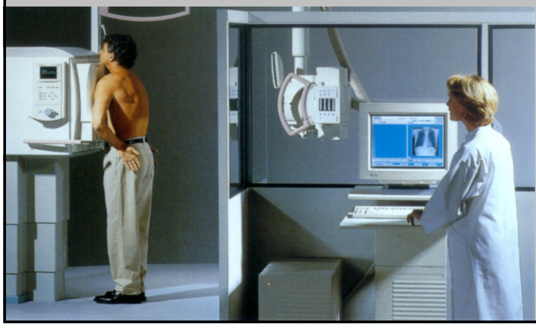
2 - Cross section through a flat panel detector. Source: Cowen et al, 2008

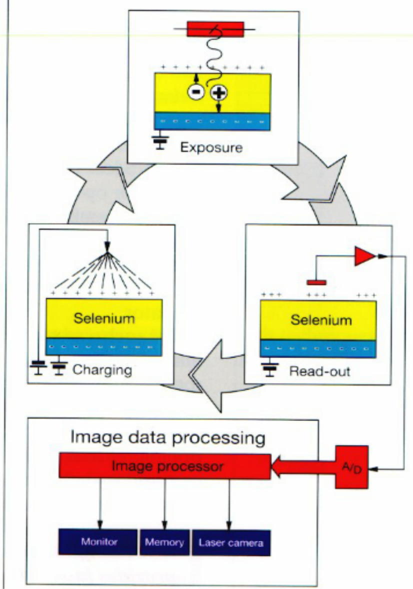
Flat Panel Detectors – **DIRECT** (a-Se)

Digital Radiography with **Selenium** Philips Thoravision

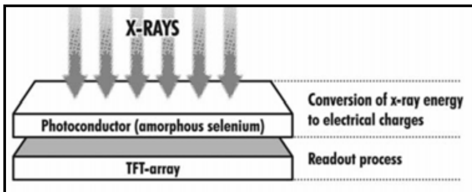
Uses amorphous Selenium (similar to xeroradiography)

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

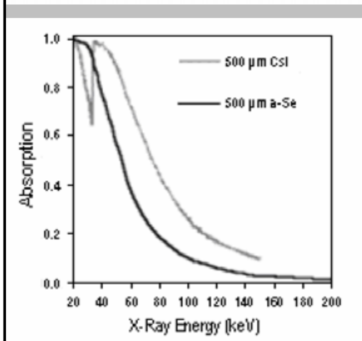




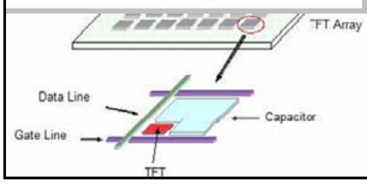
Directly linked to PACS



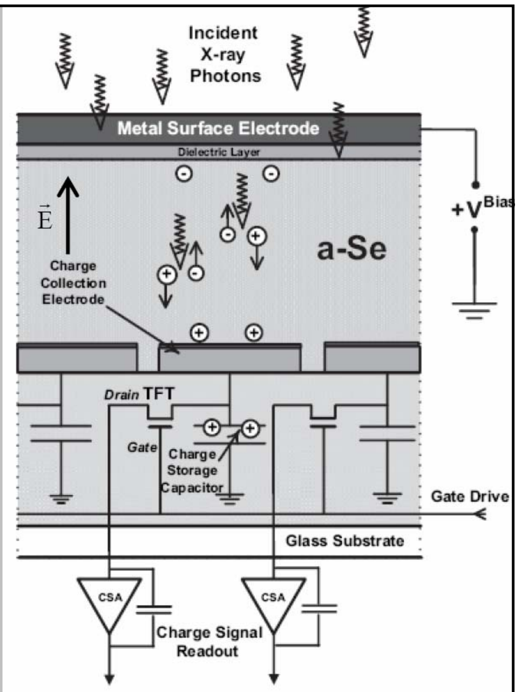
a-Se principle

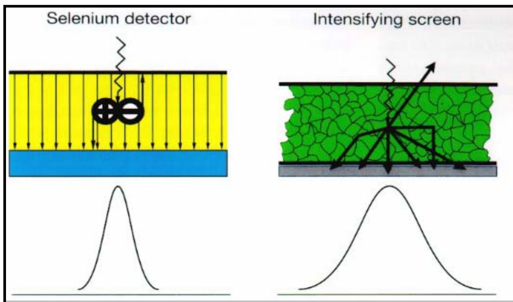


Capacitors and gates



Fill factor



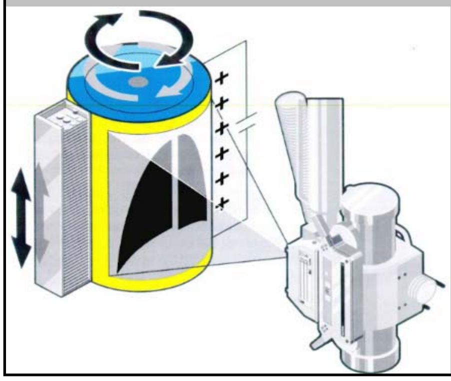


No intermediate light – the signal is transferred through electrical charge.

very good sp. Resolution (less spread)- Theoretically up to 20lp/mm for a-Se

but also less absorption in a-Se and hence reduced DQE.

The decrease fo signal leads to reduced Signal/Noise Ratio (and less contrast)



Drum with 50 cm diameter

0.5 mm Selenium (43x49cm)

Read-our array of 36 probes

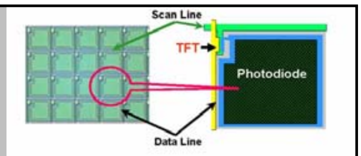
2000x2000 pixels (each 0.2mm, 14 bits)

Excellent contrast (wide dynamic)

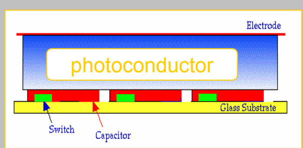
No transport of cassettes (fast radiography)

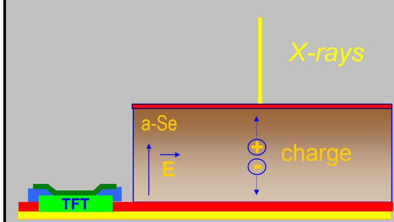
Directly linked to PACS

Direct Radiography – FPD overview



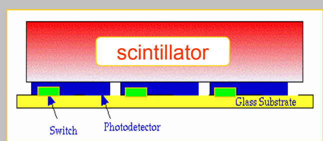
- Direct Conversion (photoconductor + TFT)

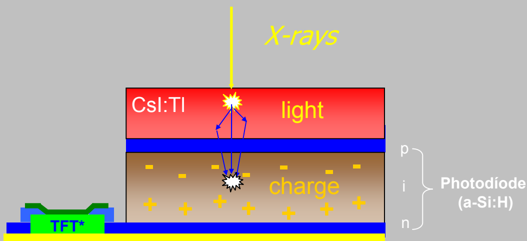




*Thin-Film Transistor

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





*Thin-Film Transistor

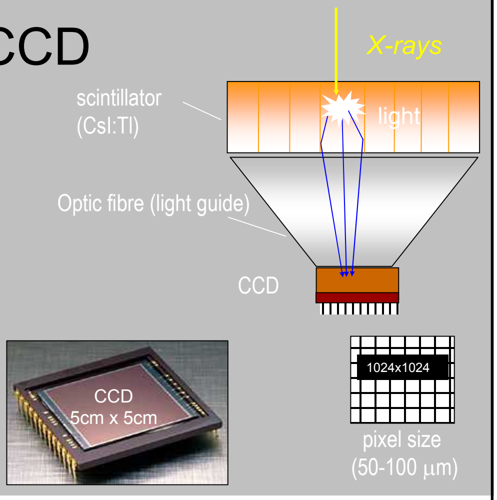
Direct Radiography– CCD

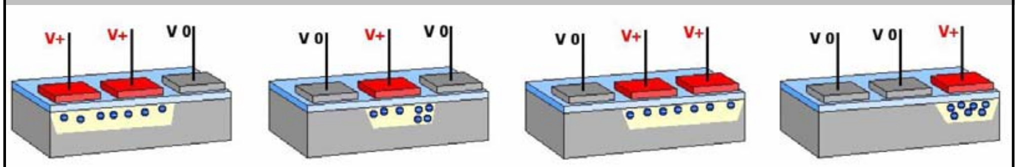
➔ Detector:

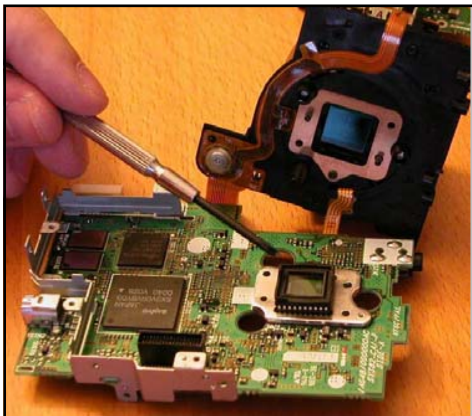
Scintillator

➔ Read-out:

Charge-Coupled Device







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.

The imaging chain includes:

-Photo optics (+colour filters)

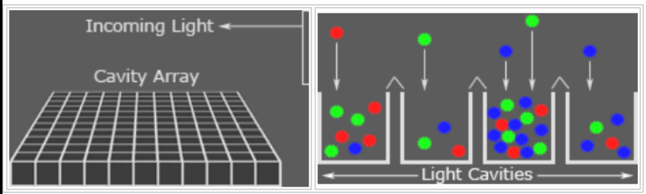
-Photo detectors (photosites)

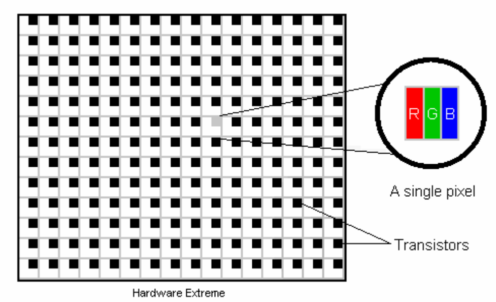
-Charge-coupled device

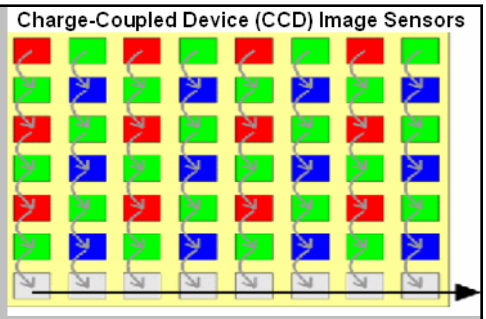
-Readout register + Imager

-Software (algorithm)

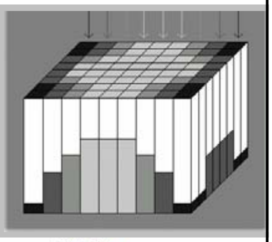
CCD basic principle



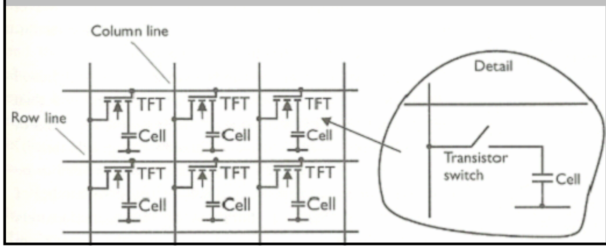


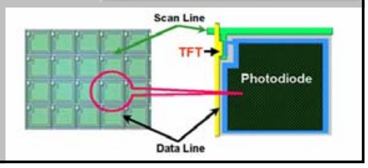


Software (algorithm)



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





CCD systems coupling

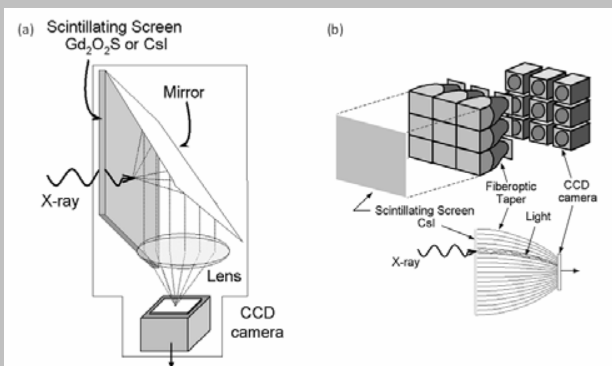
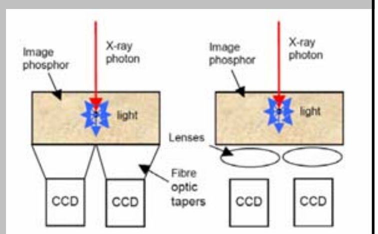
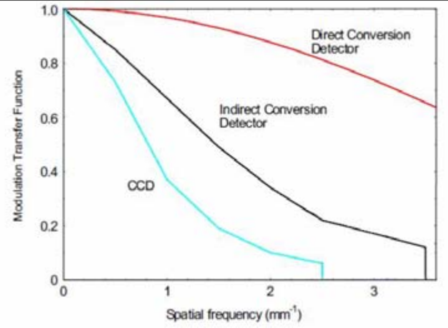


Fig.4: (a) A lens-coupled CCD system (b) A fibre optic-coupled CCD system



An array of CCDs covers the area of the image, the light coupling leads to light loss, hence decrease of SNR and reduced sp. Resolution and DQE (fibre optic is better)

Figure 5. Fibre optic tapers
(source: www.edmundoptics.com



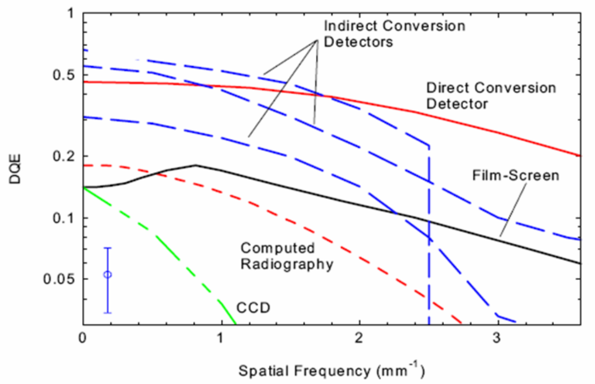
Comparison of Digital Radiography systems

Best Sp. Resolution – a-Se

Best DQE / SNR - a-Si (higher DQE = better low contrast res.)

Best Cost - CR

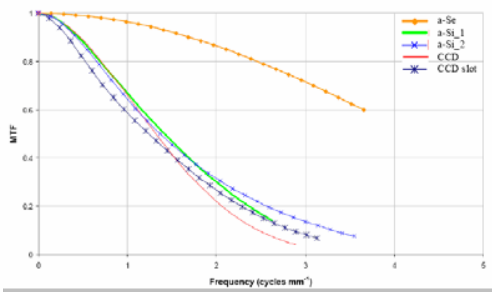
CCD - ??

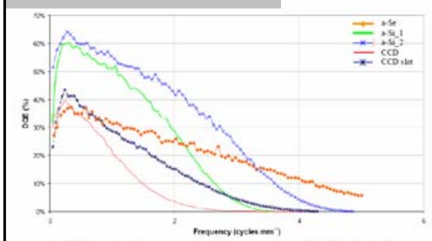


Parameters of Digital Radiography systems (different types – real figures)

Table 1: Detectors characteristics. [19]

Detector name	aSe	aSi _1	aSi_2	CCD	CCD scan
Technology	a-Se/TFT array	a_Si:H photodiode/ TFT array	a_Si:H photodiode/ TFT array	CsI scintillator optically coupled to 4 CCDs	Slot scanning system with CsI directly coupled to CCD array
X-ray absorber	a-Se	CsI:Tl	CsI:Tl	CsI	CsI:Tl
Thickness of material (µm)	500	unknown	550	600	500
Image area (cm)	35.6×42.7	41×41	43×43	35×43	44×44
Pixel array	2560×3072	2022×2022	3001×3001	2048×2560	2720×2720
Pixel size (µm)	139	200	143	169	162



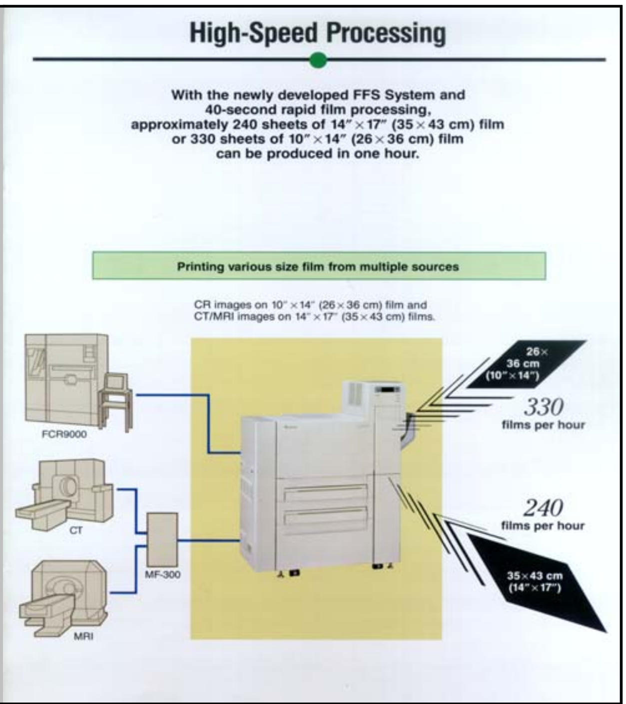


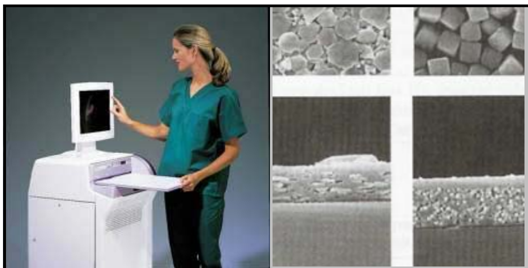
Average DQE of the various detectors acquired for x-ray settings of 70 keV and 4 µGy exposure.

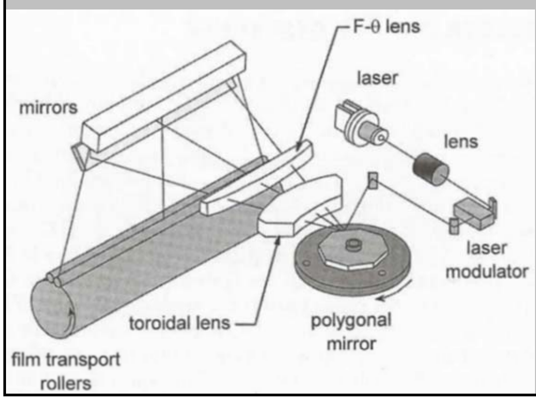
The digital image of CR allows archiving and share of images through PACS.

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….







Dry Laser Imager (Laser Camera)

Dry laser imager (78-micron laser spot spacing, 325-dpi resolution). Direct laser scanning over X-ray film. Laser is often infrared solid state laser of 820nm. In the image the Laser source intensity is constant and it is later modulated to create the necessary light intensities (grey levels).

Throughput of >100 films per hour. Image quality with selectable 50 or 100 micron pixel size, 12 bit gray-scale resolution, automatic density control and image interpolation technology. The imager can work with normal X-ray film and wet-type film processing. It allows PACS connection. The film for lasers is with cubic grains and not with tabular grains (as in screen-film)

Variations in image quality resulting from a not always optimal wet film development are frequent. A newly developed thermographic film developer for laser films without liquid powdered chemicals, is environmentally preferable and reducing operating costs.

X-ray film with dry processing methods:

1. Adherographic – laser sensitive adhesive layer + imaging layer (carbon particles), both sandwiched between 2 polyester sheets. When the laser beam scans the dry-film it causes the adhesive layer to take carbon and stick it to the polyester sheet. As a result there are 2 sheets with positive and negative image. The first is coated and used as film, the other is disposed. The adhesion process is binary and the grey tone (nuance) is produced by dithering. Normally a cell of 16x16 pels makes a pixel with 256 grey levels. This requires very thin laser and small pells (5 μm) – 16x5 = 80 μm pixel = 6.25 lp/mm

2. Thermal - a combination of silver behanate and silver halide over polyester. The scanning laser beam triggers "thermal developing process" producing a "true" gray scale. However there is no fixer – I.e. the undeveloped silver halide crystals remain on the film, what makes it thermally unstable.

* These imagers could have less grey levels as they are used with img. methods using "window"

SONY D71XR Laser Imager (Direct Thermal Printing)

Resolution: 300 dpi (with blue thermal film)

Gradation: 256 grey levels (memory: 16 MB)

Effective Print Pixels: 2743 x 2320 dots

Print Area: (232.2 x 196.4 mm); Interface: SCSI

Printing Time: Approx. 45 seconds

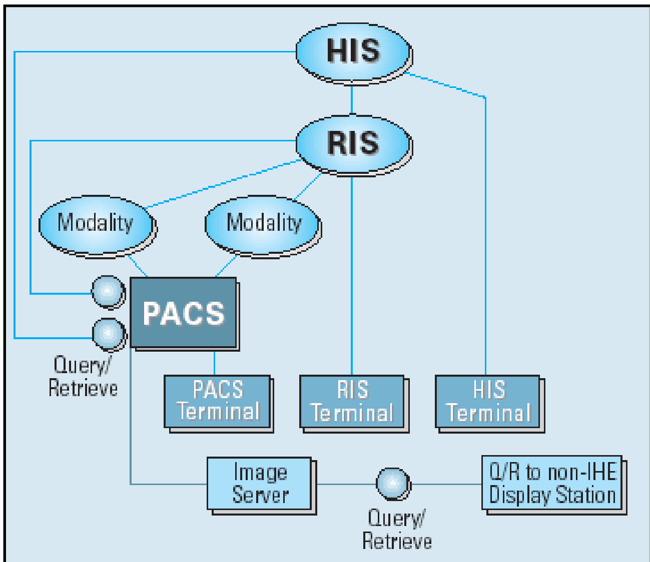
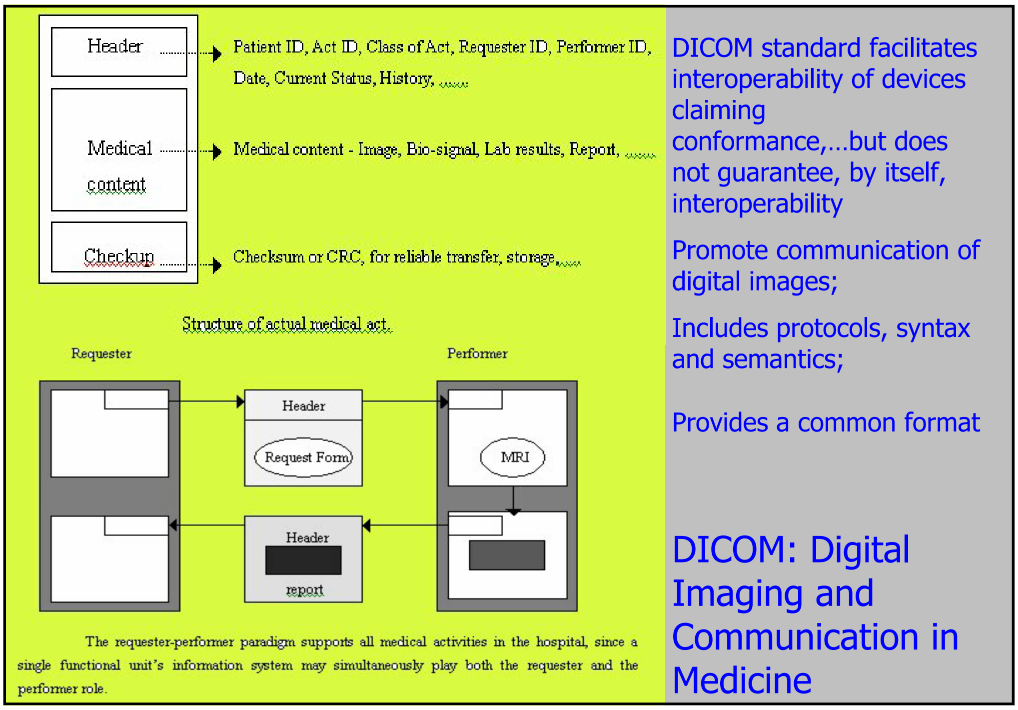


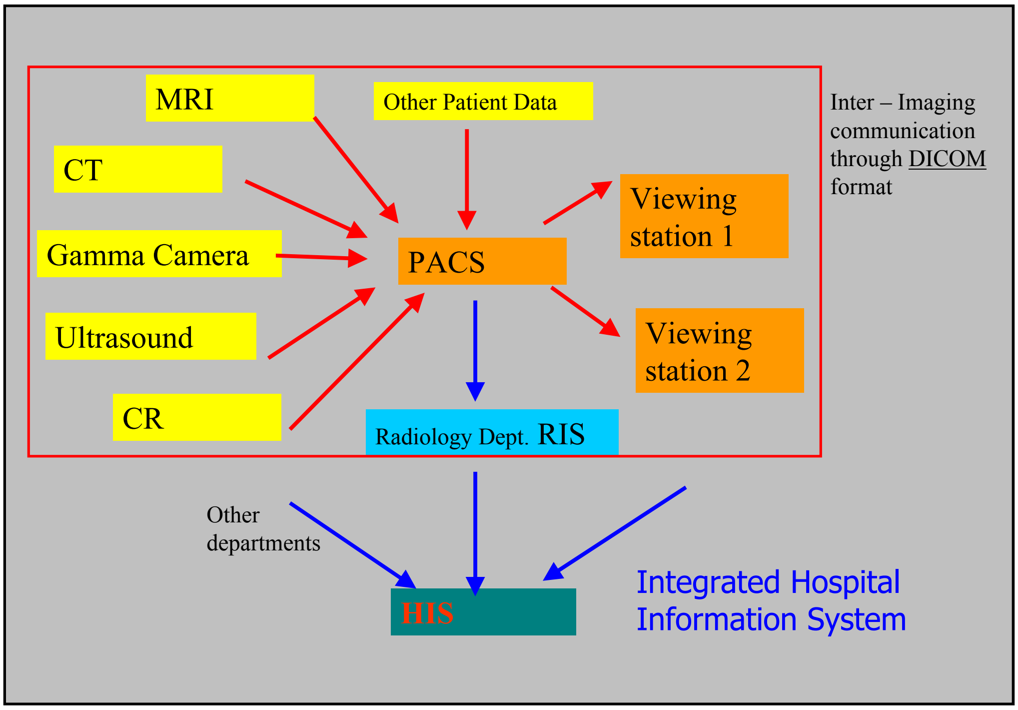
Image move (ATN): min 150 Mbits/sec;

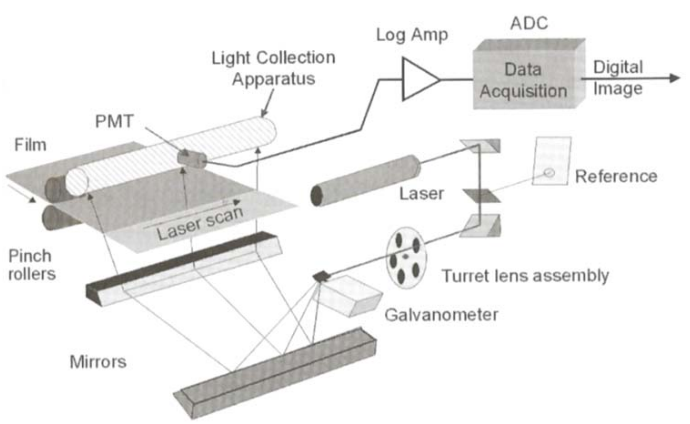
Fibre opt. 600 Mbits/sec

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

Simple PACS architecture


Header
Patient ID, Act ID, Class of Act, Requester ID, Performer ID, Date, Current Status, History, ...
Medical content
Medical content - Image, Bio-signal, Lab results, Report, ...
Checkup
Checksum or CRC, for reliable transfer, storage,......
Structure of actual medical act.
Requester
Performer
Header
Request Form
MRI
Header
report
The requester-performer paradigm supports all medical activities in the hospital, since a single functional unit's information system may simultaneously play both the requester and the performer role.
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


MRI
Other Patient Data
CT
Gamma Camera
PACS
Viewing station 1
Ultrasound
Viewing station 2
CR
Radiology Dept. RIS
Inter – Imaging communication through DICOM format
Other departments
HIS
Integrated Hospital Information System



X-ray film Laser digitizer (with signal proportional to OD)

The laser beam is focussed and moved across the film through galvano-motor (its torque is proportional to the current). The scanning beam passes through reflective tube (with narrow slit) and after many reflections reaches the PMT detector. The log amplifier corrects for OD = $\log_{10}(1/T)$

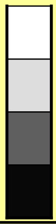
Reminder for : Light Transmittance (T) and Optical Density (OD)

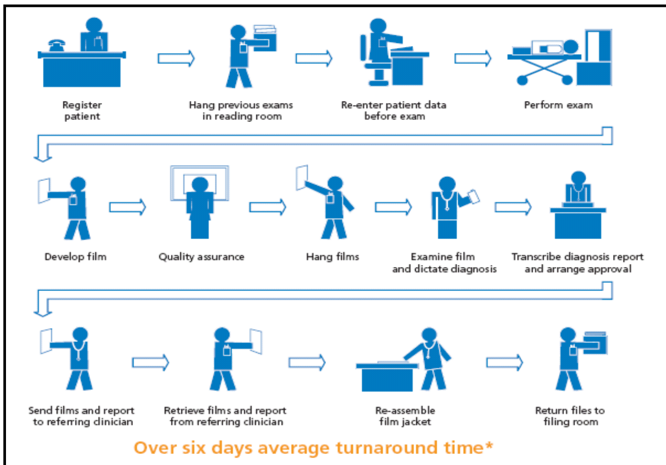
$T = I/I_o$ [max T=1] ; $OD = \log_{10}(1/T)$ [min OD = 0]

OD = 1 >>> T=0.1 medium grey

OD = 2 >>> T=0.01 dark

OD = 3 >>> T= 0.001 very dark [max.OD=4]





PACS disadvantages:
- Cost;
- Training
- Reliance on servers and networking
- PACS failure/update contingency plans

Turnaround time

Before and after PACS

http://www.connectingforhealth.nhs.uk/

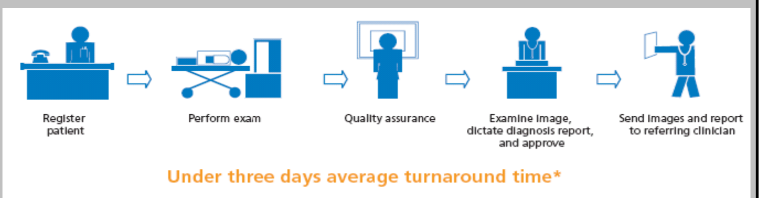
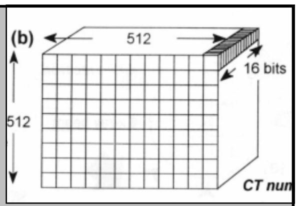


Image quality and matrix size



Matrix size of 2048x2048 pixels can present resolution for field size 400x400 mm of the order of 0.2 mm pixel size (400/2048) = 2.5 lp/mm. Smaller image field (200x200 mm), will have the same resolution (due to the geometry of the image).

For scanners the final resolution will depend not only from the matrix size, but also from the density of projections. If a CT scanner has 512 mm scanning diameter and the matrix is 1024x1024 pixels, the pixel size will be 0.5 mm (512/1024) = 1 lp/mm. If the scanner has collected sufficient number of projections, then part of this raw data can be used for subsequent reconstruction of another smaller image. For example if a ROI with diameter 128 mm is reconstructed, the pixel size of the final image will be 0.125 mm (128/1024), what will present spatial resolution of 4 lp/mm.

Matrix depth (how many bits are included in one pixel) refers to the contrast resolution. Contemporary medical imaging matrices have 16 bits of depth, of which 12 bits are used for displaying the level of grey of the pixel, and the other 4 bits are used for supporting information (text or graphs). The 12 bits present 212 = 4096 levels of grey (or colours), what is more then enough for the human visual system. 4096 levels of grey is also completely sufficient for various densitometric measurements

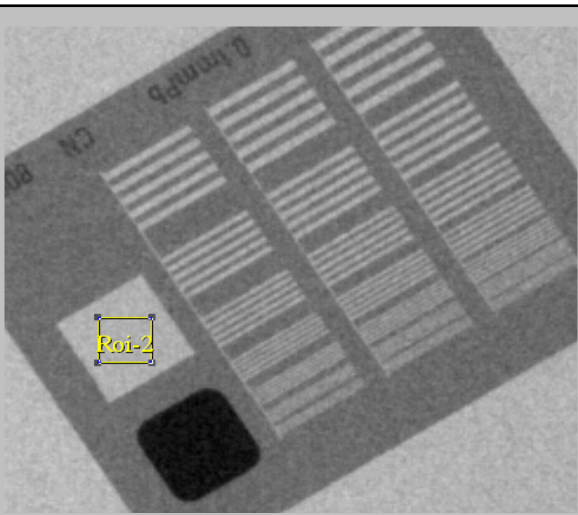
Finally a matrix size can be displayed 2048x2048x16 (4 mega pixel matrix), what will present approx. 67 Mega bits. Presented in Bytes (1 byte = 8 bits), the image file size will be 8 MB.

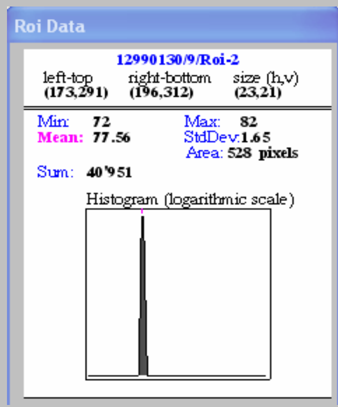
Physical aspects of image quality and Practical examples

- ➔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

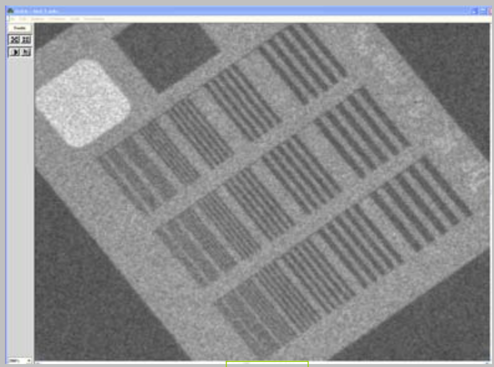




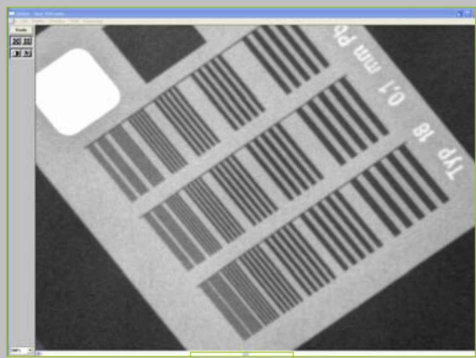
White = more radiation to the detector
Black = less radiation to the detector
Noise (absolute): 1.65
Noise (relative): 1.65 x 100 / 77.56 = 2.1%

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).



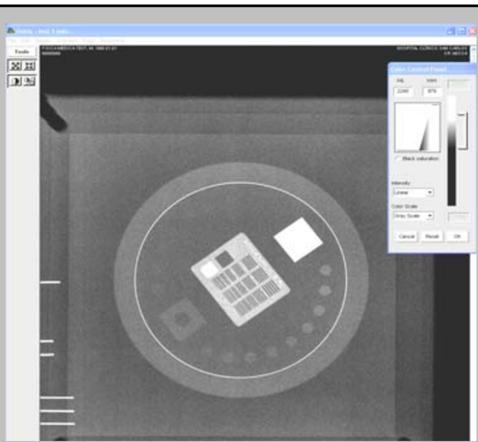
1 mAs



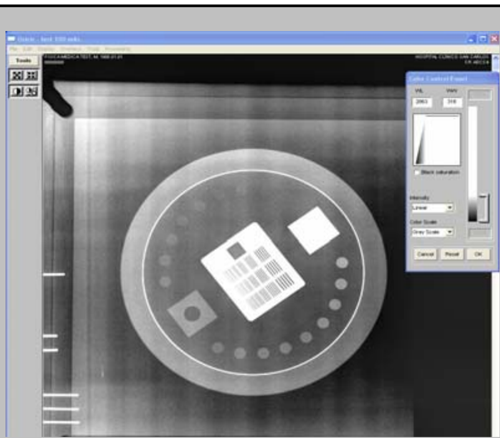
100mAs

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

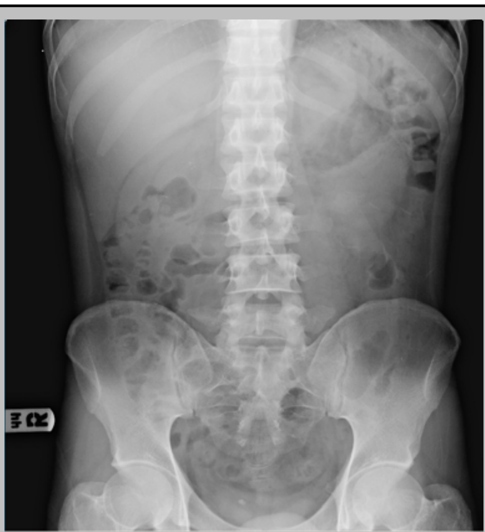
- The standard post-processing parameters offered in some CR workstations includes the noise reduction and the edge enhancement.
- Some examples are shown for the Agfa post-processing called “MUSICA” (Multi Scale Image Contrast Enhancement). This is the basic principle of MUSICA:
  - contrast enhancement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

Standard image

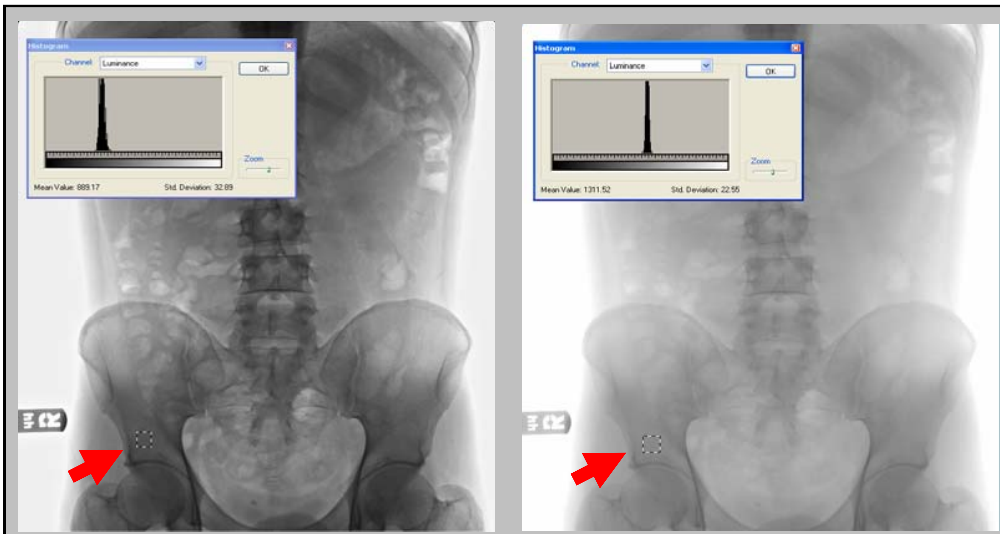
Noise reduction

Edge enhancement

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



Same image, two different postprocessings?

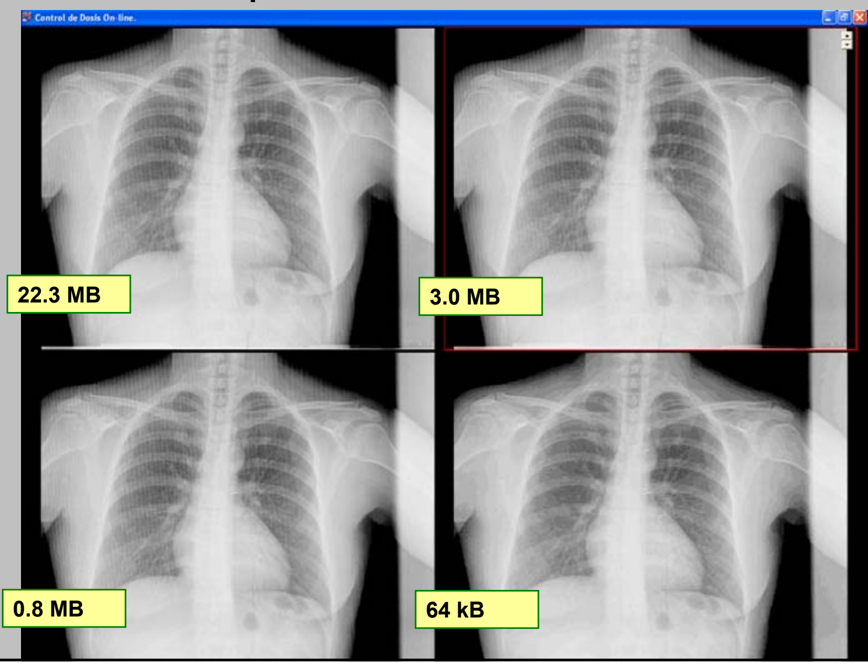


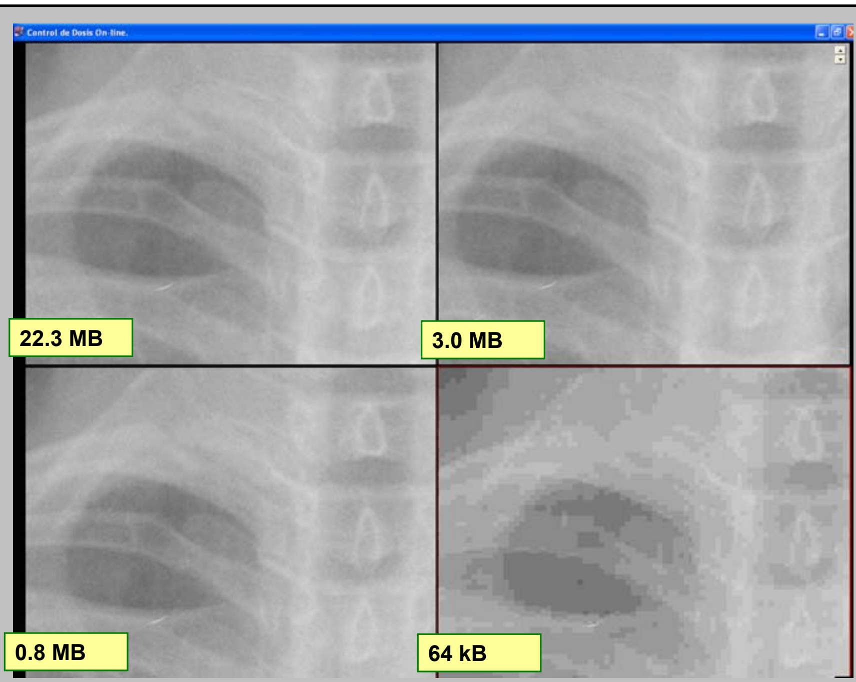
Same image, two different postprocessings?

Noise = 33/889 = 3.7%

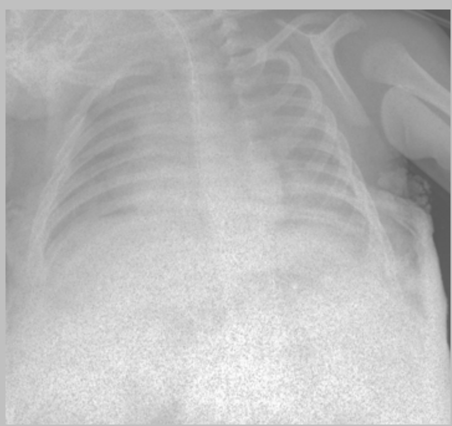
Noise = 23/1312 = 1.8%

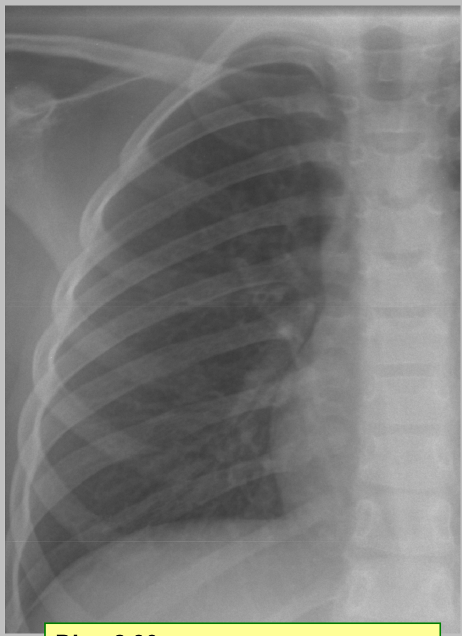
Different compression levels



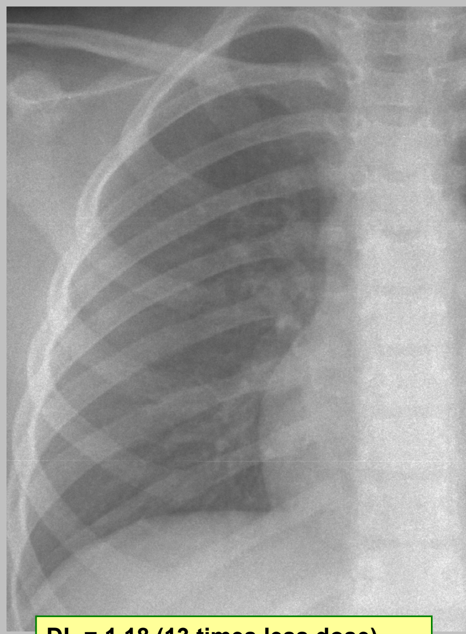


Patient Dose
More dose better image quality





DL = 2.30

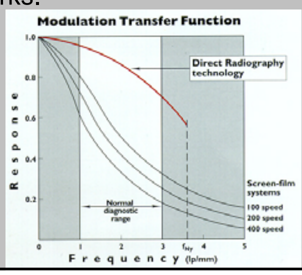


DL = 1.18 (13 times less dose)

Digital radiography and digital fluoroscopy. Differences with conventional

Advantages

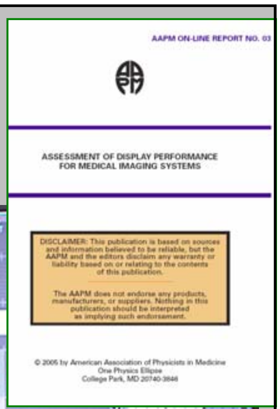
- More information can be obtained from the image (change of window and level, magnification, etc).
- Wide dynamic range (more tolerance to different dose values).
- Easy archive and transmission by networks.



Disadvantages

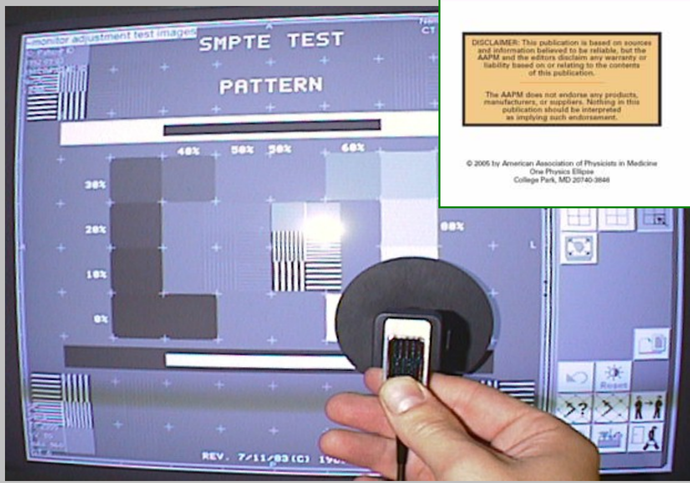
- Over exposures could not be noticed.
- Very easy to delete the files of the bad quality images.
- A tendency to obtain more images than necessary could occur.
- Audit of relevant radiation protection parameters can sometimes be difficult.

Monitors: photometer measurement

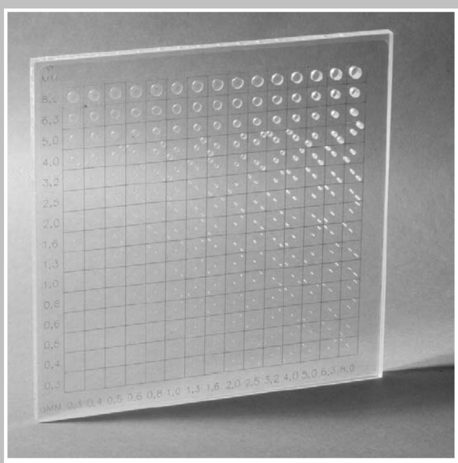


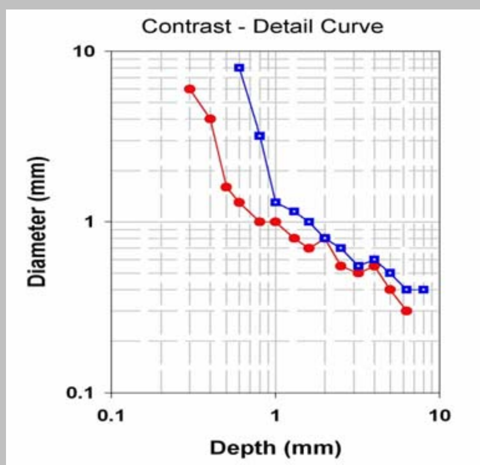
Poor conditions of the visualization monitor (e.g. lack of enough brightness or contrast, poor spatial resolution, etc) can require repetitions of exposures.

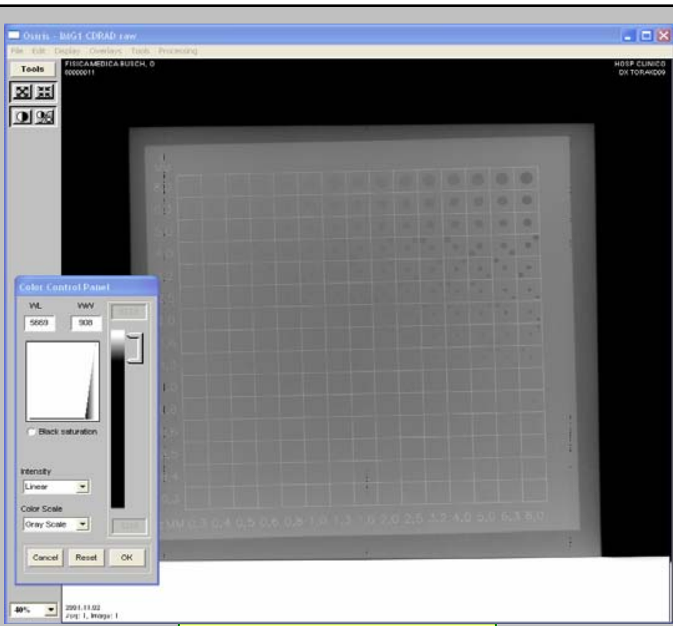
Such a visualization monitor produces sub-standard image quality.



CDRAD phantom





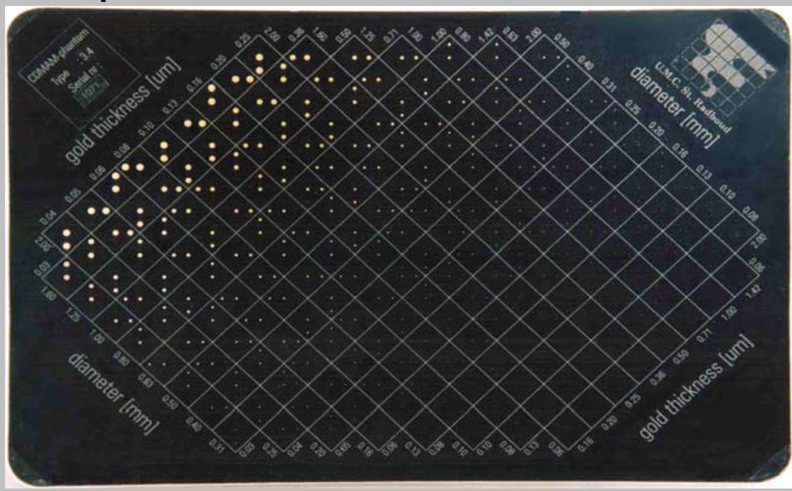


IMG1 CDRAD raw;

CDRAD phantom

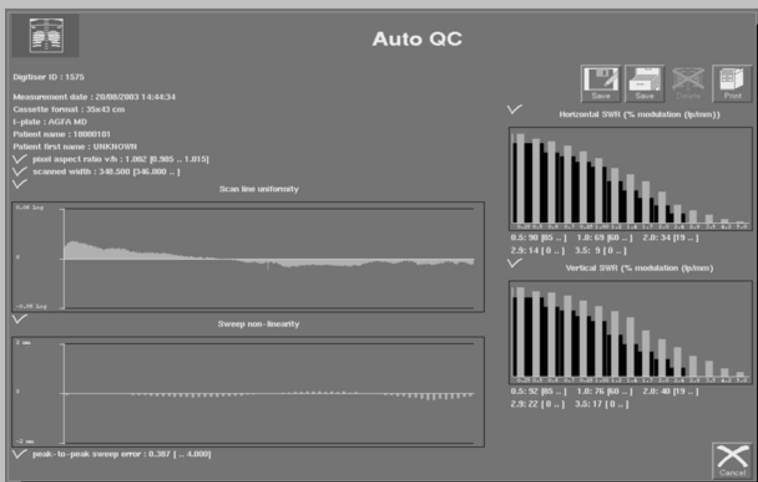
- The image shows 225 squares, 15 rows and 15 columns. In each square either one or two spots are present, being the images of the holes.
- The first three rows show only one spot, while the other rows have two identical spots, one in the middle and one in a randomly chosen corner.

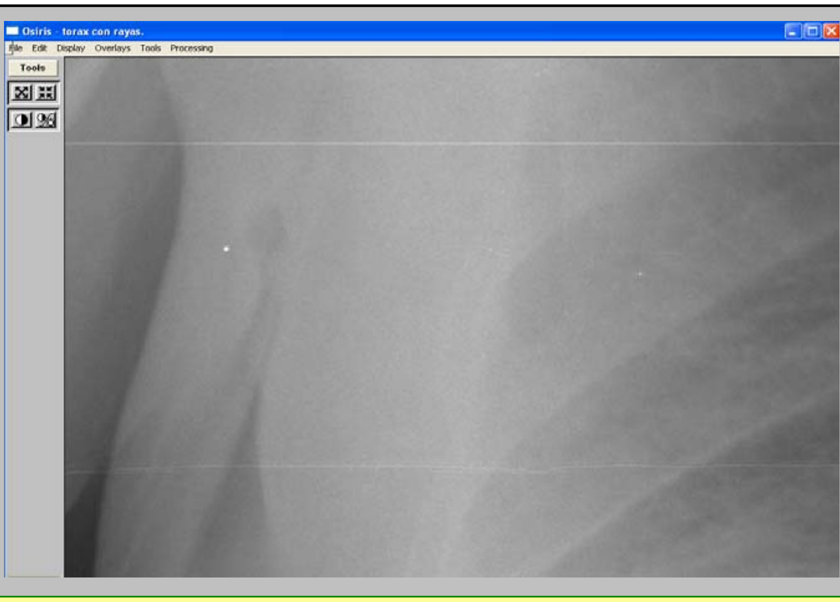
CDMAM phantom



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

Acknowledgments

- Figures from Agfa, Siemens, Philips, GE, Fuji and Toshiba systems have been used.
- Materials from IAEA Training Material on Radiation Protection in Digital Radiology have been used
- Images from Prof. Perry Sprawls, Dr. Ramon Sanchez-Jacob, Dr. Eliseo Vano-Galvan, Dr Anchali Krisanachinda, Petcharleeya Suwanpradit, and Ana Pascoal have been used
- Images from EMERALD materials and Dr S Tabakov have been used