Bushberg - Chapter 6: Screen-Film Radiography
Chapter 11: Digital Radiography

RSNA & AAPM Physics Curriculum: Module 10
X-Ray Projection Imaging Concepts and Detectors

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a copy of this lecture may be found at:
http://courses.washington.edu/radxphys/PhysicsCourse.html

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❖ **Fundamental Knowledge:**

- ✓ Describe the fundamental characteristics of all projection imaging systems that determine the capabilities and limitations in producing an x-ray image
- ✓ Review the detector types used to acquire an x-ray imaging. Describe how radiation is detected by each detector type and the different attributes of each detector for recording information

❖ **Clinical Application:**

- ✓ Demonstrate how variations in each of the fundamental characteristics of a projection imaging system affect the detected information in producing an image.
- ✓ Give examples of how each detector type performs in imaging a specific body part or view, and describe how the attributes of each detector type influence the resulting image.

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❖ **Clinical Problem-Solving:**

- ✓ What is the difference in exposure class between CR and DR systems? How does this difference affect patient dose?

- ✓ Describe some of the common artifacts seen in a portable chest x-ray image, and explain how these can be minimized.

- ✓ Describe how distance to the patient and detector affect patient dose.

- ✓ Describe how the transition from film to a digital detector systems eliminates some artifacts and creates the possibility of others.

- ✓ What are the properties of a detector system that determines its suitability for pediatric procedures?

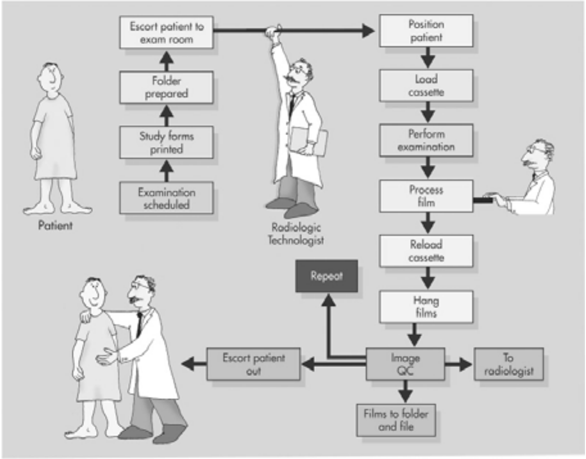
Bushberg - Chapter 11: Digital Radiography

Lecture 2

Q: What are some limiting factors of the almost 100 year old analog film-screen technology?

- limited dynamic range (only about two orders of magnitude rather than four or five) – may require more exposure repeats
- film can only be viewed in one location at a time (unless a copy film produced)
- can’t visualize the image immediately – patient has to wait while film processed and reviewed – radiographic room throughput and utilization suffer
- film processor problems and film processor QA monitoring time consuming

Workload Steps in Screen-Film Environment



c.f. RSNA/AAPM Online Physics Modules: Radiographic Image Receptors

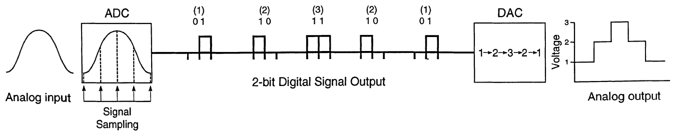
Q: What image information might we lose with the use of digital detectors?

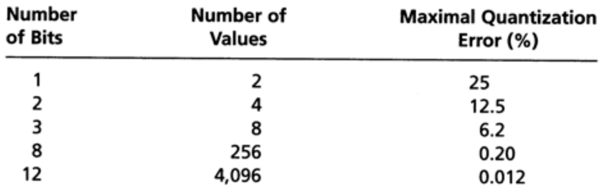
- loss of spatial resolution (limiting resolution about 2.5 lp/mm for 17”x17” general-purpose digital detector, screen-film about 8-10 lp/mm)
- quantization error (depends on how many bits are used to represent each pixel, but not as great an issue as the loss of spatial resolution)

Review: Digital Representation of Data

- Storage of Positive Integers
  - In general, n bits have 2^n possible permutations and can represent integers from 0 to 2^n-1 (the range usually denoted with square brackets):
  - n bits represents 2^n values with range $[0, 2^n-1]$
  - 8 bits represents $2^8 = 256$ values with range [0, 255]
  - 10 bits represents $2^{10} = 1024$ values with range [0, 1023]
  - 12 bits represents $2^{12} = 4096$ values with range [0, 4095]
  - 14 bits represents $2^{14} = 16384$ values with range [0, 16383]
  - 16 bits represents $2^{16} = 65,536$ values with range [0, 65535]

Review: Conversion of Analog Data to Digital Form



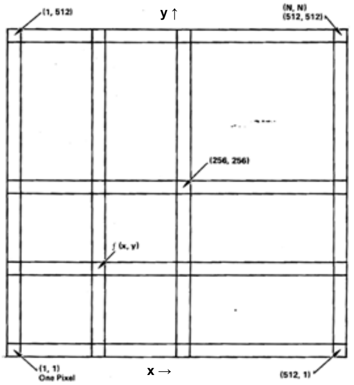


Number of Bits	Number of Values	Maximal Quantization Error (%)
1	2	25
2	4	12.5
3	8	6.2
8	256	0.20
12	4,096	0.012

c. f. Bushberg, et al., The Essential Physics of Medical Imaging, 2nd ed., p. 69.

Review: Digital Storage of Images

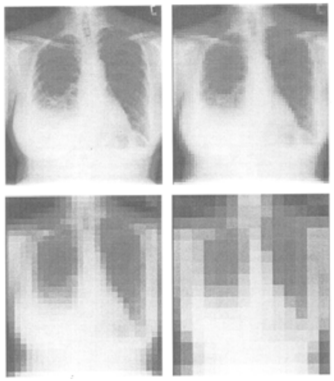
- Usually stored as a 2D array (matrix) of data, I(x,y): I(1,1), I(2,1), … I(n,m-1), I(n,m)
- Each minute region of the image is called a *pixel* (picture element) represented by one value (e.g., digital value, gray level or Hounsfield unit)
- Typical matrices:
  - CT: 512x512x12 bits/pixel
  - CR: 1760x2140x10 bits/pixel
  - DR: 2048x2560x16 bits/pixel



c.f. Huang, HK. Elements of Digital Radiology, p. 8.

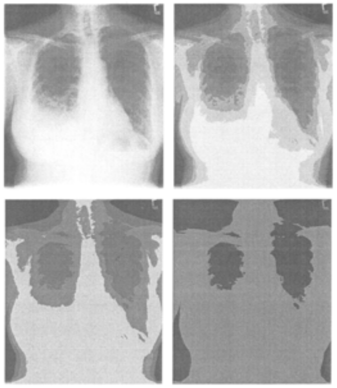
Review: Effect of Resolution and Bits per Pixel

1024^2, 64^2, 32^2, 16^2 matrices



c.f.: Bushberg, et al., The Essential Physics of Medical Imaging, 2nd ed., p. 82.

8, 3, 2, 1 bits/pixel



c.f.: Bushberg, et al., The Essential Physics of Medical Imaging, 2nd ed., p. 84.

Overview of Digital Systems

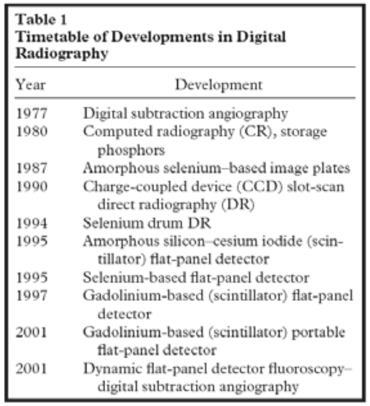


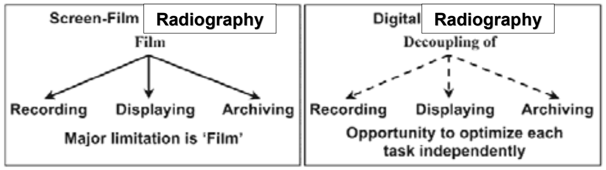
Table 1
Timetable of Developments in Digital Radiography

Year	Development
1977	Digital subtraction angiography
1980	Computed radiography (CR), storage phosphors
1987	Amorphous selenium–based image plates
1990	Charge-coupled device (CCD) slot-scan direct radiography (DR)
1994	Selenium drum DR
1995	Amorphous silicon–cesium iodide (scintillator) flat-panel detector
1995	Selenium-based flat-panel detector
1997	Gadolinium-based (scintillator) flat-panel detector
2001	Gadolinium-based (scintillator) portable flat-panel detector
2001	Dynamic flat-panel detector fluoroscopy–digital subtraction angiography

c.f. **RadioGraphics 2007;** 27:675–686 c

Advantages of going Digital

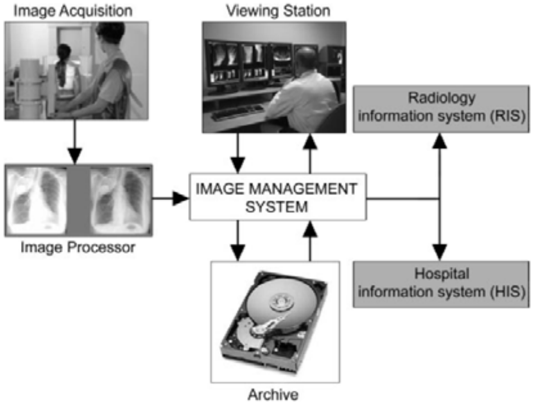
- Wider dynamic range
- Implementation of PACS, softcopy read and thus teleradiology
- Higher patient throughput
- Increased dose efficiency



Physical Principles of Digital Radiography

- Energy from x-rays is absorbed by the digital detector and transformed into electrical charges
- Electrical charges are recorded, digitized and quantified into a gray scale that represents the amount of x-ray energy deposited on the detector
- Postprocessing software is needed for organizing the raw data into clinically meaningful image
- Images are sent to archive and linked to the patient demographic information
- Sent to workstation for review

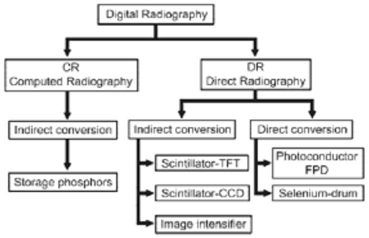
Physical Principles of Digital Radiography



c.f. RSNA/AAPM Online Physics Modules: Radiographic Image Receptors

Various Types of Digital Detectors

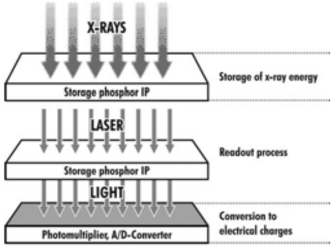
- Computed radiography (CR) systems use storage-phosphor image plates with a separate image readout process
- Direct digital radiography (DR) is a way of converting x-rays into electrical charges by means of a direct readout process
- Indirect DR is a way of converting x-rays into electrical charges by means of a scintillator



c.f. **RadioGraphics 2007;** 27:675–686

CR

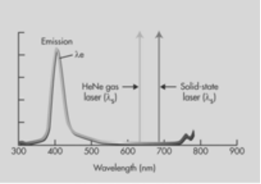
- In CR, the response to x-ray interaction is seen as trapped electrons in a higher-energy metastable state
- Some electrons will return to the ground state promptly but will not contribute to the latent image
- Barium fluorohalide (BaFBr:Eu or BaFI:Eu) is an example of a photostimulable phosphor (PSP)
- The remaining metastable electrons return to the ground state over time causing the electronic latent image to fade
- Consequently, the IP must be processed soon after exposure
- CR signal loss is objectionable after approximately 8 hours



c.f. **RadioGraphics 2007;** 27:675–686

CR

- Processing the IP begins with stimulating the PSP with a finely focused beam of infrared light from a laser
- The diameter of the laser beam affects the spatial resolution of the CR system
- Photomultiplier tubes (PMT) and photodiodes (PD) are the light detectors of choice for CR causing the latent electronic image to be made visible
- If any residual signal remained on the IP following reading, ghosting could appear on subsequent use of the IP
- Any residual latent image is removed by flooding the phosphor with very intense white light from a bank of specially designed fluorescent lamps

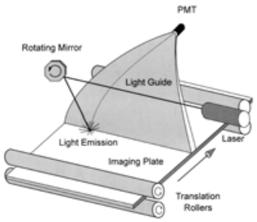


The red laser light is used to stimulate the PSP. When the trapped electron energy is released, a broad spectrum of blue-green light is emitted.

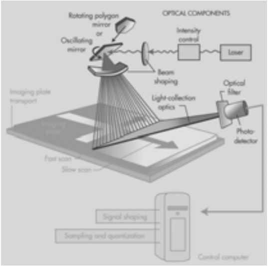
c.f. RSNA/AAPM Online Physics Modules: Radiographic Image Receptors

CR

c.f. Bushberg, et al. The Essential Physics of Medical Imaging, 2nd ed., p. 294.

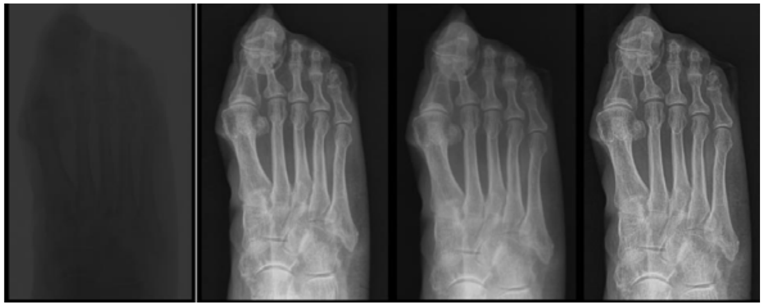


- The output of the photodetector is processed for amplitude, scale, and compression
- Then the analog signal is digitized, with attention paid to proper **sampling** (time between samples) and **quantization** (the value of each sample)
- Sampling and quantization are the process of analog-to-digital conversion (ADC)



c.f. RSNA/AAPM Online Physics Modules: Radiographic Image Receptors

Image Processing



L to R: Raw Image, Contrast enhanced, Contrast reduction, Edge Enhancement

c.f. RSNA/AAPM Online Physics Modules: Radiographic Image Receptors

Q: What features of CR have made it useful as a transition between analog film/screen and digital radiography?

- uses the same sized cassettes as in film/screen radiography, so little impact to the then current film/screen radiography suite equipment (i.e., Bucky tray and phototimers)
- initial implementations of CR used film printers to make hardcopy for review (keep old lightboxes and film alternator equipment – didn't have to spend $ on PACS)

Radiographic Suite

CR Cassette

CR Reader

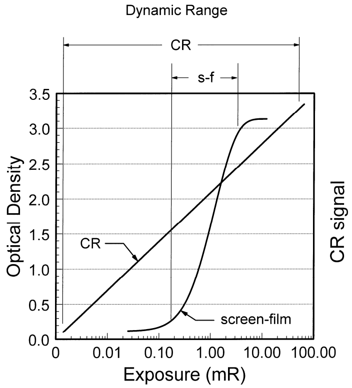
Film Alternator

c.f. http://homepages.cae.wisc.edu/~bme402/time_stamp/images/CR_casette_2.jpg

c.f. http://www.aomedical.se/alternatorAccess.htm

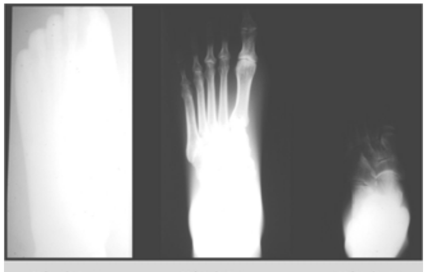
CR

- IP dynamic range = 10^4, about 100x that of S-F (10^2)
- Very wide latitude → flat contrast
- Image processing required
- CR's wide latitude and image processing capabilities produce reasonable exposure for either under or overexposed exams
- Dose creep possible
- Helps in portable radiography: where the tight exposure limits of S-F are hard to achieve



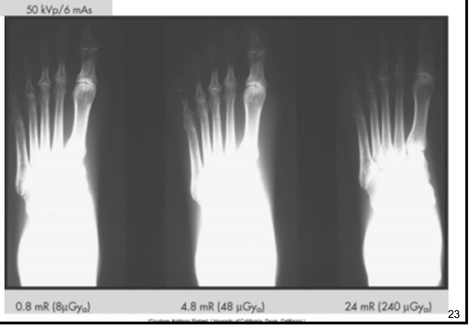
c.f. Bushberg, et al. The Essential Physics of Medical Imaging, 2^{nd} ed., p. 296.

SF vs. CR

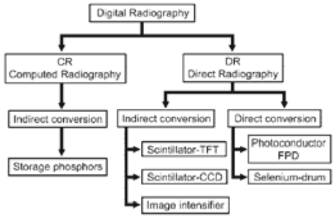


c.f. RSNA/AAPM Online Physics Modules: Radiographic Image Receptors

Digital radiographs of a foot phantom using the same radiographic techniques as above show the maintenance of contrast over a wide range of patient radiation dose.



Various Types of Digital Detectors



Indirect DR – create visible light photons from x-rays with scintillator then produce electrons with photodiodes typically lower spatial resolution than direct DR and lower dose efficiency than direct DR due to limiting the phosphor thickness so as not to adversely impact spatial resolution

Direct DR - directly create electrons from absorbed x-rays, typically higher spatial resolution than indirect DR and higher dose efficiency than indirect DR due to electric field lines constraining electron lateral drift (can make photoconductor thick - > high x-ray absorption)

c.f. **RadioGraphics 2007;** 27:675–686

Indirect DR

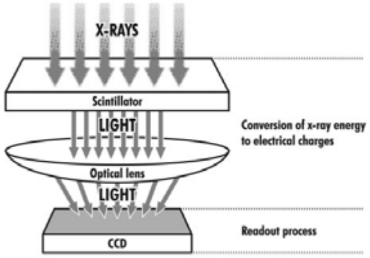


Figure 5. CCD-based indirect conversion DR system. **(a)** Drawing illustrates a lens-coupled CCD-based system. The incident x-ray energy is converted into light by a scintillator. The emitted light has to be bundled by an optical lens to fit the size of the CCD chip, which subsequently converts the light energy into electrical charges.

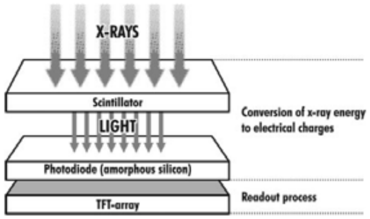
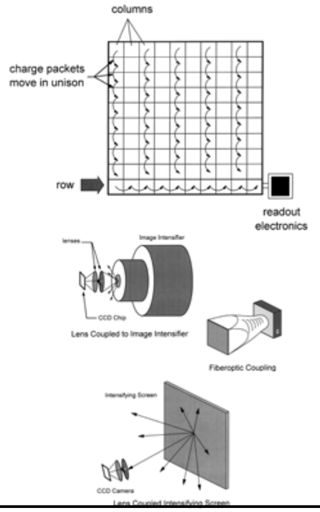


Figure 6. Drawing illustrates an amorphous silicon–based indirect conversion DR system. X-ray energy is converted into visible light in a scintillator layer. The emitted light is then converted into electrical charges by an array of silicon-based photodiodes and read out by a TFT array.

c.f. **RadioGraphics 2007;** 27:675–686

Charged-Coupled Devices (CCD)

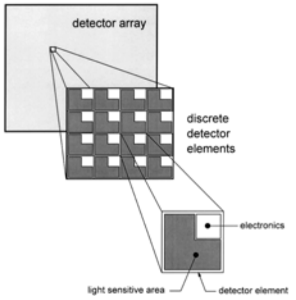
- CCDs form images from visible light
- Used in videocams & digital cameras
- CCD chip is an integrated circuit with discrete pixel electronics
- CCD chip made of crystalline silicon which is photosensitive
- As light falls on the each pixel, electrons are liberated and build up in the pixel
- The electronic charge electronically readout via 'bucket brigade'
- Light focused using lenses or fiber-optics – fluoroscopy, cine, digital mammo
- 1K and 2K CCDs used



c.f. Bushberg, et al. The Essential Physics of Medical Imaging, 2nd ed., pp. 298-299.

Indirect Flat Panel Detectors
Thin-film transistors (TFT)

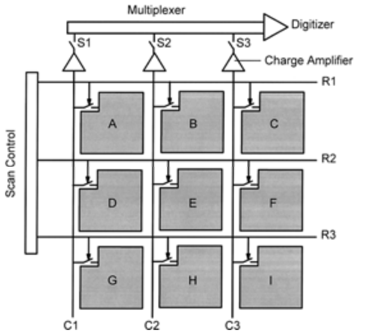
- Similar to that used in laptop screens: thin-film transistors (TFT)
- Use an intensifying screen - e.g., Gd_2O_2S or CsI - to generate light photons from an x-ray exposure
- Light photons absorbed by TFT array
- Each element of the array (pixel) consists of transistor (readout) electronics and a photodetector area



c.f. Bushberg, et al. The Essential Physics of Medical Imaging, 2nd ed., pp. 234 & 301.

Thin-Film Transistors (TFT)

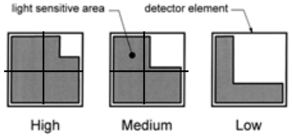
- During exposure, charge is build up in each detector element and stored
- After the exposure is complete charge in each detector element is read out using electronics shown to the right (gate, source, drain)



c.f. Bushberg, et al. The Essential Physics of Medical Imaging, 2nd ed., p. 301.

Resolution and Fill Factor

- Dimension of detector element largely determines spatial resolution
- 200μm and 100μm pixel size typical
- For dimension of 'a' mm - Nyquist frequency: $F_N = 1/(2a)$
- If a = 100μm → F_N = 5 cycle/mm or lp/mm
- Fill factor = (light sensitive area)/(detector element area)
- Trade-off between spatial resolution and contrast resolution



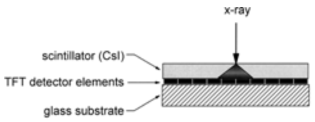
$$\text{Fill Factor} = \frac{\text{light sensitive area}}{\text{area of detector element}}$$

c.f. Bushberg, et al. The Essential Physics of Medical Imaging, 2nd ed., p. 303.

Direct Flat Panel Detectors

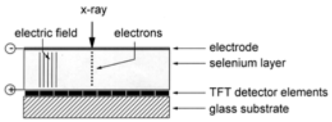
- Use a layer of photoconductive material (e.g., α-Se) atop a TFT array
- Electrons are released in the detector layer from x-ray interactions used to form the image directly
- High degree of e^- directionality through application of E field
- Electrons are collected on the detector elements and then readout
- Photoconductive material can be made thick w/o significant degradation of spatial resolution
- Photoconductive materials
  - Selenium commonly used

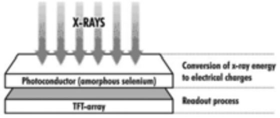
c.f. Bushberg, et al. The Essential Physics of Medical Imaging, 2nd ed., p. 304.



Indirect Flat Panel Detector (for comparison)

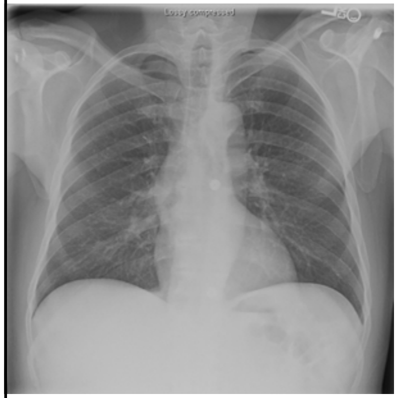
Direct Flat Panel Detector



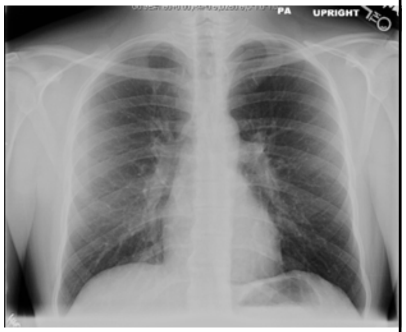


c.f. **RadioGraphics 2007;** 27:675–686

CR Versus DR



DR



CR

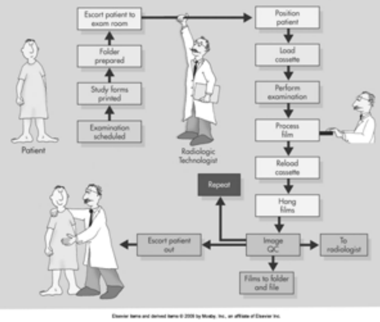
Patient Dose Considerations

- With FS systems, over- or under- exposure provides immediate feedback and requires repeating the exposure
- With digital systems, over- or under- exposure will still produce a decent image due to wide dynamic range
- High exposures may go unnoticed – "dose creep"
- QC essential
- FS – 400 speed for general radiography
- CR – half as efficient, more like a 200 speed
- DR – 2 times more efficient that FS system with 1/2 to 1/3rd the dose of FS

Digital versus Analog Processes & Implementation

- Although some of the previous image reception systems were labeled ‘digital’, the initial stage of all those devices produce an analog signal that is later digitized
- CR: x-rays→metastable state → VL→PMT→ADC
- CCD, direct & indirect digital detectors: stored e^- → ADC
- Benefits of CR
  - Same exam process and equipment as screen-film radiography
  - Many exam rooms serviced by one reader, lower initial cost
- Benefits of DR
  - Throughput ↑: radiographs available immediately for QC & read
  - Reduced radiation dose: 2-3x compared with CR
  - Greater spatial resolution possible

Going from SF to Digital



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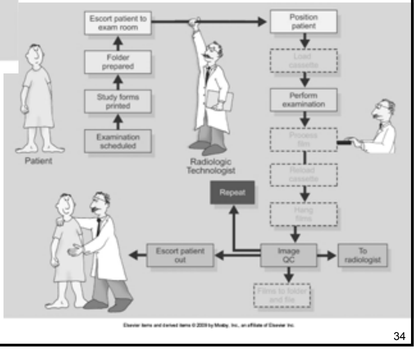

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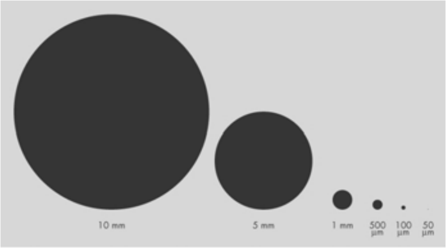
Spatial Resolution



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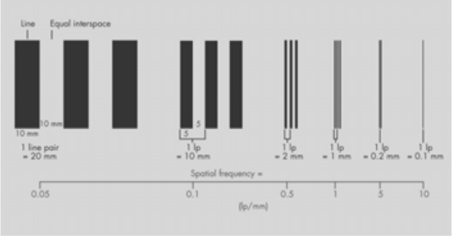

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c.f. RSNA/AAPM Online Physics Modules: Radiographic Image Receptors

Approximate Spatial Resolution for Various Medical Imaging Systems

Gamma Camera	0.1 lp/mm
Magnetic Resonance Imaging	1.5 lp/mm
Computed Tomography	1.5 lp/mm
Diagnostic Ultrasound	2 lp/mm
Fluoroscopy	3 lp/mm
Digital Radiography	5-7 lp/mm
Computed Radiography	5 lp/mm
Screen-Film Radiography	8-10 lp/mm
Mammography	15 lp/mm

Contrast Resolution

- The principal descriptor for contrast resolution is grayscale, also called dynamic range
- The dynamic range of a digital radiographic system is identified by the bit capacity of each pixel as shown in the following table

- Dynamic Range of Digital Medical Imaging Systems

Imaging System	Bit Depth	Shades of Gray
Screen-Film		30 (0-3 OD)
Diagnostic Ultrasound	2^{8}	256
Nuclear Medicine	2^{10}	1,024
CT	2^{12}	4,096
MRI	2^{12}	4,096
Digital Radiography	2^{14}	16,384
Digital Mammography	2^{16}	65,536

Review Question

- Photostimulable phosphors [systems] do NOT include:

- A. Analog-to-digital converters
- B. Barium fluorohalide
- C. Light detectors (blue)
- D. Red light lasers
- E. Video cameras

Review Question

- Which of the following x-ray detector materials emits visible light:

- A. Xenon
  - X-ray→e^-→ADC
- B. Mercuric iodide
  - X-ray→e^-→TFT→ADC
- C. Lead iodide
  - X-ray→e^-→TFT→ADC
- D. Selenium
  - X-ray→e^-→TFT→ADC
- E. Cesium iodide
  - X-ray→e^-→VL→e^-→TFT→ADC

Review Question

- Concerning computed radiography (CR), which of the following is true?
- A. Numerous, small solid-state detectors are used to capture the x-ray exposure patterns.
- B. It has better spatial resolution than film.
- C. It is ideal for portable x-ray examinations, when phototiming cannot be used.
- D. It is associated with high reject/repeat rates.
- E. The image capture, storage, and display are performed by the receiver.

Review Question

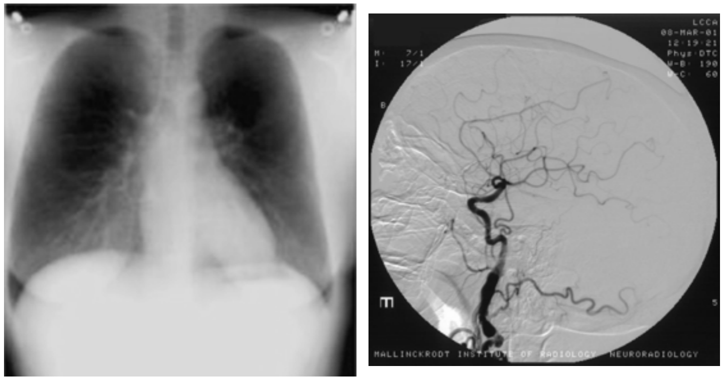
- Photoconductors convert x-ray energy directly into:
- A. Light
- B. Current
- C. Heat
- D. Charge
- E. RF energy

Image Processing

Why is Image Processing Important or Necessary?

- Want to do some neat things:
  - Remove bone, see only contrast, mimic CT
  - Increase sharpness, enhance contrast
- Digital detector elements don’t operate identically
- Some pixels go bad or have a large offset
- The wide dynamic range of digital detectors → flat image
- Must manipulate image digitally to view → window/level
- May wish to reduce image noise
- May wish to enhance the spatial frequency → MTF
- Try to automatically set gray-scale mapping of image

Why is Image Processing Important or Necessary?



Digital Image Correction (1)

c.f. Bushberg, et al. The Essential Physics
of Medical Imaging, 2nd ed., p. 310

Digital Image Correction (2)

- Interpolation to fill in dead pixel and row/column defects
- Subtracting out average dark noise image $D_{avg(t)}(x,y)$

1	9	5
4	6	2
7	3	8

D(t=0)(x,y): dark noise

4	5	3
5	6	4
7	4	5

$D_{avg}(t)(x,y)$: dark noise avg

- Differences in detector element digital values for flat field
  - Gain image: $G(x,y) = G'(x,y) - D_{avg(t)}(x,y)$; $G_{avg} = (1/N) \cdot \sum\sum G(x,y)$

126	112	116
108	125	117
127	120	111

G'(x,y): raw gain image

122	107	113
103	119	113
120	116	106

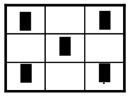
G(x,y): corrected gain image

113	113	113
113	113	113
113	113	113

G_{avg}: average gain image

Digital Image Correction (3)

- Make corrections for each detector element (map)
  - $I(x,y) = G_{avg} \cdot [I_{raw}(x,y) - D_{avg(t)}(x,y)] / G(x,y)$



Target phantom: X

1285	226	1209
209	1296	228
1268	235	1059

$I_{raw}(x,y)$: raw image of X

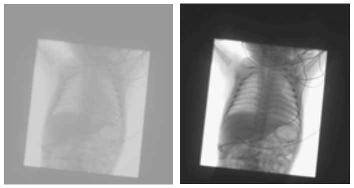
1201	239	1206
228	1202	226
1206	227	1209

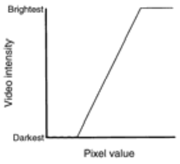
I(x,y): corrected image of X

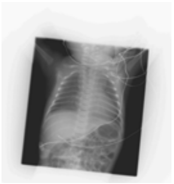
- Done for DR and in a similar manner for CT (later) – and in your digital camera!
- Not performed for CR on a pixel by pixel basis, although there are corrections on a column basis for differences in light conduction efficiency in the light guide to the PMT

Global Processing

- Most common global image processing: window/level
- Global processing algorithm
  - $I'(x,y) = c \cdot [I(x,y) - a]$: essentially $y = mx + b$
  - Level (brightness) set by a
  - Window (contrast) set by c
  - $I' = [2^N/ww] \cdot [I-\{wl-(ww/2)\}]$, where ww = window width and wl = window level
  - Need threshold limits when max/min $[2^N-1, 0]$ digital values encountered
  - If $I'(x,y) > T_{max} \rightarrow I'(x,y) = T_{max}$
  - If $I'(x,y) < T_{min} \rightarrow I'(x,y) = T_{min}$

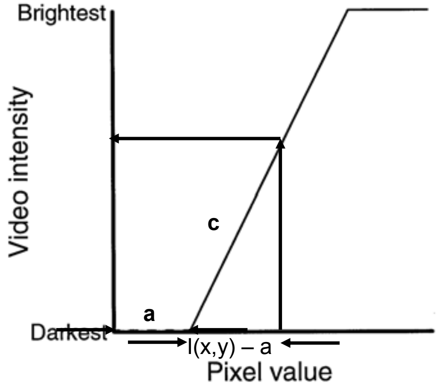






c.f. Bushberg, et al. The Essential Physics
of Medical Imaging, 2nd ed., pp. 92 and 331

I’(x,y) = c · [I(x,y) – a]



Window and Level Controls

- Modification of the translation table causes changes in the displayed image brightness and contrast and is usually done through window (contrast) and level (brightness) controls (e.g., under mouse control)
  - In the example (below), the window is kept constant as the level is increased, causing the image to become darker and darker
  - The narrower the window, the greater the displayed image contrast, but more of the image saturated in either toe or shoulder regions of LUT

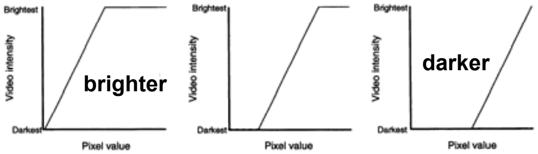
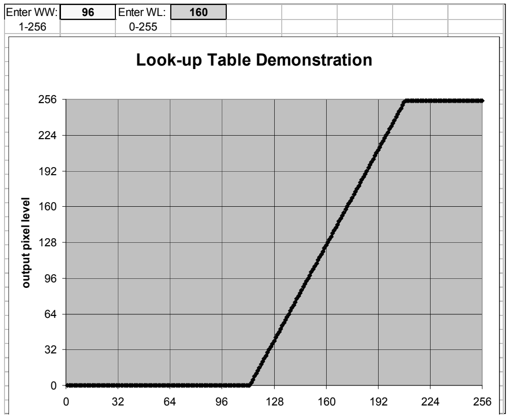


FIGURE 4-16. Windowing an image with pixel values from 0 to 200 to enhance contrast in pixels from 0 to 100 (*left*), 50 to 150 (*center*), and 100 to 200 (*right*).

c.f.: Bushberg, et al., The Essential Physics
of Medical Imaging, 2nd Ed., p. 92

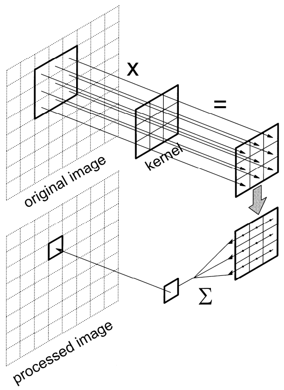
Window/Level Demo



Enter WW: 96 1-256
Enter WL: 160 0-255

Image Processing Based on Convolution

- Convolution: Ch. 10 - Image Quality and Ch. 13 - CT
- Defined mathematically as passing a N-dimensional convolution kernel over an N-dimensional numeric array (e.g., 2D image or CT transmission profile)
- At each location (x, y, z, t, ...) in the number array multiply the convolution kernel values by the associated values in the numeric array and sum
- Place the sum into a new numeric array at the same location



c.f. Bushberg, et al. The Essential Physics
of Medical Imaging, 2nd ed., p. 312

Image Processing Based on Convolution

- Delta function kernel

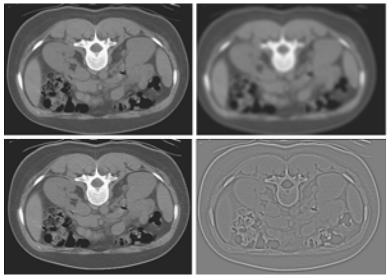
0	0	0
0	1	0
0	0	0

- Blurring kernel (normalization) also known as low-pass filter

1/9	1/9	1/9
1/9	1/9	1/9
1/9	1/9	1/9

- Edge sharpening kernel

-1	-1	-1
-1	9	-1
-1	-1	-1

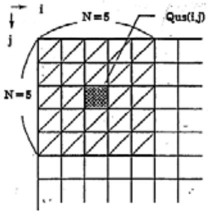


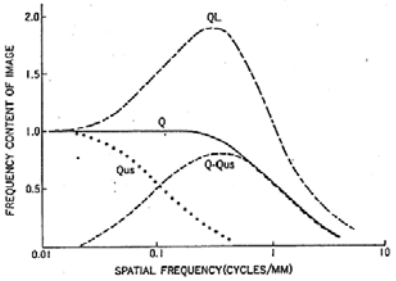
c.f. Bushberg, et al. The Essential Physics of Medical Imaging, 2nd ed., p. 313.

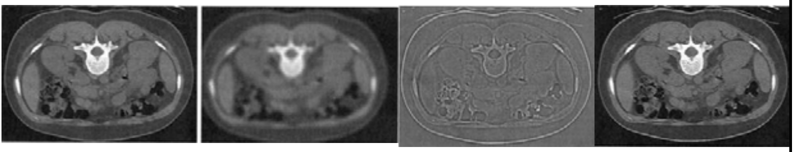
Image Processing Based on Convolution

- Convolution kernels can be much larger than 3 x 3, but usually N x M with N and M odd
- Can also perform edge sharpening by subtracting blurred image from original → high-frequency detail (harmonization)
- The edge sharpened image can then be added back to the original image to make up for some blurring in the original image: CR unsharpmasking - freq. processing
- The effects of convolution cannot in general be undone by a 'de-convolution' process due to the presence of noise, but a deconvolution kernel can be applied to produce an approximation: ^{19}F MRI

Unsharpmasked Spatial Frequency Processing







c.f. Bushberg, et al. The Essential Physics of Medical Imaging, 2nd ed., p. 313.

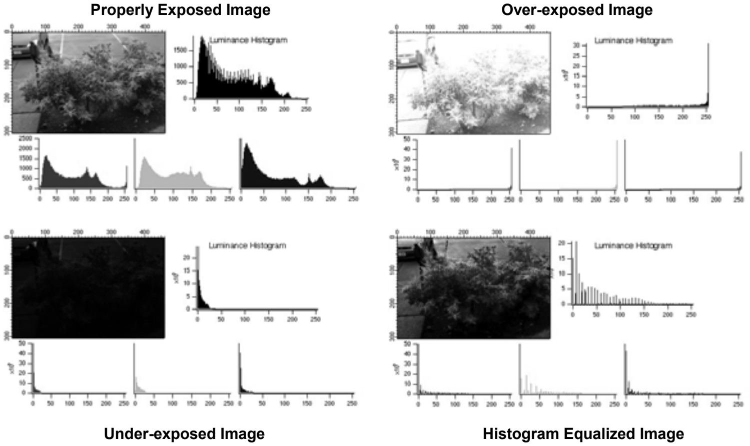
Median and Sigma Filtering

- Convolution of an image with a kernel where all the values are the same, e.g. (1/NxM), essentially performs an average over the kernel footprint
- Smoothing or noise reduction
- This can make the resulting output value susceptible to outliers (high or low)
- Median filter: rank order values in kernel footprint and take the median (middle) value
- Sigma filter: set sigma (σ) value (e.g., 1) and throw out all values in kernel footprint > $\mu + \sigma$ or < $\mu - \sigma$ and then take the average and place in output image

Multiresolution/Multiscale Processing and Adaptive Histogram Equalization (AHE)

- Some CR systems (Agfa/Fuji) make use of multiresolution image processing (AKA unsharpmasking) to enhance spatial resolution
- Wavelet or pyramidal processing on multiple frequency scales
- Histogram equalization re-distributes image digital values to uniformly span the entire digital value range [2^N-1,0] to maximize contrast
- AHE does this on a spatial sub-region basis in an image rather than the entire image
- Fuji ‘Dynamic Range Control’ (DRC) a version of AHE that operates on sub-regions of digital values

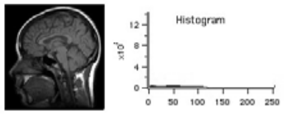
Histogram Equalization



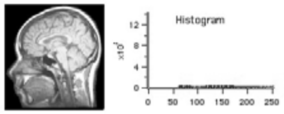
c.f. http://www.wavemetrics.com/products/igorpro/imageprocessing/imagetransforms/histmodification.htm

Global and Adaptive Histogram Equalization

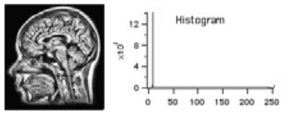
The following images illustrate the differences between global and adaptive histogram equalization.



MR image with the corresponding gray-scale histogram. The histogram has a peak at minimum intensity consistent with the relatively dark nature of the image.



Global histogram equalization and the final gray-scale histogram. Comparing the results with the figure above we can see that the distribution was shifted towards higher values while the peak at minimum intensity remains.



Adaptive histogram equalization shows better contrast over different parts of the image. The corresponding gray-scale histogram lacks the mid-levels present in the global histogram equalization as a result of setting a high contrast level.

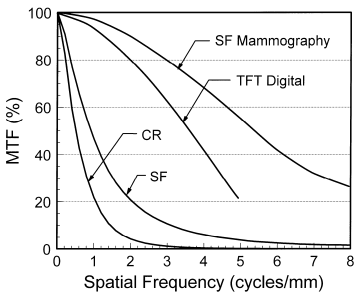
c.f. http://www.wavemetrics.com/products/igorpro/imageprocessing/imagetransforms/histmodification.htm

Why is Image Processing Important or Necessary?

- Digital detector elements don’t operate identically (gain)
- Some pixels go bad or have a large offset (correction)
- The wide dynamic range of digital detectors → flat image
- Must manipulate image digitally to view → window/level
- May wish to reduce image noise (convol., median, σ)
- May wish to enhance the spatial frequency → MTF
  - Convolution processing (unsharpmasking)
  - Fourier and wavelet processing
- Try to automatically set gray-scale mapping of image
  - Histogram equalization

Contrast vs. Spatial Resolution in Digital Imaging

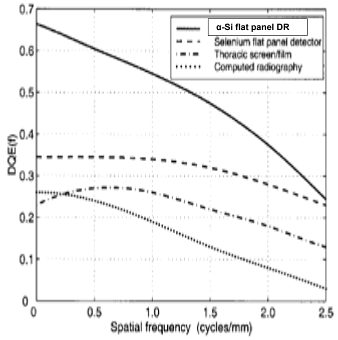
- S-F mammography can produce images w/ > 20 lp/mm
- According to Nyquist criterion would require 25 μm/pixel resulting in a 7,200 x 9,600 image (132 Mbytes/image)
- Digital systems have inferior spatial resolution
- However, due to wide dynamic range of digital detectors and image processing capabilities, digital systems have superior contrast resolution



c.f. Bushberg, et al. The Essential Physics of Medical Imaging 2nd ed., p. 315

Digital Imaging Systems and DQE

- Remember the equation for DQE(f)?
- DQE(f) =
- How can we account for this?
- Both CR and the screens in film/screens made thin
- Film higher spatial resolution than CR
- DQE higher for α-Si systems using CsI rather than α-Se (mean x-ray E & Z)
- α-Si DQE falling off more rapidly than α-Se (indirect)



c.f. Granfors and Aufrichtig. Medical Physics 27(2000), p. 1330

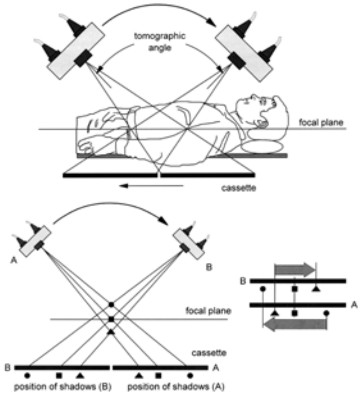
Review Question

- **15**. Which of the following does NOT involve image processing:
- A. Background subtraction
- B. Energy subtraction
- C. Histogram equalization
- D. K-edge filtering
- E. Low-pass filtering

Review Question

- **14**. Processing a digital x-ray image by unsharpmask enhancement would increase the:
- A. Bit depth per pixel
- B. Limiting spatial resolution
- C. Matrix size
- D. Patient dose
- E. Visibility of edges

Geometric (Linear) Tomography

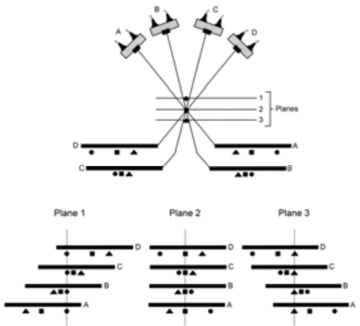


- With the advent of CT, geometric tomography has only limited clinical utility where only one or a few planes of objects with high contrast are desired, e.g., IVP
- Desired slice through patient set at pivot point (focal plane)
- The tomographic process blurs out regions outside the focal plane, but still contributes to overall loss of contrast
- Larger tomographic angles result in a lessening of out of plane contributions
- High dose, comparable to CT for many tomographic slices

c.f. Bushberg, et al. The Essential Physics of Medical Imaging, 2nd ed., p. 318

Digital Tomosynthesis

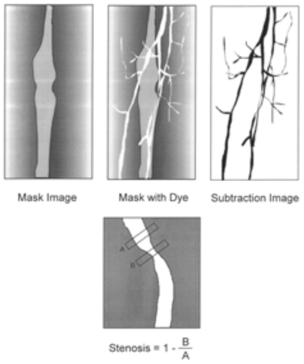
- Improved version of geometric tomography where a digital detector saves an image at each of several tube angles
- This allows reconstruction of multiple planes through the object through shifting the various images through a certain distance before summing them
- Much more dose efficient, but still suffers from out of plane blurring effects
- Either CR or DR used



c.f. Bushberg, et al. The Essential Physics of Medical Imaging, 2nd ed., p. 320

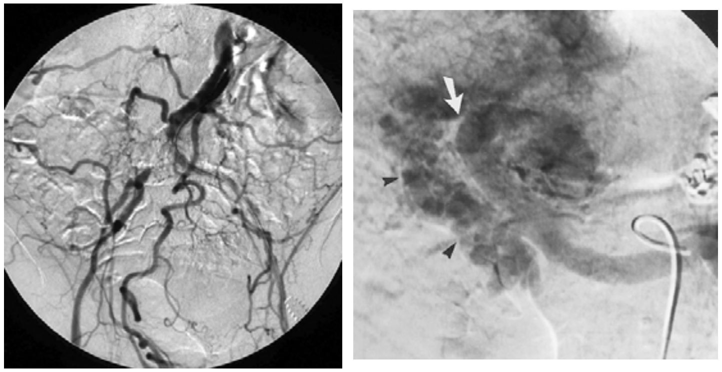
Temporal Subtraction

- Digital Subtraction Angiography (DSA) – usually 1K resolution
- Mask (background) subtracted from images during/post contrast injection: Δ < 1% trans. visualized
- Motion can cause misregistration artifacts
- Digital value proportional to contrast concentration and vessel thickness
    - $I_s = \ln(I_m) - \ln(I_c) = \mu_{vessel} \cdot t_{vessel}$
- Temporal subtraction works best when time differences between images is short
- Possible to spatially warp images taken over a longer period of time



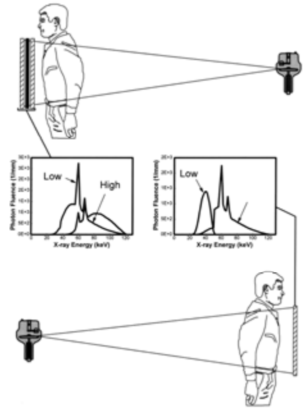
c.f. Bushberg, et al. The Essential Physics of Medical Imaging, 2nd ed., p. 322

Bas-relief Effect from Subtraction Misregistration



Dual-Energy Subtraction

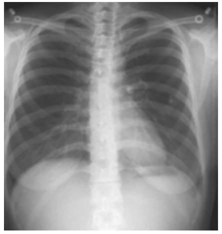
- Exploits differences between the Z of bone ($Z_{eff} \approx 13$) and soft tissue ($Z_{eff} \approx 7.6$)
- Images taken either at two different kVp (two-shot), or
- One image (one-shot) taken with energy separation provided by a filter (sandwich)
- $I_{out} = \log_e(I_{low}) - R \cdot \log_e(I_{high})$, where R is altered to produce soft-tissue predominant or bone predominant images
- Chest DR @ UWMC/SCCA



c.f. Bushberg, et al. The Essential Physics of Medical Imaging 2nd ed., p. 324.

Dual-Energy Subtraction

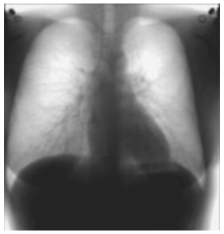
60 kVp Exposure



120 kVp Exposure

Bone Computed DE Image

Soft Tissue Computed DE Image



c.f. Bushberg, et al. The Essential Physics of Medical Imaging 2nd ed., p. 325.

Review Question

- **22**. The matrix size in a DSA image is typically:

- A. 128 x 128
- B. 256 x 256
- C. 512 x 512
- D. 1024 x 1024
- E. 2048 x 2048

Review Question

- **25**. Changing the DSA matrix from 1024^2 to 2048^2 would NOT increase the:

- A. Data digitization rate
- B. Data storage requirement
- C. Image processing time
- D. Pixel size
- E. Spatial resolution

Review Question

- **D51**. A flat panel digital radiographic detector has a square 20 x 20 cm image receptor field. The full field of the detector is coupled to a nominal 1024 x 1024 CCD array. The relative spatial resolution (lp/mm) when going from a 20 x 20 cm to a 10 x 10 cm field of view is:
- A. Four times better
- B. Twice as good
- C. The same
- D. Half as good
- E. One fourth as good

Review Question

- **17**. The Nyquist frequency for a 1K digital photospot image (25 cm image intensifier diameter) is:

- A. 1 lp/mm
- B. 2 lp/mm
- C. 4 lp/mm
- D. 8 lp/mm
- E. 10 lp/mm

- F_N (lp/mm) = 1/2a = 1/2(250mm/1024 lines) = 2.048 $\approx$ 2