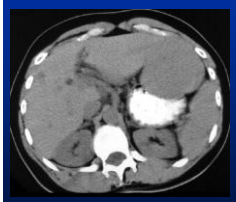
Dual-Energy CT: Acquisition and Processing

Norbert J. Pelc, Sc.D.
Departments of Radiology and Bioengineering
Stanford University

Acknowledgments

- Material from many colleagues
- Research support:
  - GE Healthcare
  - Philips Healthcare

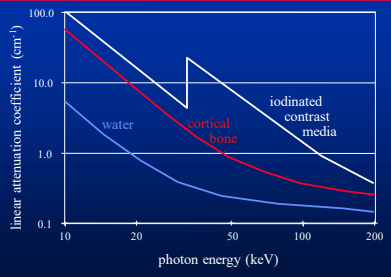
Motivation



www.uhrad.com/
ctarc/ct153b2.jpg

- Conventional CT measures μ at one effective energy
- Does lower HU mean lower density or lower Z?
- Can we get more material specificicity
  - more diagnostic information
  - extrapolation to other energies

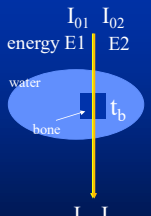
attenuation coefficients

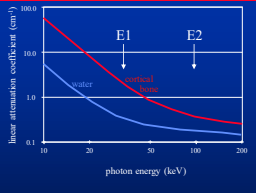


OUTLINE

- Physical principles of multi-energy x-ray measurements
- Data processing
- Methods for obtaining spectral measurements
- Closing comments

unknown thickness two known materials



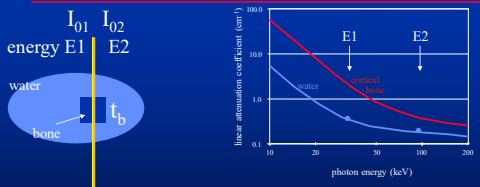


$$I_1 = I_{01}\, e^{-(\mu_{w1} t_w + \mu_{b1} t_b)}$$

$$I_2 = I_{02}\, e^{-(\mu_{w2} t_w + \mu_{b2} t_b)}$$

solve for t_w and/or t_b

unknown thickness two known materials



$$t_b = A\{ \ln(I_{01}/I_1) - (\mu_{w1}/\mu_{w2})(\ln(I_{02}/I_2)\}$$

scale for lost bone signal

subtract water

makes the water contribution at E2 match that at E1

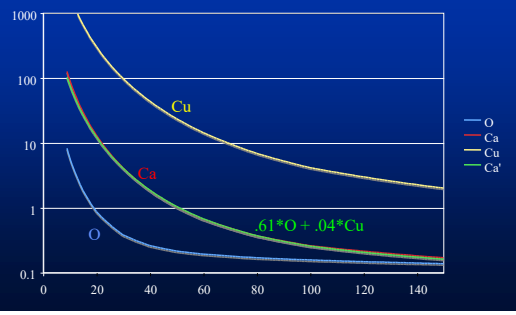
material analysis with absorptiometry

- 2 energies ⇨ 2 materials
- can we generalize this? N energies for N materials?
- limitation: two strong interaction mechanisms
  Compton scattering and photoelectric absorption
  ~ Same energy dependence for all elements

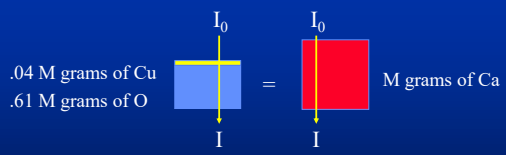
basis material decomposition

- Barring a K-edge:
  $\mu(E) \sim a*\text{Compton}(E) + b*\text{Photoelectric}(E)$
  2 fundamental parameters determine material behavior
  effective atomic number, electron density
- any material can be modeled as a weighted sum of two other materials
  $\mu(E) \sim \alpha^* \mu_i(E) + \beta^* \mu_j(E)$
  basis material decomposition
- in any projection measurement, we can only isolate two materials

Basis material decomposition



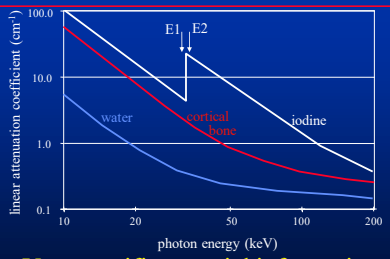
Basis material decomposition



Indistinguishable at any x-ray energy above their K-edge
Common "basis functions"

- Photoelectric and Compton (Z_{eff} and electron density)
- Basis "materials" (water and iodine, aluminum and plastic, etc)

K-edge subtraction

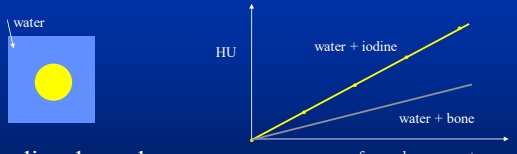


Very specific material information
Ideally uses very narrow spectra

Dual-energy processing

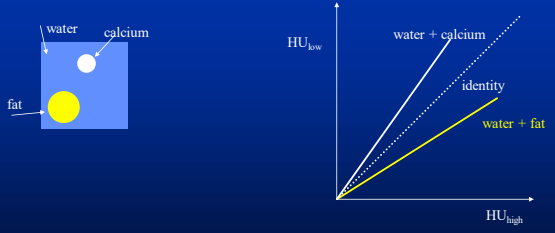
- goals:
  tissue characterization, material canceled images,
  high SNR ("conventional") images,
  equivalent monoenergetic images
- reconstruct images in the normal manner, and combine HU images
  easy to implement
- combine projection data prior to reconstruction
  somewhat more difficult
  requires aligned projections
  enables "exact" beam hardening correction

Two known materials single energy CT



line depends on:
the two materials
x-ray energy
how they mix, e.g., displace or dissolve

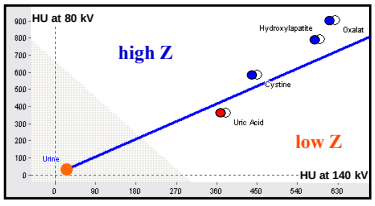
Analyzing dual energy CT

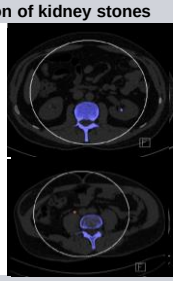


Applications of Dual Energy CT

SIEMENS

Another image based application : **characterization of kidney stones**

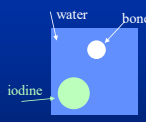


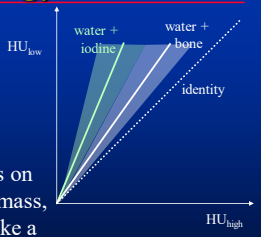


- Uric acid stones can be differentiated from other renal calculi

Courtesy of University Hospital of Munich - Grosshadern / Munich, Germany

Three known materials dual energy CT



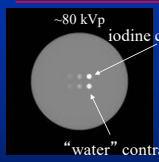


segmentation accuracy depends on the atomic number difference, mass, energy difference, and noise (like a subtraction process)

Dual-energy processing

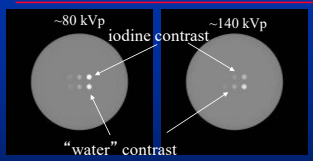
- goals:
  tissue characterization, material canceled images,
  high SNR ("conventional") images,
  equivalent monoenergetic images

Dual energy processing



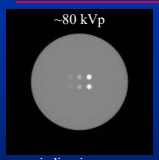
monochromatic 55 keV simulation
comparable to ~ 80 kVp

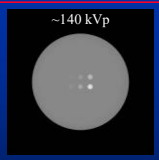
Dual energy processing

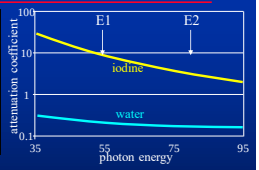


monochromatic simulation
comparable to 80/150 kVp

Dual energy processing

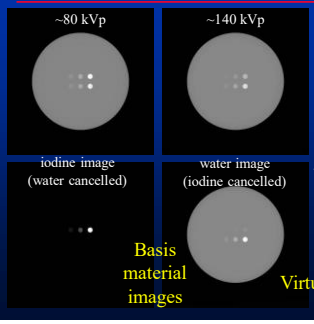






Iodine image = $a \cdot (Image_{80} - b \cdot Image_{140})$

Dual energy processing



Water image =
$c \bullet (Image_{80} - d \bullet Image_{140})$

Virtual non-contrast (VNC) image

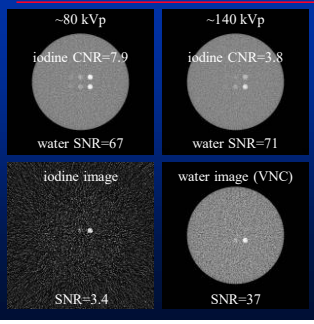
VNC ≠ NC

CT numbers for non-water tissues are not the same

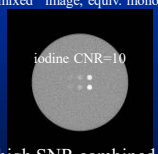
Difference in CT number between VNC and true non-contrast image depends on the tissue and the choice of basis materials.

SNR of VNC image is much lower

Dual energy processing



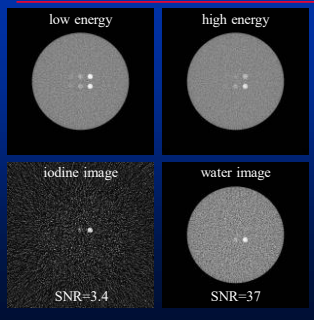
optimal combination "gray scale" image
("mixed" image, equiv. monoE image)



high SNR combined image

material cancelled images
have increased noise

Noise

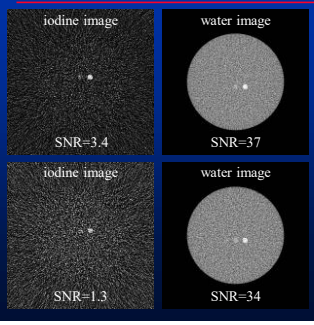


Iodine data ~
a • $Data_{low}$ - b • $Data_{high}$

$\sigma^2 \sim a^2 \sigma_{low}^2 + b^2 \sigma_{high}^2$

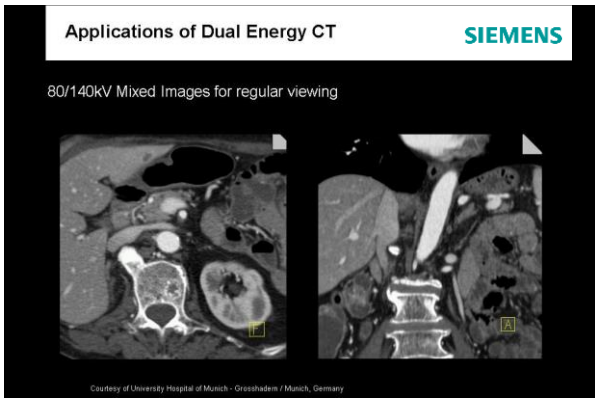
depends on:
- specific energies
- allocation of dose to the two measurements

Noise depends on dose allocation



with 80/140 kVp dose allocation that maximizes iodine SNR

↓ 80 kVp dose, ↑140 kVp dose
same total dose



Applications of Dual Energy CT

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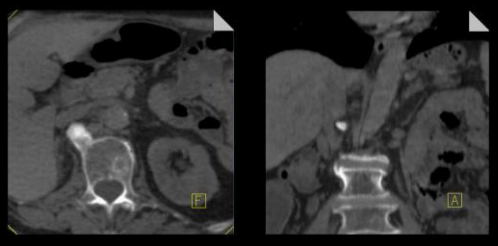
Virtual Non-Contrast Image



Courtesy of University Hospital of Munich - Grosshadern / Munich, Germany

Applications of Dual Energy CT

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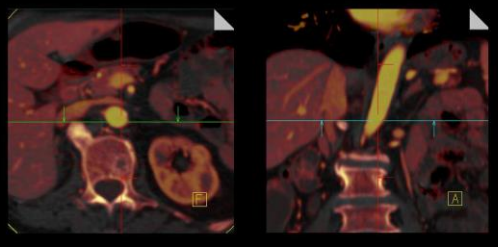
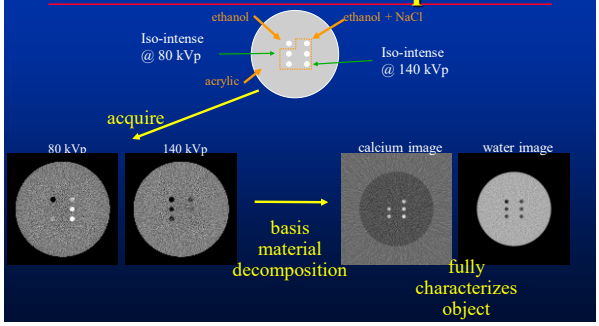
Virtual Non-Contrast Image with Iodine (Overlay in Color)

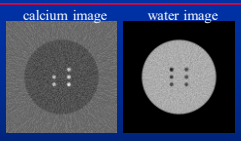


Courtesy of University Hospital of Munich - Grosshadern / Munich, Germany

Dual energy - Basis material decomposition



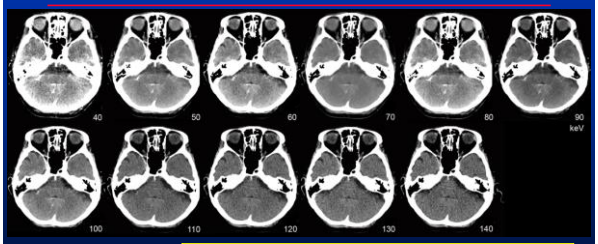
Equivelent monoenergetic images



fully characterizes object

image (E) =
(calcium image)*μ_{Ca}(E) + (water image)*μ_{W}(E)

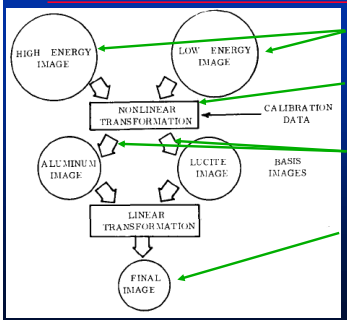
Equivelent monoenergetic images



Postma et al, 2015

Monochromatic energy is a display parameter.
Contrast and SNR vary with selected energy.
Highest SNR image is similar to "mixed" image

Pre-reconstruction processing



polychromatic data

includes accurate beam hardening

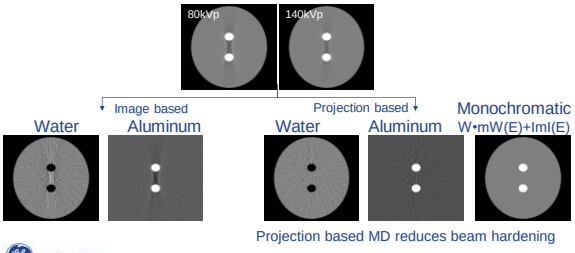
reconstruction

material cancelled images

monoenergetic images

Lehmann et al: Med Phys 8, 659-67, 1981.

Images from projection-based recon



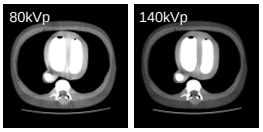
Projection based MD reduces beam hardening

Courtesy of Uri Shreter, GE Healthcare

Monochromatic CT from projection-based recon

Heart Chamber Phantom, 8.3%

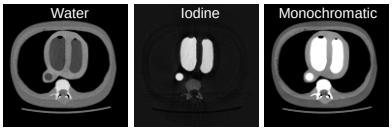
L = 0, W = 350 HU



Material separation

Monochromatic CT – keV tuned

Natively eliminates beam hardening CT # shifts



Courtesy of R. Senzig, GE Healthcare

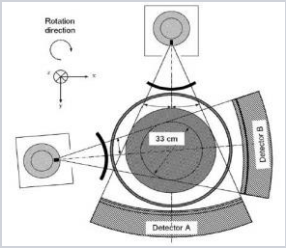
Data acquisition implementations

- Sequential scans at different kVp
  motion sensitivity > scan time

syngo Dual Energy
- Principle of Dual Energy

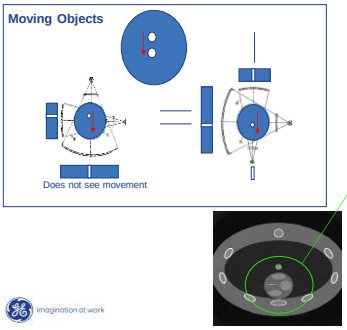
SIEMENS

SOMATOM Definition Flash

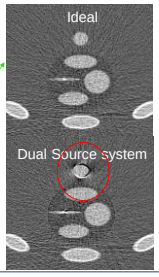


Two X-ray tubes, 1st: 80 or 100 kVp, 2nd: 140 kVp

Dual Source Challenge: Inconsistent scans



Moving Phantom Simulation



Courtesy of R. Senzig, GE Healthcare

Rapid kVp switching Dual energy CT

- Requires fast generator and detectors
- Dose allocation controlled by dwell time
- Difficult to switch filters

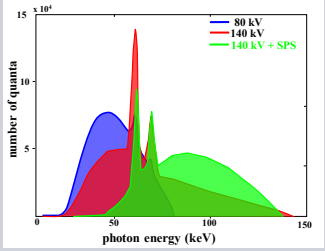
Adapted from slide courtesy of Uri Shreter, GE Healthcare

syngo Dual Energy
- Principle of Dual Energy

SIEMENS

SOMATOM Definition Flash

S_1: 80 kV S_2: 140 kV + SPS



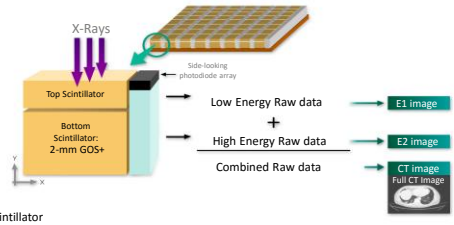
SPS= Selective Photon Shield

PHILIPS

NanoPanel Prism

Perfect alignment

Simultaneous alignment in time and space



Top scintillator
- Effective atomic number small but does not sacrifice light output
- Thickness optimized for energy separation and low-energy image noise

Bottom scintillator absorbs 99.5% of high-energy spectrum

Layered detector

- relatively poorer spectral separation
- simultaneous dual energy sensing

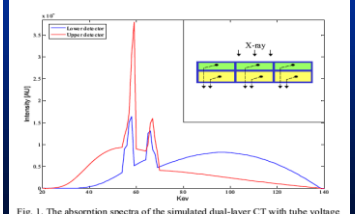
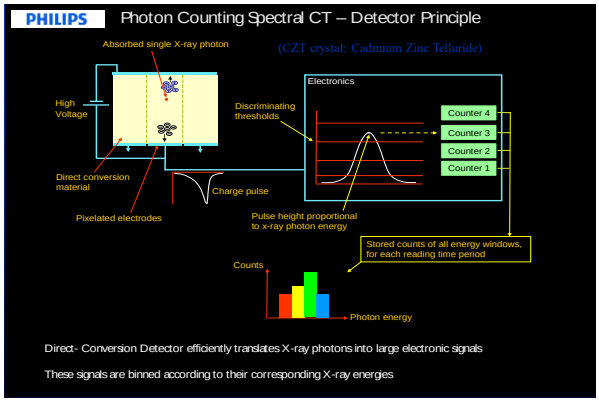


Fig. 1. The absorption spectra of the simulated dual-layer CT with tube voltage of 140 Kv. Insert shows detector configuration with partial absorber (green) and full absorber (yellow).

Carmi R, Naveh G, and Altman A: IEEE NSS M03-367, 1876-78, 2005

Twin-beam dual energy

- different detector rows have different filters
- helical acquisition with lower pitch
- modest spectral separation unless most of the flux is discarded



Photon counting detectors

- very promising
- main challenges:
  - count rate capability (count loss and pile-up)
  - signal sharing across neighboring elements
  - imperfect and count-rate dependent energy response
  - cost

Spectral separation

- very critical for SNR efficiency, separation robustness, etc.
- implementations
  - photon counting with K-edge filter (ideal)
  - photon counting with energy analysis (ideal)
  - different kVp and filtration
  - different kVp
  - layered detector, different filters

↑ better spectral separation and dose efficiency

Data acquisition implementations

- Sequential scans at different kVp
  motion sensitivity > scan time
- Split-filter helical
  motion sensitivity ~ T_{rot}
- Two sources at 90º on the same gantry
  some motion sensitivity (~ 25% T_{rot})
- Switching kVp within a single scan
- Energy discriminating detectors
  layered detector, photon counting

↓ better immunity to motion

Summary of commercial systems

- Siemens has two implementations
  - two sources, different kVp (80 or 100 /140) and filtration, direct control of mA
  - split-filter helical
- GE: single source with rapid switching, same filter for both kVps, control mAs by dwell time
- Philips: dual-layer detector, complete kVp mAs control
- Lots of R&D work, especially on photon counting detectors

Summary

- spectral CT does not require higher radiation dose
- perfect beam hardening correction (pre-recon)
- effective monoenergetic images, more accurate RTP and PET attenuation correction
- some material specificity (e.g., average atomic number, some disease specificity
- improved image segmentation

Summary

- virtual noncontrast image
  perfectly registered and simultaneously acquired
  Beware of noise propagation. Separate optimized scans probably have lower total dose
- isolate contrast media from calcified plaque
  difficult for small amounts of either
- material specific images are noisy
- "tomochemistry" or molecular imaging?
  only for high concentrations (10^2 MRI, $>10^8$ PET)

Thank You