Dual Energy CT for Density Measurements

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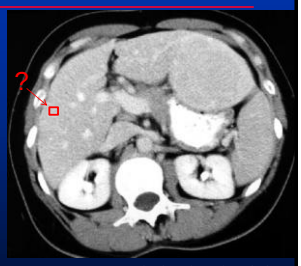
What is in a voxel?

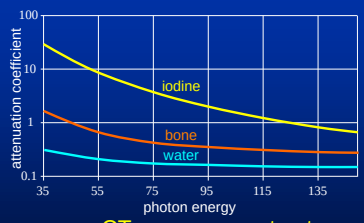
CT number depends on:
- inherent tissue properties (chemical composition, density)
- x-ray spectrum
- administered contrast media

Can we be more specific?



www.uhrad.com/ ctarc/ct153b2.jpg

Attenuation coefficients depend on photon energy

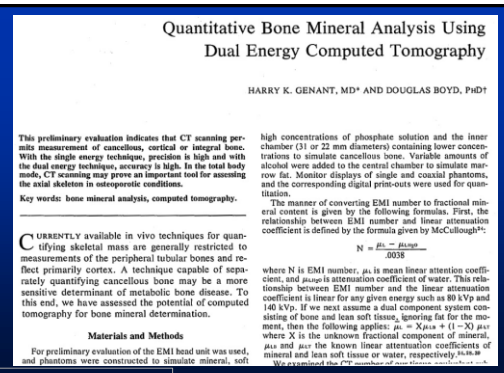


use CT measurements at multiple energies for material specificity and improved quantitation

Motivation

"Two pictures are taken of the same slice, one at 100 kV and the other at 140 kV...so that areas of high atomic numbers can be enhanced... Tests carried out to date have shown that iodine (Z=53) can be readily differentiated from calcium (Z=20)".

G.N. Hounsfield, BJR 46, 1016-22, 1973.



Quantitative Bone Mineral Analysis Using Dual Energy Computed Tomography

HARRY K. GENANT, MD* AND DOUGLAS BOYD, PhD†

This preliminary evaluation indicates that CT scanning permits measurement of cancellous, cortical or integral bone. With the single energy technique, precision is high and with the dual energy technique, accuracy is high. In the total body mode, CT scanning may prove an important tool for assessing the axial skeleton in osteoporotic conditions.

Key words: bone mineral analysis, computed tomography.

CURRENTLY available in vivo techniques for quantifying skeletal mass are generally restricted to measurements of the peripheral tubular bones and reflect primarily cortex. A technique capable of separately quantifying cancellous bone may be a more sensitive determinant of metabolic bone disease. To this end, we have assessed the potential of computed tomography for bone mineral determination.

Materials and Methods

For preliminary evaluation of the EMI head unit was used, and phantoms were constructed to simulate mineral, soft

high concentrations of phosphate solution and the inner chamber (31 or 22 mm diameters) containing lower concentrations to simulate cancellous bone. Variable amounts of alcohol were added to the central chamber to simulate marrow fat. Monitor displays of single and coaxial phantoms, and the corresponding digital print-outs were used for quantitation.

The manner of converting EMI number to fractional mineral content is given by the following formulas. First, the relationship between EMI number and linear attenuation coefficient is defined by the formula given by McCullough[14]:

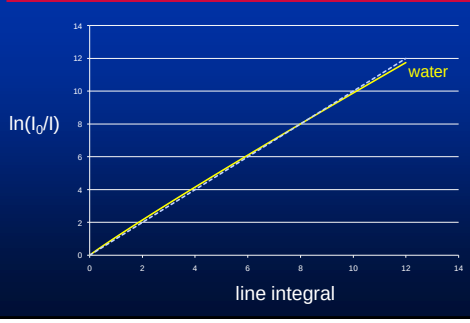
$$N = \frac{\mu_L - \mu_{LWP}}{.0038}$$

where N is EMI number, μ_L is mean linear attention coefficient, and μ_{LWP} is attenuation coefficient of water. This relationship between EMI number and the linear attenuation coefficient is linear for any given energy such as 80 kVp and 140 kVp. If we next assume a dual component system consisting of bone and lean soft tissue, ignoring fat for the moment, then the following applies: $\mu_L = X\mu_{LB} + (1-X)\,\mu_{LT}$ where X is the unknown fractional component of mineral, μ_{LB} and μ_{LT} the known linear attentuation coefficients of mineral and lean soft tissue or water, respectively.[14,19,20]

We examined the CT number of our tissue equivalent ...

Genant et al, Inv Radiol, 1977.

Beam hardening

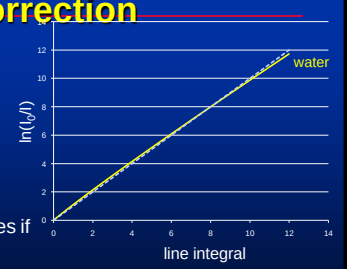


Single energy CT: "water" beam hardening correction

Accurate for "water-like" materials of any density

Methods:
MLE is accurate
polynomials are fast:
$p = a_1L + a_2L^2 + a_3L^3 + \ldots$

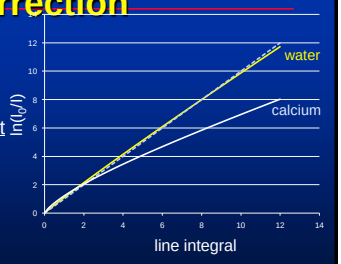
Provides accurate densities if Z_{eff} is known and uniform



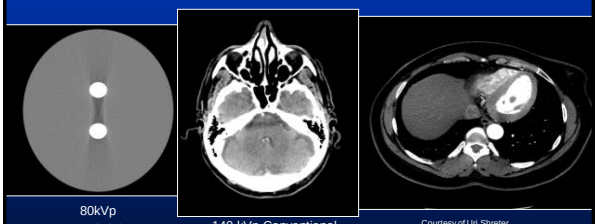
Single energy CT: "Water" beam hardening correction

Correction fails with materials of very different Z_{eff}

Results in local and distant errors and artifacts



Beam hardening



80kVp
Courtesy of Uri Shreter
GE Healthcare

140 kVp Conventional CT
Courtesy of Dr. Joseph Hoxworth
Mayo Clinic Arizona

Courtesy of Uri Shreter
GE Healthcare

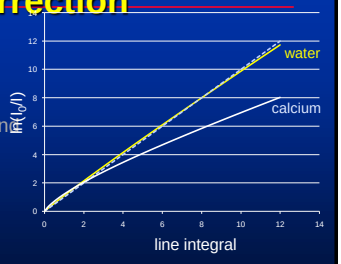
Single energy CT: "Water" beam hardening correction

Correction fails with materials of very different Z_{eff}

Results in density errors and artifacts

Multi-pass corrections can help if materials and distribution are known

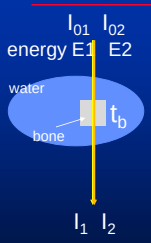
Can we do better?

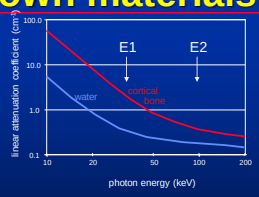


OUTLINE

- Physical principles of multi-energy x-ray measurements
- Signal processing
- Quantitation opportunities and challenges
- Data acquisition
- Summary

unknown thickness two known materials



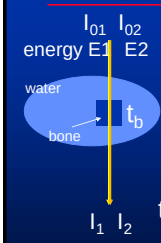


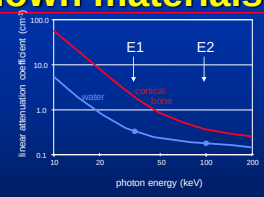
$$I_1 = I_{01}\, e^{-(\mu_{w1} t_w + \mu_{b1} t_b)}$$

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

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

dual energy x-ray absorptiometry (DEXA)

2 energies ⇨ 2 materials

material analysis with absorptiometry

- 2 energies ⇨ 2 materials
- can we generalize this? N energies for N materials?
- limitation: two strong interaction mechanisms

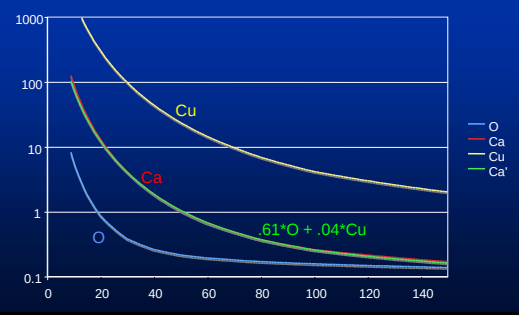
  Compton scattering and photoelectric absorption

  Barring a K-edge in the spectrum, the energy dependence of each is the same for all

basis material decomposition:

- barring a K-edge:
  - any material acts like a combination of pure photoelectric and Compton
  - any material can be modeled as a weighted sum of two other materials

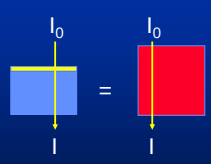
Basis material decomposition



Basis material decomposition

.04 M grams of Cu

.61 M grams of O



M grams of Ca

Indistinguishable at any x-ray energy above their K-edge

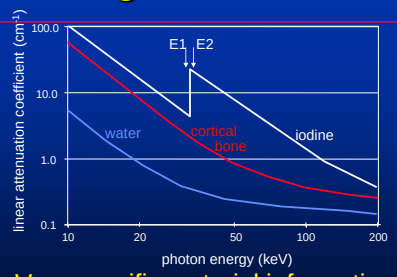
Common "basis materials":

iodine and water, aluminum and plastic

basis material decomposition:

- barring a K-edge:
  - any material acts like a combination of pure photoelectric and Compton
  - any material can be modeled as a weighted sum of two other materials
  - "basis material" decomposition
  - in any projection measurement, we can only isolate two materials
- Material parameters:
  - effective atomic number and electron density
  - amounts of two basis materials
  - can be measured using 2 x-ray energies

K-edge subtraction

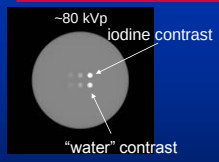


Very specific material information
2 narrow spectra, or 3 spectra

Dual-energy processing

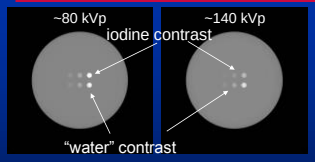
- reconstruct images in the normal manner, and combine HU images
  - easy to implement
- combine projection data prior to reconstruction
  - somewhat more difficult
  - requires aligned projections
  - enables “exact” beam hardening correction

Dual energy processing



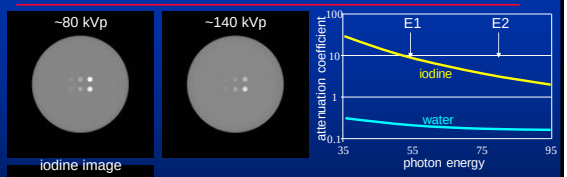
monochromatic 55 keV simulati
comparable to ~ 80 kVp

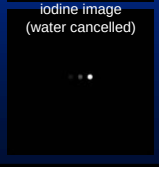
Dual energy processing



monochromatic simulation
comparable to 80/150 kVp

Dual energy processing



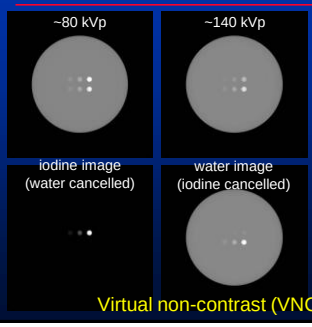


Iodine image = a • (Image$_{80}$ - b • Image$_{140}$)

restore reduced iodine signal

amplify 140 kVp water signal

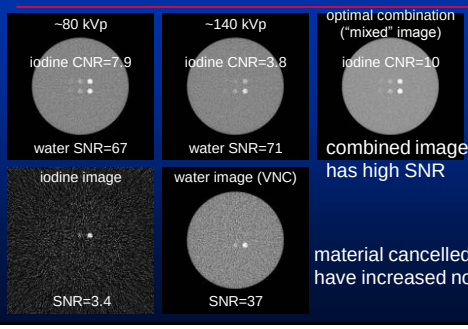
Dual energy processing



Water image = c • (Image$_{80}$ - d • Image$_{140}$)

Virtual non-contrast (VNC) image

Dual energy processing



combined image has high SNR

material cancelled images have increased noise

Applications of Dual Energy CT

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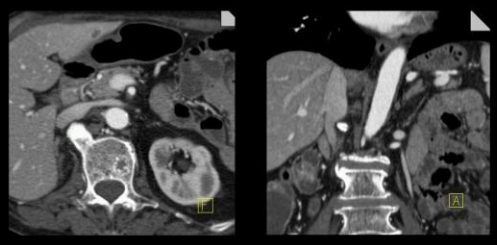
80/140kV Mixed Images for regular viewing

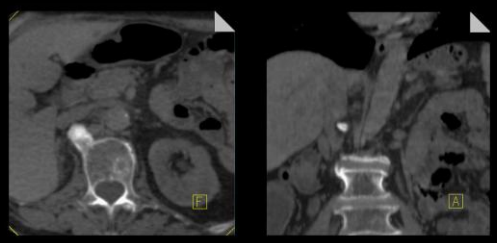


Courtesy of University Hospital of Munich - Grosshadern / Munich, Germany

Applications of Dual Energy CT

SIEMENS

Virtual Non-Contrast Image



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

Applications of Dual Energy CT

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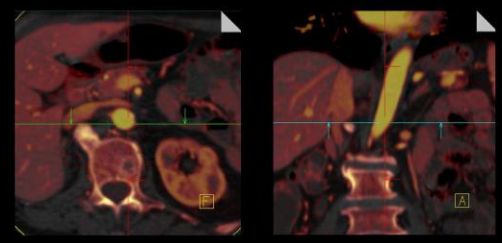
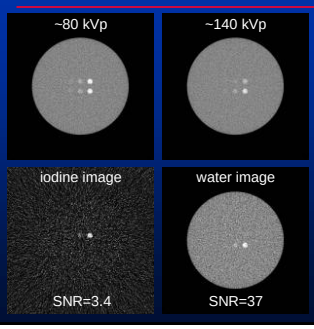
Virtual Non-Contrast Image with Iodine (Overlay in Color)



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

Noise

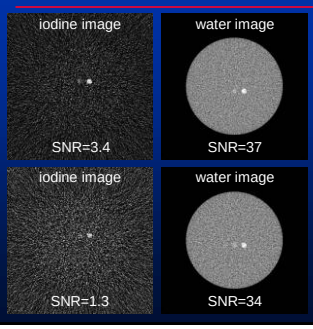


Iodine image =
$a \cdot (Image_{80} - b \cdot Image_{1}$

$\sigma^2 = a^2 \cdot (\sigma_{80}^2 + b^2 \cdot \sigma_{140}^2)$

depends on dose allocation to the 80 kVp and 140 kVp images

Noise depends on dose allocation



dose allocation that maximizes iodine SNR

↓ 80 kVp dose ↑ 140 kVp dose
same total dose

Dual energy processing

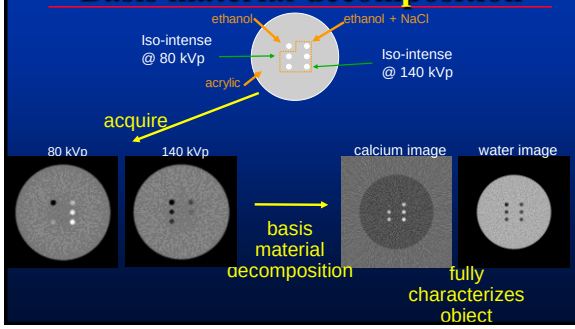
$L_L = \ln(I_{0L}/I_L)$, $L_H = \ln(I_{0H}/I_H)$
m_A, m_B = amounts of basis materials

With monochromatic beams, L's are linear functions of m's, so

$$m_A = a_0 L_L + a_1 L_H, \quad m_B = b_0 L_L + b_1 L_H$$

With polychromatic beams, functions are nonlinear. Approaches:
1) iterative solutions (e.g., MLE)
   accurate but slow
2) polynomial approximation
   $m_A = a_0 L_L + a_1 L_H + a_2 L_L^2 + a_3 L_H^2 + a_4 L_L L_H + \ldots$
   accuracy depends on polynomial order, dynamic range, etc.

Dual energy - Basis material decomposition



Dual energy - Basis material decomposition

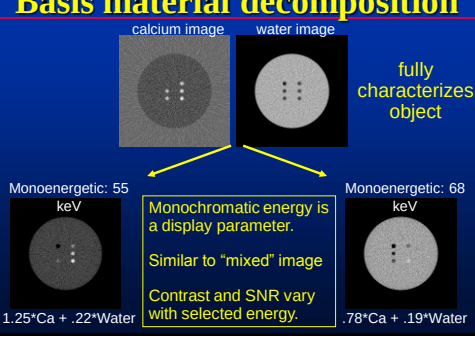
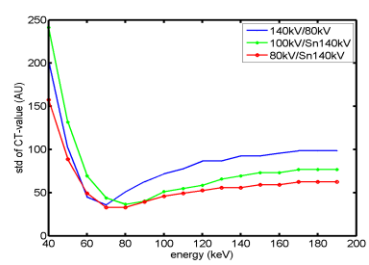


Image Based Methods SIEMENS

- **Calculation of Energy-Selective Images: Noise @ different energies**



B: Heismann, B. Schmidt, T. Flohr
Short course 987
SPIE Medical Imaging Conference 2010

Application of monoenergetic images

- image interpretation
  hig
- extra
  MV

Proc Soc Photo Opt Instrum Eng. 2010 May 22; 7622: 762225–762228. doi:10.1117/12.844075.

Noise and Bias Properties of Monoenergetic Images from DECT used for Attenuation Correction with PET/CT and SPECT/CT

Ting Xia[1], **Adam M Alessio**, and **Paul E Kinahan**
University of Washington, Seattle, WA USA 98195

Abstract

We evaluate the energy dependent noise and bias properties of monoenergetic images synthesized from dual-energy CT (DECT) acquisitions used to estimate attenuation coefficients at PET or SPECT energies. This is becoming more relevant with the increased used of quantitative imaging by PET/CT and SPECT/CT. There are, however, variations in the noise and bias properties of synthesized monoenergetic images as a function of energy. We used analytic approximations and simulations to estimate the bias and noise of synthesized monoenergetic images of a water-filled cylinder from 10 to 525 keV. The dual-energy spectra were based on the GE Lightspeed VCT scanner at 80 and 140 kVp. Both analytic calculations and simulations for increasing energy the relative noise plateaued near 140 keV (i.e. SPECT with ^{99m}Tc), and then remained constant with increasing energy up to 511 keV and beyond (i.e. PET). If DECT is being used for attenuation correction at higher energies, there is a noise amplification that is dependent on the energy of the synthesized monoenergetic image of linear attenuation coefficients. For SPECT and PET imaging the bias and noise levels of DECT based attenuation correction is unlikely to affect image quality.

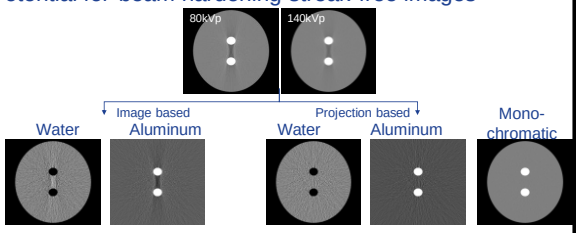
Beam hardening correction

If basis material assumption holds (e.g., no K-edge materials),

- nonlinear decomposition exactly handles polychromaticity
- exact beam hardening correction
- projection domain processing is preferred

Monochromatic CT from HDCT projection-based recon

Potential for beam hardening streak-free images



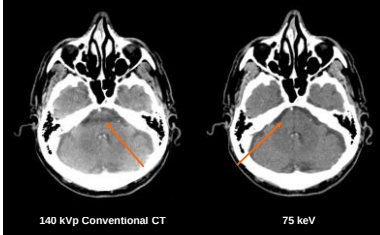
Projection based MD reduces beam hardening

imagination at work

Courtesy of Uri Shreter, GE Healthcare

Discovery CT750 HD

Posterior Fossa Artifact Reduction



140 kVp Conventional CT

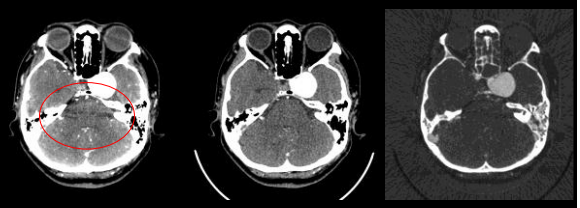
75 keV

Clinical Value

- Spectral or Monochromatic images have a reduced beam hardening effect vs polychromatic reconstructed images
- Beam hardening artifacts can obstruct the the clinician's interpretation of important brain anatomy
- Notice the visualization improvement of the brain anatomy between the petrous bones

Images Courtesy of Dr. Joseph Hoxworth
Mayo Clinic Arizona

Posterior Fossa Streak Artifacts are Removed / Lesion Verification

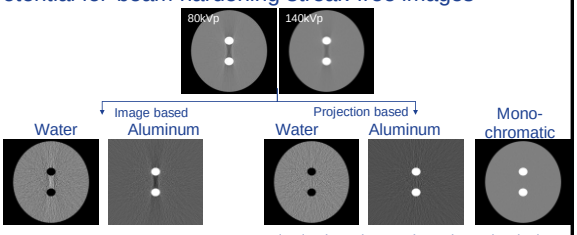


80 kVp | 100 keV | MD Iodine

Images Courtesy of Dr. Amy Hara, Mayo Clinic, Scottsdale, AZ

Monochromatic CT from HDCT projection-based recon

Potential for beam hardening streak-free images



Projection based MD reduces beam hardening

imagination at work

Courtesy of Uri Shreter, GE Healthcare

Beam hardening correction

If basis material assumption holds (e.g., no K-edge materials),

- nonlinear decomposition exactly handles polychromaticity
- exact beam hardening correction
- projection domain processing is preferred
- starting with image domain data is possible

Exact dual energy material decomposition from inconsistent rays (MDIR)

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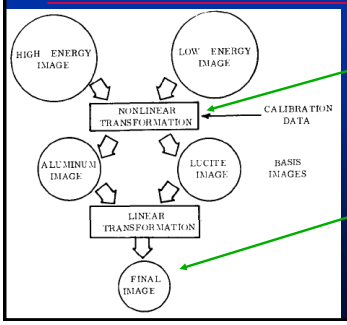
(Received 16 March 2010; revised 17 November 2010; accepted for publication 14 December 2010; published 11 January 2011)

Purpose: Dual energy CT (DECT) allows calculating images that show the spatial distribution of the electron density and the atomic number or, more common, images of two basis material densities. In contrast, the Hounsfield unit that is shown in standard CT images is a measure of the x-ray attenuation, which is a function of the atomic number and electron density. To acquire additional

Maaß et al, Med Phys, 2011

Dual energy processing



accurate beam hardening built-in

material cancelled images

monoenergetic images

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

Quantitation

- Decomposition yeilds local density of basis materials
- Can convert basis materials – i.e., calculate density of any two materials
- Quantifies actual materials only if they match basis materials

- Suppose bases are a and b, but voxel has a and c
  - Truth: (m_a, m_c)
  - Appearance: $(m_a,0)+ (m_a', m_b') = ([m_a+m_a'] , m_b')$

- (Local) errors results if the materials are wrong (e.g., tissue vs. water vs. fat)
- Virtual non contrast ≠ precontrast image

Applications of Dual Energy CT

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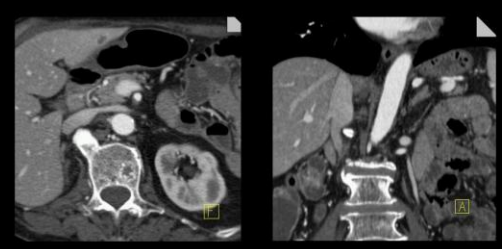
80/140kV Mixed Images for regular viewing

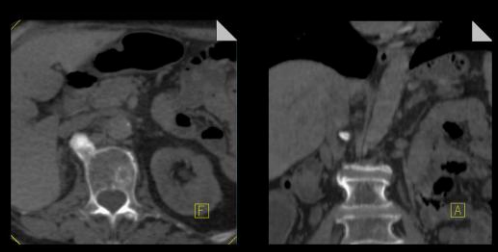


Courtesy of University Hospital of Munich - Grosshadern / Munich, Germany

Applications of Dual Energy CT

SIEMENS

Virtual Non-Contrast Image

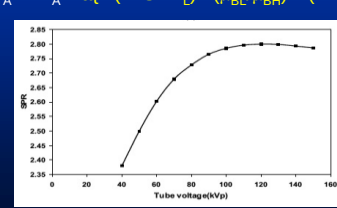


Courtesy of University Hospital of Munich - Grosshadern / Munich, Germany

Scatter

ideal: $m_A = a\{L_L - (\mu_{BL}/\mu_{BH})\ L_H\}$

w/ scatter: $m_A' = m_A - a\{\ln(1+SPR_L) - (\mu_{BL}/\mu_{BH})\ \ln(1+SPR_H)\}$



20 cm water absorber

Ghafarian et al, IEEE Nuclear Science Symposium, 2007.

Scatter

Correction for scattered radiation and other background signals in dual-energy computed tomography material thickness measurements

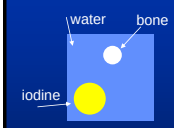
J. R. Vetter and J. E. Holden[a)]
Departments of Medical Physics and Radiology, University of Wisconsin, Madison, Wisconsin 53706

(Received 16 December 1987; accepted for publication 12 May 1988)

Both beam hardening and the detection of scattered radiation cause nonlinearity errors in x-ray computed tomography (CT), leading to artifacts and CT number inaccuracies. Dual-energy measurements can be used to correct beam hardening effects to a high degree of accuracy. However, in the imaging of thick body sections the transmitted intensity of the primary beam is low, making scatter the most significant cause of CT number inaccuracy. Furthermore, the scatter-to-primary ratio is energy dependent, causing a shift in the apparent effective atomic number of the absorbing material. We have measured scatter under a variety of conditions on a third generation CT scanner with dual-energy capability in order to determine its effect on the accuracy of quantitative measurements. The effects of off focus radiation, detector cross-talk, and detector dark current were also accounted for in the analysis of the measured scatter data. Our results indicate that on this well collimated system, an accurate correction for scatter can be made based on the detected intensity in projections.

Vetter et al, Med Phys 1998.

three known materials



$$\mu_1 = \mu_{w1} f_w + \mu_{b1} f_b + \mu_{I1} f_I$$

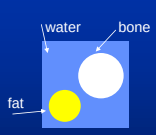
$$\mu_2 = \mu_{w2} f_w + \mu_{b2} f_b + \mu_{I2} f_I$$

$$f_w + f_b + f_I = 1$$

solve for f_w , f_b , f_I

D. Tran and D. Fleischmann

three materials



- Bone mass from single energy CT is inaccurate if fat fraction is unknown

Quantitative Computed Tomography Scanning for Measurement of Bone and Bone Marrow Fat Content

A Comparison of Single- and Dual-Energy Techniques Using a Solid Synthetic Phantom

MITCHELL M. GOODSITT, PhD,* AND DANIEL I. ROSENTHAL, MD

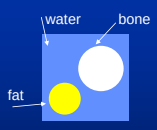
Goodsitt MM, Rosenthal DI. Quantitative Computed Tomography scanning for measurement of bone and bone marrow fat content: a comparison of single- and dual-energy techniques using a solid synthetic phantom. Invest Radiol 1987;22:799–810.

Quantitative CT (QCT) has become a popular method for estimat-

THERE HAS BEEN continuing interest in the capability of quantitative CT (QCT) to determine the amount of skeletal mineral. Most clinical attention has focused on single-energy techniqes, because of greater ease and precision.[1,2] The accuracy of single-energy CT methods has been

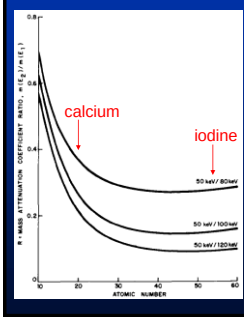
Goodsitt et al: Inv Radiol 22, 799, 1987.

three materials



- Bone mass from single energy CT is inaccurate if fat fraction is unknown
- DE can be more accurate but is less precise

Three known materials dual energy CT



Reliable separation requires large

$$(R_1 - R_2)^2$$

where $R = \mu_{high}/\mu_{low}$, depends on material and energies

Works best for one high Z and one lower Z material, and very different x-ray energies

Kelcz et al. Med Phys 6, 418-25, 1979.

Choice of photon energies

- ΔE, E_L, E_H
- very critical for SNR efficiency, separation robustness, etc.
- implementations
  - different kVp and/or filtration
  - layered detector
  - photon counting with energy analysis

Photon counting detectors

- main challenges:
  count rate capability, translates to scan speed
  imperfect and count-rate dependent energy response
- very promising in the long term but not yet ready for clinical use

Spectral separation

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

better spectral separation and dose efficiency

* assumes good energy response

Dual energy implementations

- Sequential scans at different kVp
  - motion sensitivity > 50% 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

contrast material sensitivity - very low concentration CNR

- optimized single energy CT
  - advantage: highest CNR
  - limitation: inhomogeneous background
- temporal subtraction (post - pre)
  - advantage: perfect background suppression (w/o motion)
  - limitation: motion misregistration
    lower CNR at the same dose
- rapid dual energy CT
  - advantage: motion immunity
  - limitations: nonuniform Z background?
    lowest CNR at the same total dose

Summary

- more material specificity than single energy CT(e.g., average material properties, material cancelled images)
- perfect beam hardening correction (pre-recon)
  - effective monoenergetic images, more accurate RTP and PET attenuation correction
- significant challenges but also many opportunities

Thank You

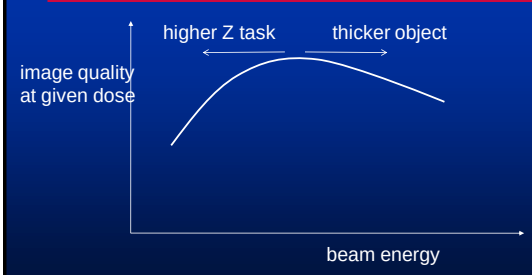
Summary

- virtual pre-contrast 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, especially for small amounts of either
- lower dose?
  - not likely, compared to optimized protocols
- molecular imaging?
  - I don't think so

Dose

- Two scans. Do we have to double the dose?
  - Depends on the goal
  - Start by splitting the dose to both energies

Dose



both spectra of DE system can't be optimal!

Dose

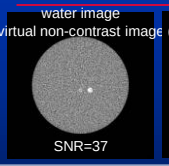
- Two scans. Do we have to double the dose?
  - Depends on the goal
  - Start by splitting the dose to both energies
- DE not likely to provide the same noise performance as optimized single energy protocols
- penalty, if any, is low

Summary of commercial systems

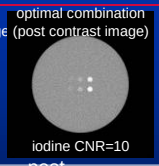
- Siemens: two sources, different kVp and filtration, image-based processing
- GE: single source with rapid switching, same filter for both kVps, projection-based processing
- Philips: sequential scans at different kVp, same filtration, image-based processing
- Lots of R&D work

Dose comparison

water image (virtual non-contrast image)



optimal combination (post contrast image)



DUAL ENERGY
55/80 keV acquisition
ideal dose allocation (~1:1)

total dose = D

pre-contrast dose ~ 0.15D

post-contrast dose ~ 0.79D

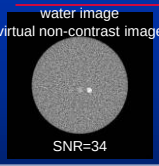
CONVENTIONAL
55 keV pre/post contrast scans

total dose 0.94 D

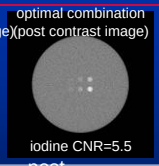
Under ideal conditions, DE scan has slightly higher dose

Dose comparison

water image (virtual non-contrast image)



optimal combination (post contrast image)



DUAL ENERGY
55/80 keV acquisition
poor dose allocation (~1:8)

total dose = D

pre-contrast dose ~ 0.13D

post-contrast dose ~ 0.23D

CONVENTIONAL
55 keV pre/post contrast scans

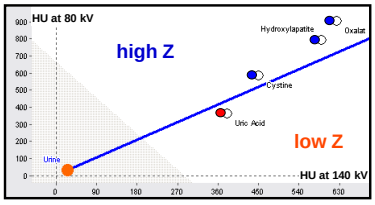
total dose 0.36 D

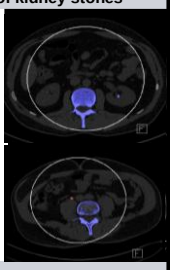
Under non-ideal conditions, DE scan has much higher dose

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

Principle of Dual Energy CT – Image Based Evaluation

Each material is characterized by its „Dual Energy Index“ $\frac{x_{80} - x_{140}}{x_{80} + x_{140} + 2000}$

x_{80} and x_{140} are the Hounsfield numbers at 80 kV and 140 kV, resp.

Material	DEI
Bone	0.1148
Liver	0.0011
Lung	-0.0021
Soft Tissue	-0.0052
Skin	-0.0064
Proteins	-0.0087
Fat	-0.0194
Gall fluid	-0.0200

based on subtraction, noisy

Dual energy CT can measure chemical composition!

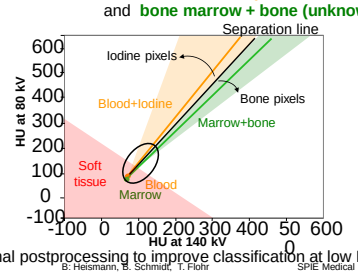
SIEMENS

Courtesy of B. Krauss, B. Schmidt, and Th. Flohr, Siemens Medical Solutions

Image Based Methods

SIEMENS

- **Modified 2-material decomposition: Separation of two materials**
  → Assume mixture of **blood + Iodine (unknown density)**
  and **bone marrow + bone (unknown density)**



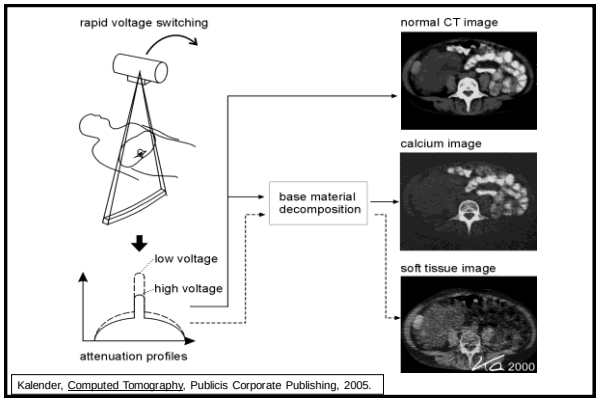
- Additional postprocessing to improve classification at low HU numbers

Image Based Methods

SIEMENS

- **Modified 2-material decomposition: Separation of bone and Iodine**
- **Automatic bone removal without user interaction**
  → Clinical benefits in complicated anatomical situations:
  - → Base of the skull
  - → Carotid arteries
  - → Vertebral arteries
  - → Peripheral runoffs

Courtesy of Prof. Pasovic, University Hospital of Krakow, Poland



Kalender, Computed Tomography, Publicis Corporate Publishing, 2005.

noise in processed images

low energy

high energy

material 1

high, correlated noise

material 2

usually high noise

can be low noise

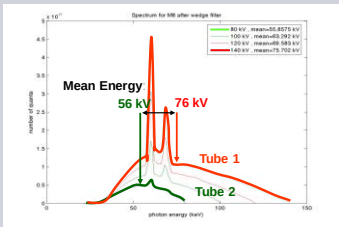
material cancelled images

equivalent monoenergetic images

Principle of Dual Energy CT

SIEMENS

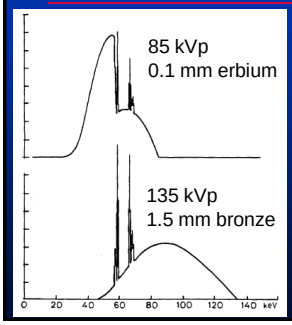
Data acquisition with different X-ray spectra: **80 kV** / **140 kV**



Different mean energies of the X-ray quanta

Courtesy of B. Krauss, B. Schmidt, and Th. Flohr, Siemens Medical Solutions

Dual kVp, dual filtration



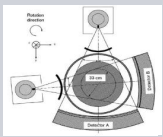
- switched filtration improves separation
- different mA helps apportion dose

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

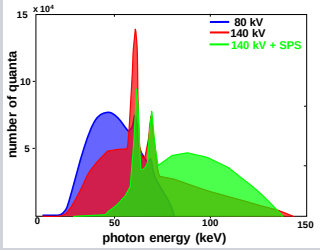
syngo Dual Energy
- Principle of Dual Energy

SIEMENS

SOMATOM Definition Flash

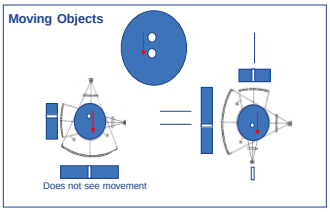


S_1: 80 kV S_2: 140 kV + SPS

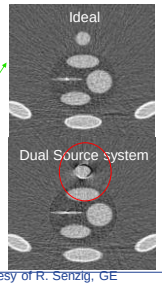


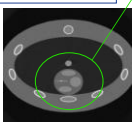
SPS= Selective Photon Shield

Dual Source Challenge: Inconsistent scans



Moving Phantom Simulation





imagination at work

Courtesy of R. Senzig, GE Healthcare

Dual energy implementations

- Sequential scans at different kVp
  motion sensitivity > 50% T_{rot}
- Two sources at ~90º on the same gantry
  some motion sensitivity (~ 25% T_{rot})
- Switching kVp within a single scan[1, 2]

1. Lehmann et al: Med Phys 8, 659-67, 1981.
2. Kalender et al, Med Phys 13, 334, 1986.

Rapid kVp switching Dual energy CT

Requires fast generator and detectors

Dose allocation controlled by dwell time

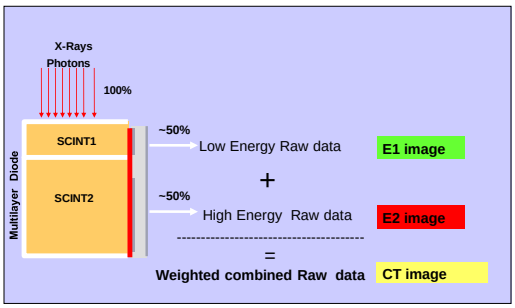
Difficult to switch filters

imagination at work

Courtesy of Uri Shreter, GE Healthcare

PHILIPS

Multilayer Detector



Investigational device

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Layered detector

- simultaneous dual energy sensing
- relatively poor spectral separation

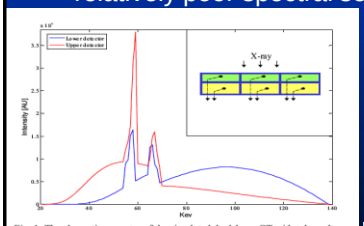


Fig. 1. The absorption spectra of the simulated dual-layer CT with tube voltage of 140 Kv. Insert shows detector configuration with partial absorber (green) and full absorber (yellow).

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

Dose efficiency

3 component decomposition, each technique optimized

TABLE IV. Relative entrance exposure requirements.

	kVp	Filtration	Relative exposure	Relative std dev
	70	3.0 mm Al	= 33	
Two kVp	180	3.0 mm Al, 0.5 mm Cu, 0.5 mm Sn	= 18	24.1
		Total Exposure	= 51	
Two crystal	180	3.0 mm Al	= 100	36.2

4X overall difference in dose efficiency

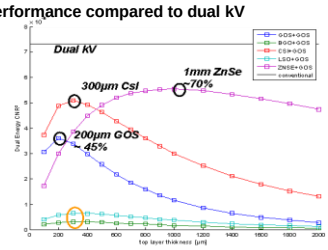
Kelcz et al: Med Phys 6, 418-25, 1979.

Dual Layer Detectors - Challenges

→ Complex technical realization
→ **Reduced dual energy performance compared to dual kV**

→ Performance depends on scintillator types and thicknesses

→ Dual energy performance less than 70% compared to dual kV technology (same radiation dose!)



Courtesy S. Kappler, Siemens

Page 84

B: Heismann, B. Schmidt, T. Flohr

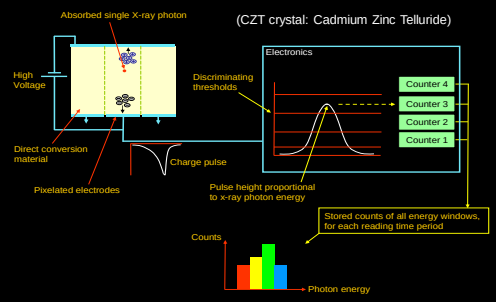
Short course 1b/1

SPIE Medical Imaging Conference 2010

PHILIPS

Photon Counting Spectral CT – Detector Principle

(CZT crystal: Cadmium Zinc Telluride)



Direct- Conversion Detector efficiently translates X-ray photons into large electronic signals

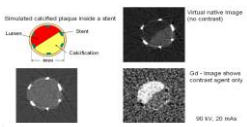
These signals are binned according to their corresponding X-ray energies

PHILIPS

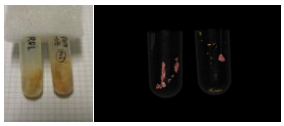
Photon Counting Prototype*

Early results

- Two systems in Philips Research labs
- Research system installed at Washington University-Saint Louis in Nov. 2008



'K-Edge Imaging' w/ Univ Ulm (*Radiology* Oct 08)



Novel contrast agents targeting fibrin (clots) w/ Wash U

*Works-in-Progress: Pending commercial availability and regulatory clearance

86

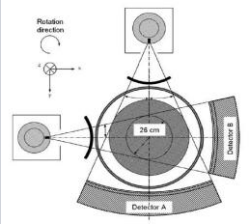
Dual energy implementations

- Sequential scans at different kVp
  motion sensitivity > 50% T_{rot}. Helical?
- Two sources at ~90º on the same gantry

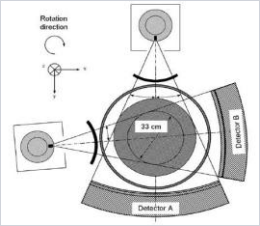
syngo Dual Energy
- Principle of Dual Energy

SIEMENS

SOMATOM Definition



SOMATOM Definition Flash



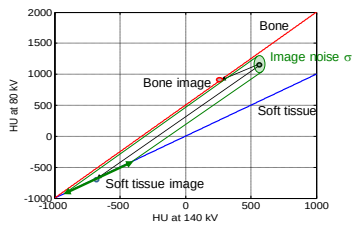
Two X-ray tubes, one at 80 kVp, second at 140

Image Based Methods

SIEMENS

- **Calculate material specific images: 2 materials of unknown density**

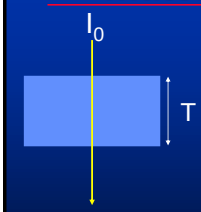
→ Noise amplification in the material specific images!



B: Heismann, B. Schmidt, T. Flohr

Short course 987
SPIE Medical Imaging Conference 2010

attenuation and material identification



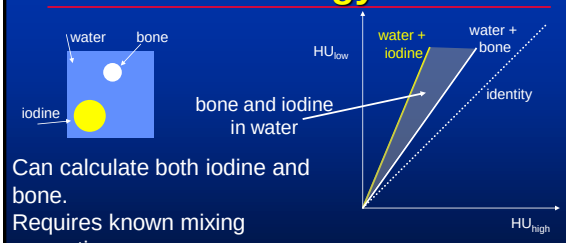
But what material is it?

Even more confusing if T is unknown

$I = I_0 e^{-\mu T}$

$\mu T = \ln(I_0 / I)$

Three known materials dual energy CT



Can calculate both iodine and bone.
Requires known mixing properties
and consistent water density

$m_b = A\{ HU_{low} - (M_{1,l} / M_{2,l}) HU_{high}\}$

Kelcz et al. Med Phys 6, 418-25, 1979.