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CT Clinical Innovation Center

Dual-energy CT Protocols

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Outline

- Basic principle of dual-energy CT
- Major components of dual-energy CT protocols
- Major considerations in dual-energy CT protocols
  - Radiation dose
  - Scan techniques and spectrum selection
  - Material-generic applications (virtual monochromatic)
  - Material-specific applications (Virtual unenhanced, iodine, stone, etc.)
- Summary

2

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Basic principle of dual-energy CT

- In diagnostic energy range, linear attenuation coefficient of a material can be decomposed into (Alvarez and Macovski, 1976):

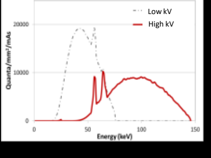
$$\mu(r,E) = \boxed{a_1(r)\cdot\frac{1}{E^3}} + \boxed{a_2(r)\cdot f_{KN}(E)}$$

Photon-electric effect — Compton effect

- Equivalent to two basis-material decomposition (Lehman et al, 1981):

$$\mu(r,E) = \rho_1(r)\cdot\left(\frac{\mu}{\rho}\right)_1(E) + \rho_2(r)\cdot\left(\frac{\mu}{\rho}\right)_2(E)$$

- To solve the density maps of the two basis materials, measurements from at least two-energy spectra are required:



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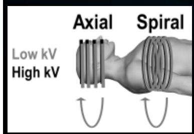
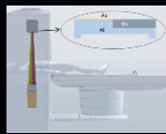
Dual-energy CT Techniques

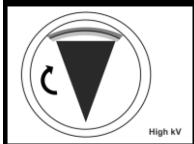


Slow kV switching

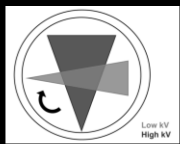
Fast kV switching



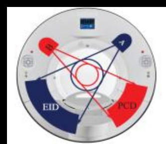
Twin-beam



Dual-layer detector



Dual x-ray source



Photon-counting

McCollough et al, Radiology, 2015
Saba et al, 2015
Kappler et al, 2010; Yu Z et al, PMB, 2016

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Strength and weakness of different Approaches

Technical Approaches for Acquiring Dual-Energy Data Set

Technology	Strengths	Weaknesses
Temporally sequential scanning of the entire scan volume	Can be performed on any CT scanner (no special hardware required).	Any patient motion occurring between the two scans may cause severe degradation of the resultant images and material composition information.
Temporally sequential scanning of a single axial rotation	Can be performed on any CT scanner (no special hardware required). Reduced interscan delay between the low- and high-energy images.	Best temporal resolution obtained with partial-scan reconstructions, which are more susceptible to errors in CT numbers; this might degrade material composition accuracy. The susceptibility to motion misregistration between the low- and high-energy single axial rotation may limit the value of this approach for imaging of vascular processes or tissues and organs susceptible to motion. For scanners having relatively narrow z-axis coverage, the increase in total scan time associated with an axial acquisition mode may be a limiting factor.
Rapid switching of the x-ray tube potential	Near-simultaneous data acquisition of the low- and high-energy data set. Allows dual-energy material-decomposition algorithms to be implemented by using either projection data or reconstructed images. Reduces beam-hardening artifacts in calculated "virtual monoenergetic" images.	Requires specialized hardware. Relatively high overlap of the energy spectra.
Multilayer detector	Simultaneous data acquisition of the low- and high-energy data set. All image data are acquired in a manner that supports material-specific imaging.	Requires specialized hardware. Relatively high overlap of the energy spectra. Noise level may differ between low- and high-energy images.
Dual x-ray sources	Tube current and tube filtration can be optimized for each tube potential independently. Relatively low degree of spectral overlap, which improves contrast-to-noise ratios in material-specific images. Beam-hardening corrections are applied prior to image reconstruction, allowing material-specific images to be created in the image domain.	Requires specialized hardware. A 90° phase shift between low- and high-energy data. Simultaneous use of both x-ray sources allows scattered radiation whose original primary photon came from one tube to be detected by the detector of the other tube, requiring specialized scatter correction.
Photon-counting detectors	Uses energy-specific measurements and energy thresholds to reject electronic noise. Facilitates new imaging approaches, such as k-edge subtraction.	Requires specialized hardware, which is not anticipated to be commercially available for some time, if at all.

McCollough et al, Radiology, 2015

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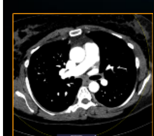
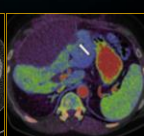
Many Important Clinical Applications

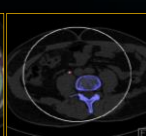


Virtual Monochromatic

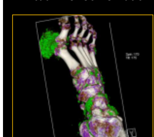
Virtual non-contrast



Iodine imaging



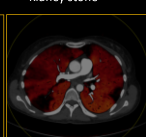
Kidney stone



Gout

Breast implant

Auto bone removal



Lung PBV

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Major components of dual-energy CT protocols

- Scanning and reconstruction (scanner platform dependent)
  - Radiation dose
  - Spectra and technique selection
  - Material decomposition
- Post-processing (clinical application dependent)
  - Material-generic (virtual monochromatic)
  - Material-specific (Virtual unenhanced, iodine, stone, gout, etc.)

7

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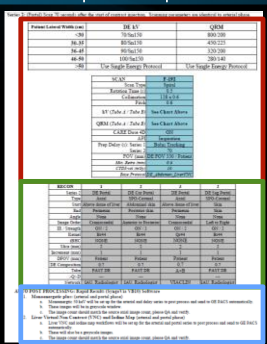
An example DE protocol



Scanning parameters: rotation time, collimation, pitch, AEC, dose, etc, also including a DE spectrum selection chart

Reconstruction parameters: Series for DE post-processing needs to be quantitative kernel

DE post-processing: mono, virtual non-contrast, iodine, etc.

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Major Considerations in Dual-energy CT Protocols

- Radiation dose
- Scan techniques and spectrum selection
- Material-generic applications (virtual monochromatic)
- Material-specific applications (Virtual non-contrast, iodine, stone, etc.)

9

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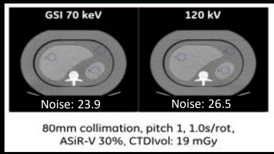
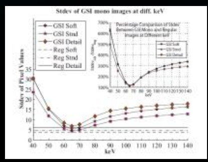
Radiation dose neutral?

GE GSI

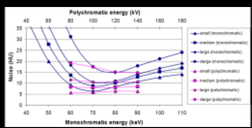


GE Revolution CT GSI Xtream white paper, 2017



Zhang D et al, Med Phys, 2011

Siemens DS



Yu L et al, Med Phys, 2009, 2011

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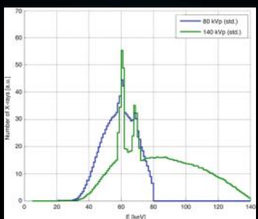
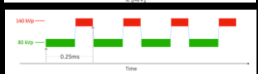
Spectrum and technique selection – GE GSI





- Fixed at 80/140 kVp
- Longer time projections for 80 kVp to balance the radiation output between the 2 energies
- GSI Assist to select the scan technique from a preset list

Saba et al, 2015
GE Revolution CT GSI Xtream white paper, 2017

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Spectrum and technique selection – GE GSI

GSI preset list: GSI assist selects a technique to match CTDIvol in a non-GSI acquisition for a target non-GSI noise index

GSI Preset	Area	SFOV	Rotation Time (s)	Beam collimation (mm)	$CTDI_{w}$**
1	Body	Large Body	0.5	40	17.30
2	Body	Medium Body	0.9	20	31.81
3	Body	Medium Body	0.5	40	18.08
4	Body	Large Body	0.9	40	28.04
5	Body	Large Body	1.0	40	31.55
6	Body	Medium Body	1.0	40	32.83
7	Body	Large Body	1.0	20	34.46
8	Body	Medium Body	1.0	20	35.82
9	Head	Head	0.9	20	69.23
10	Body	Large Body	0.8	40	24.77
11	Body	Medium Body	0.8	40	25.87
12	Body	Large Body	0.8	20	26.99
13	Body	Medium Body	0.8	20	28.16
14	Head	Head	0.8	20	61.22
15	Body	Large Body	0.6	40	21.20
16	Body	Medium Body	0.6	40	22.08
17	Body	Large Body	0.6	20	23.08
18	Body	Medium Body	0.6	20	24.09
19	Head	Head	0.6	20	52.16
20	Head	Head	0.5	20	47.80
21	Head	Small Head	0.5	20	37.81
*22	Body	Large Body	0.7	40	14.16
*23	Body	Medium Body	0.7	40	15.37
*24	Body	Large Body	0.7	20	15.65
*25	Body	Medium Body	0.7	20	16.49
*26	Head	Head	0.7	20	34.87
27	Body	Large Body	0.8	40	22.87
28	Body	Medium Body	0.8	40	24.23
29	Body	Medium Body	0.8	20	26.31
30	Head	Head	0.8	20	55.10
*31	Body	Large Body	0.6	40	12.52
*32	Body	Medium Body	0.8	40	13.42
*33	Body	Large Body	0.8	20	13.49
*34	Body	Medium Body	0.6	20	13.95
35	Body	Medium Body	0.5	20	19.60
*36	Body	Large Body	0.8	40	9.93

GSI Preset	Area	SFOV	Rotation Time (s)	Beam collimation (mm)	$CTDI_{w}$**
*37	Body	Medium Body	0.8	40	10.61
*38	Body	Medium Body	0.8	20	11.47
*39	Head	Head	0.8	20	24.93
*40	Body	Large Body	0.6	40	11.82
*41	Body	Medium Body	0.6	40	12.56
*42	Body	Medium Body	0.6	20	13.76
*43	Head	Head	0.6	20	29.17
*44	Body	Large Body	0.7	40	10.37
*45	Body	Medium Body	0.7	40	10.94
*46	Body	Medium Body	0.7	20	11.81
*47	Head	Head	0.7	20	25.58
*48	Body	Large Body	0.7	40	8.74
*49	Body	Medium Body	0.7	40	9.37
*50	Body	Medium Body	0.7	20	10.03
*51	Body	Large Body	0.5	40	9.96
*52	Body	Medium Body	0.5	40	10.34
*53	Body	Medium Body	0.5	20	11.29
*54	Body	Large Body	0.6	40	8.82
*55	Body	Medium Body	0.6	40	9.38
*56	Body	Medium Body	0.6	20	10.18
57	Cardiac	Large Cardiac	0.35	40	8.68
58	Cardiac	Medium Cardiac	0.35	40	8.71
59	Cardiac	Small Cardiac	0.35	40	7.40
60	Cardiac	Large Cardiac	0.35	40	14.35
61	Cardiac	Medium Cardiac	0.35	40	14.93
62	Cardiac	Small Cardiac	0.35	40	12.68
63	Cardiac	Large Cardiac	0.35	40	17.36
64	Cardiac	Medium Cardiac	0.35	40	17.86
65	Cardiac	Small Cardiac	0.35	40	15.27
*66	Head	Small Head	0.8	20	21.99

12

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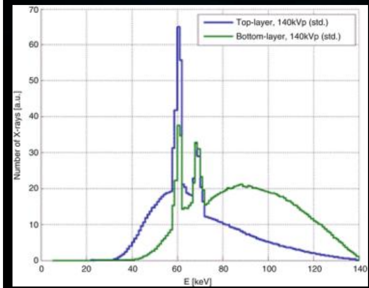
Spectrum and technique selection – Philips iQon



- Scan technique just like single energy
- 120 kV or 140 kV

Saba et al, 2015

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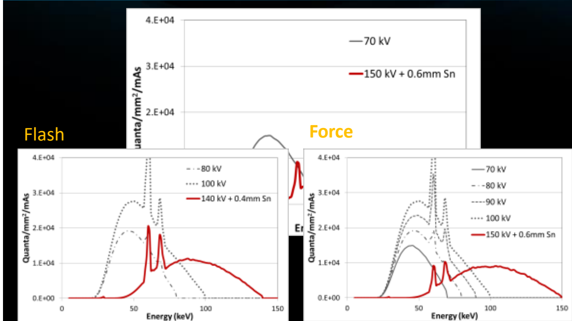
Spectra and technique selection – Siemens DS



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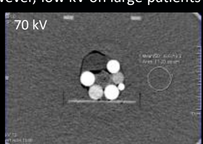
Spectra and kV pair selection – Siemens Dual-source

- Better spectra separation → better dual-energy performance (Primak A et al, Med Phys, 2009)
- However, low kV on large patients → limited radiation output, causing artifacts.



Michalak G et al, Med Phys 2016

Patient Lateral Width	DE kV	Quality reference mAs
<30 cm	70/Sn150	800/200
30-35 cm	80/Sn150	450/225
36-45 cm	90/Sn150	320/200
46-50 cm	100/Sn150	280/140
>50 cm	Use SE	Use SE

Mayo dual-energy liver kV chart

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Two Categories of Dual-energy CT Applications

- Material-generic imaging
  - Virtual monochromatic image
- Material-specific imaging
  - Basis material decomposition
  - PE-Compton decomposition

⇩

Reduce artifacts, improve quantitative accuracy, Improve contrast and CNR

⇩

Expand CT clinical applications
- Material quantification (e.g., iodine, bone)
- Material classification (e.g., bone/iodine, uric acid/non-uric acid)

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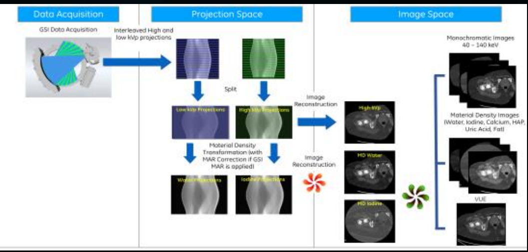
GE – Basis material decomposition in projection domain



- Projection domain basis material decomposition
- Material density maps (Iodine, water, calcium, HAP, uric acid, and fat) and VUE
- Generated as material basis pair (e.g., water/iodine; uric acid/calcium)

GE Revolution CT GSI Xtream white paper, 2017

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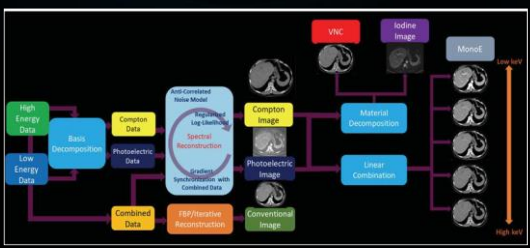
Philips – PE/Compton decomposition in projection domain



- Projection domain decomposition → PE and Compton data → PE and Compton images
- Material decomposition → VNC, Iodine image, etc.
- Anti-correlation model to reduce noise

Rassouli et al, 2017

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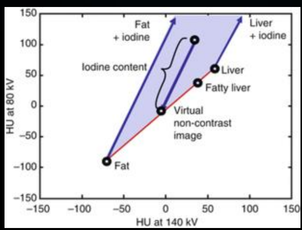
Siemens – 2 or 3-material decomposition in image domain



- Material groups:
  - Liver virtual non-contrast (VNC): soft tissue/iodine/fat
  - Lung PBV: soft tissue/iodine
  - Virtual non-calcium: soft tissue/calcium/fat
  - Virtual un-enhanced: water/iodine
- Classification task relies on certain threshold of dual-energy ratio
  - Kidney stone characterization
  - Gout diagnosis

Saba et al, 2015

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Virtual monochromatic imaging

- Reduce beam-hardening artifacts and improve quantitative accuracy
- Low keV to improve contrast or CNR
- Medium keV to minimize noise
- High keV to reduce metal artifacts

20

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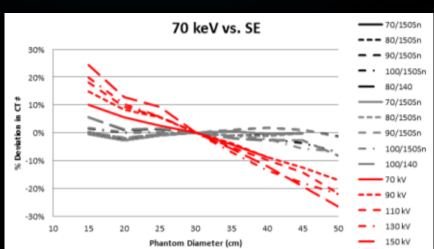
Virtual Monochromatic Imaging – CT Number Stability



Michalak et al, Med Phys 2016

21

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Improved Iodine Enhancement at low keV

120 kV | 50 keV

From Philips iQon CT, Rassouli et al, 2017

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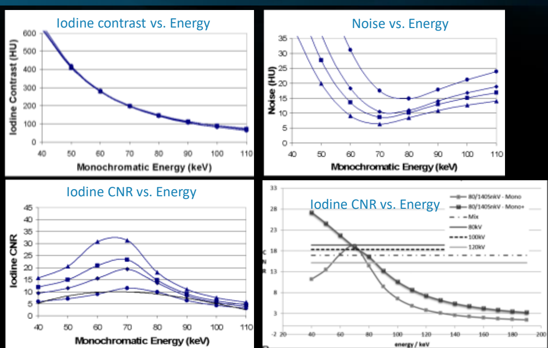
Virtual Monochromatic Imaging – Contrast, Noise, and CNR



Yu et al, Med Phys 2011 | Grant et al, Invest Rad, 2014

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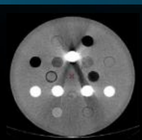
Optimal energy for virtual monochromatic images



40 keV

70 keV

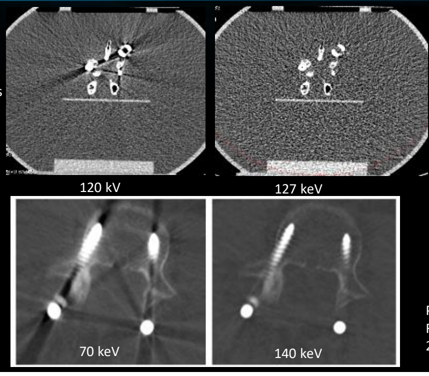
- Artifacts and noise might be problematic at too low monochromatic energies
- Optimal energy for virtual monochromatic images
  - For contrast and contrast-noise-ratio enhancement: ~50 keV
  - For minimum noise and soft tissue contrast: 65-75 keV

Goodsitt et al, Med Phys 2011

24

Metal artifact reduction

Siemens DS

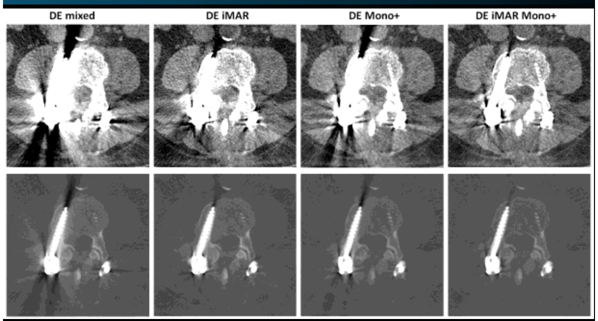


Yu et al, AJR, 2012

GE GSI

Pessis et al, RadioGraphics, 2013

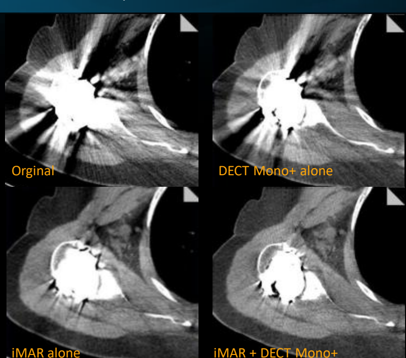
Combined with iMAR, best for spine implants



Long et al, AJR, 2018

26

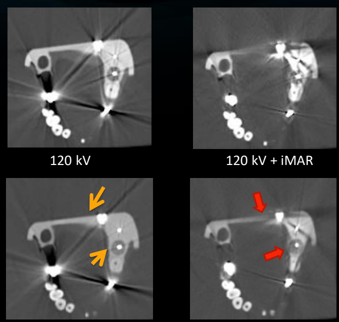
Total Shoulder Arthroplasties: iMAR+mono was the best



Glazebrook et al, RSNA 2017

27

Mono was the best, iMAR introduced artifacts in dental



120 kV | 120 kV + iMAR

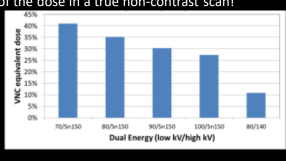
DE mono 130 keV | DE iMAR mono 130 keV

Long et al, 2018

Virtual non-contrast (VNC) – Dose consideration

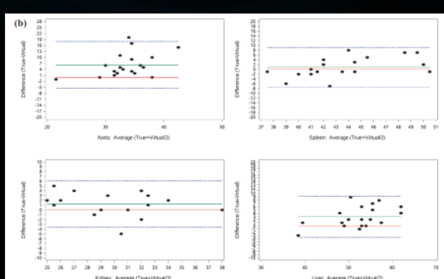
Contrast image | VNC image | True non-contrast

- Dose saving by VNC should be quantified as the dose that is required in single energy scan to generate the image quality equivalent to VNC
- Dose saving depends on dose distribution, spectrum separation and patient size: At most 40% of the dose in a true non-contrast scan!



Yu L et al, RSNA 2014

Virtual non-contrast (VNC) – CT Number Accuracy



- Vary by 5-9 HU on a per-patient basis, which is small, but could still potentially lead to mischaracterization of a mass as benign or potentially malignant (GE GSI).

Kaza et al, Acad Rad, 2017

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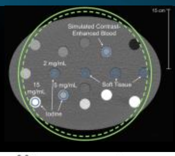
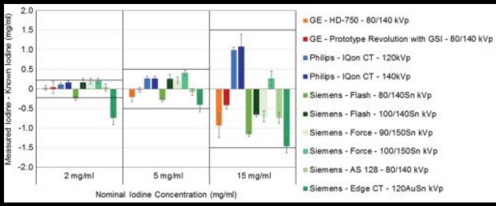
Iodine quantification accuracy



- Varies among systems, mostly within 10%
- Dual-source and fast kV switching most accurate in large phantom
- Split filter DE had the lowest accuracy



Jacobson et al, Radiology, 2018

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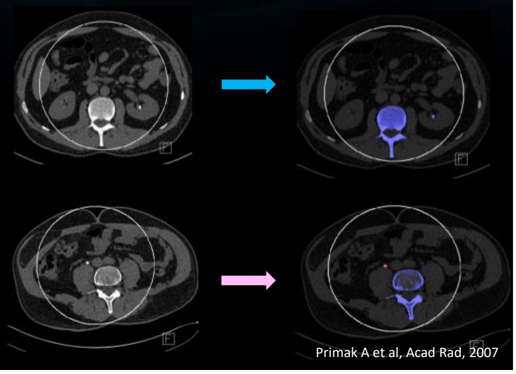
Renal Stone Composition Characterization



Primak A et al, Acad Rad, 2007

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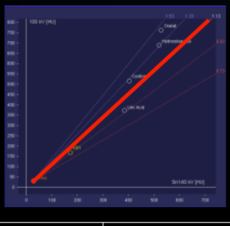
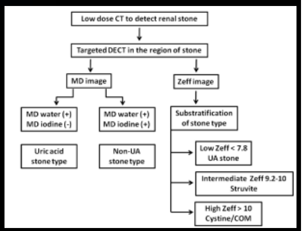
Threshold for stone classification

Siemens DE



Flash		Force	
80/140	1.15	70/Sn150	1.29
80/Sn140	1.21	80/Sn150	1.23
100/Sn140	1.13	90/Sn150	1.19
		100/Sn150	1.15

GE GSI



Kulkarni NM et al, JCAT 2013

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34

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Summary

- Major components of DE protocols
- Scanner platform-dependent considerations
  - Scanning parameter optimization (kV, dose, etc)
  - Material decomposition (methods, material types, etc)
- Clinical application-dependent considerations
  - Material-generic (virtual monochromatic)
    - Optimal keV depends on applications
  - Material-specific (VNC, iodine, stone, etc)
    - VNC dose and image quality
    - Iodine quantification accuracy
    - Stone characterization

35

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36