X-ray Computed Tomography for Medical Imaging

Jiang Hsieh, Ph.D.
and several hundred colleagues and collaborators inside and outside GE

GE Healthcare, Waukesha, Wisconsin
University of Wisconsin, Madison, Wisconsin

CT Development

- 1956 Derived mathematic for reconstruction (Harvard sabbatical)
- 1957 First lab testing (South Aferica)
- 1963 Repeated the lab experiment and published results (Tufts University)
- 1979 Shared Nobel Price in Physiology and Medicine

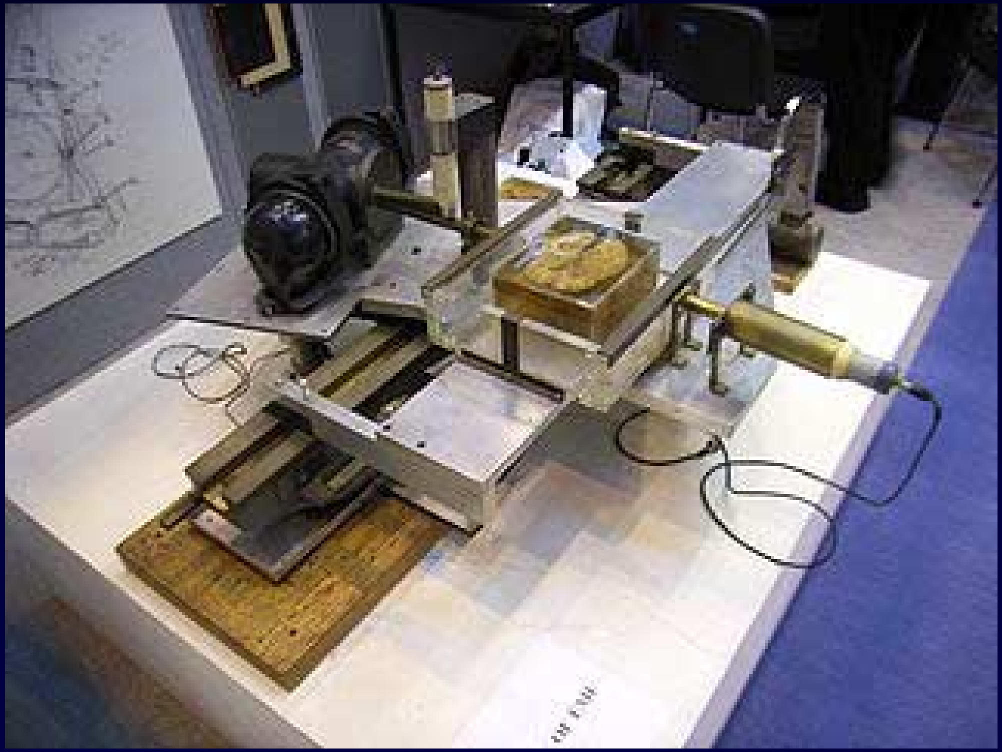
"There was virtually no response. The most interesting request for a reprint came from the Swiss Center for Avalanche Research."

Allan M. Cormack

CT Scanner Development

- The development of the first clinical CT scanner began in 1967 with Godfrey N. Housfield at the Central Research Laboratories of EMI.

Godfrey N. Hounsfield



Technological Advancements in CT

	1971	2007	Factor
Scan speed	270 sec	0.3 sec	900 X
Z-resolution	10 mm	0.5 mm	20 X
Coverage (30s)	1 cm	314 cm	314 X

1971

2007

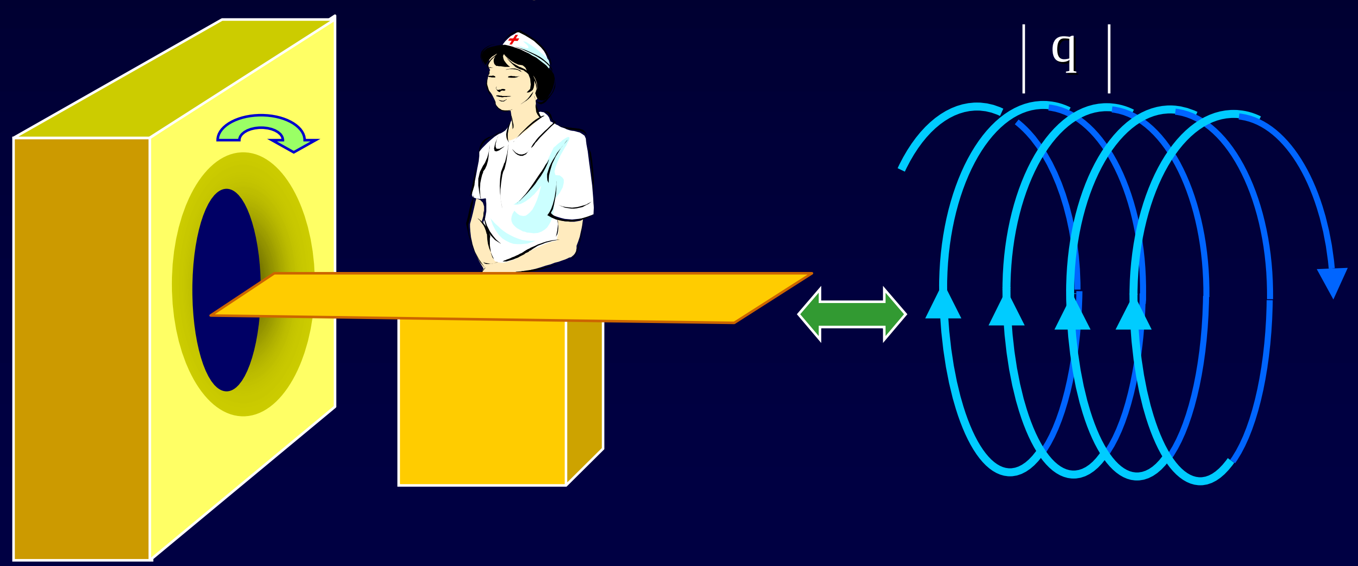
Helical Scanning

- In helical scanning, the patient is translated at a constant speed while the gantry rotates.
- Helical pitch:

$$h = \frac{q}{d}$$

q — distance gantry travel in one rotation

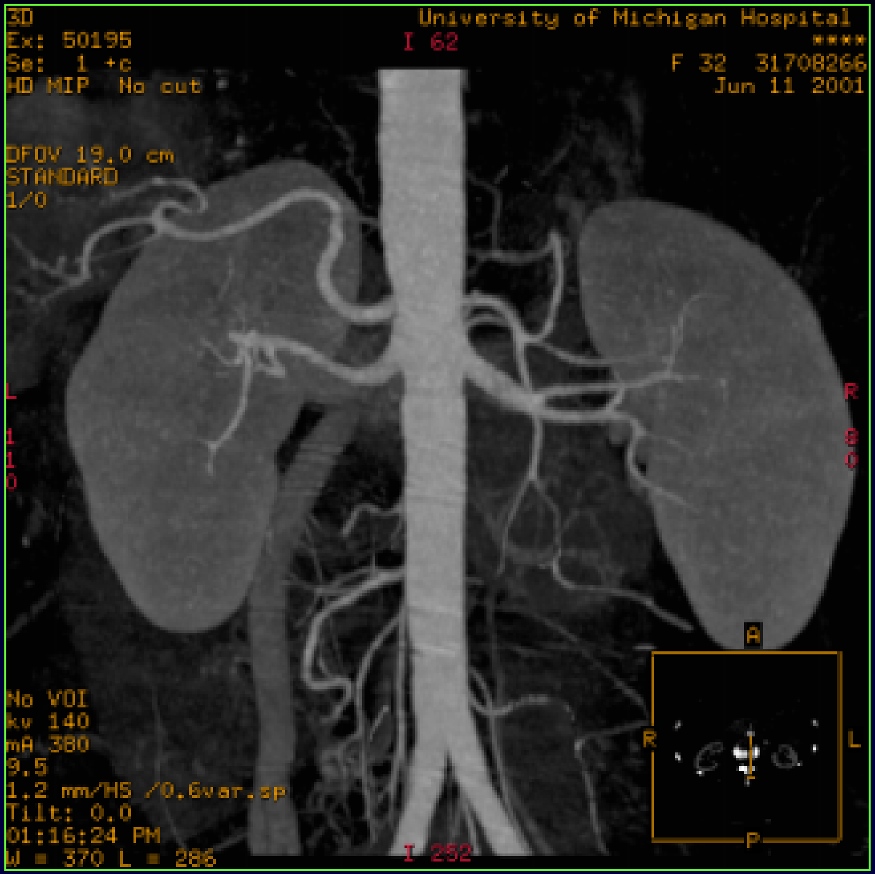
d — collimator aperture



Gantry Drive

- The key performance parameters for the gantry is the angular accuracy, stability, and speed.
- The encoder is accurate to 0.003^{o}.
- Diameter of the gantry is about 1 meter.
- Vibration needs to be a small fraction of the minimum slice thickness of image (0.625mm)

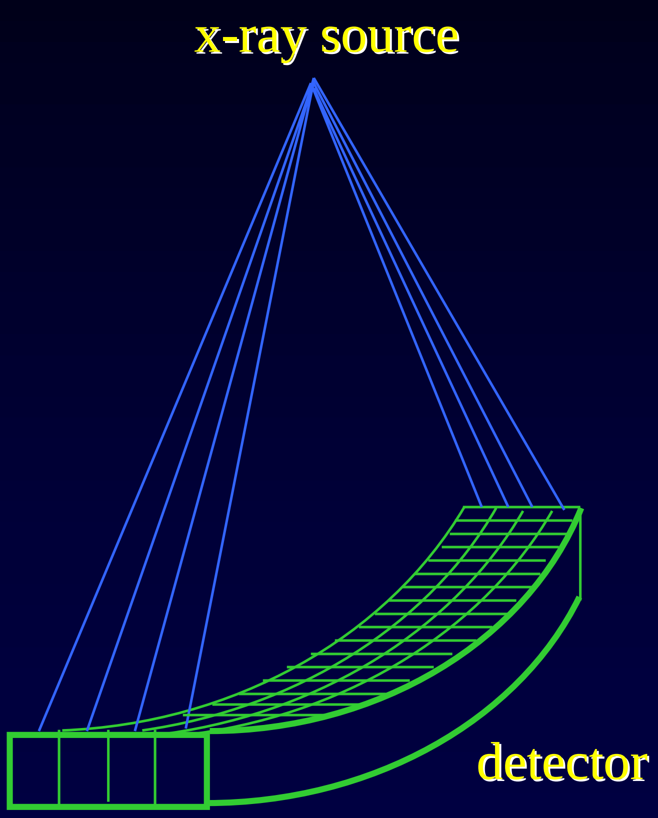
Clinical Examples



Organ Coverage in a Breath-hold

Multi-slice CT

- Multi-slice CT contains multiple detector rows.
- For each gantry rotation, multiple slices of projections are acquired.
- Similar to the single slice configuration, the scan can be taken in either the step-and-shoot mode or helical mode.
- Unlike the single slice, the slice thickness is defined by detector aperture.



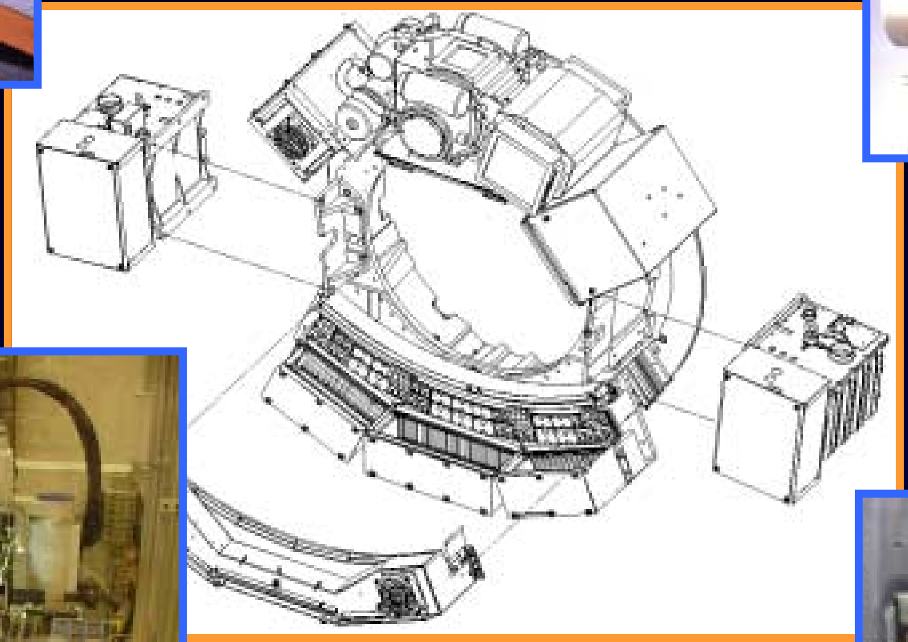
Advantages of Multi-slice

- Large coverage and faster scan speed
- Better contrast utilization
- Less patient motion artifacts
- Isotropic spatial resolution

Isotropic Volume Coverage Anytime, Anywhere

Technology Challenges

since 1990
- 64x connection
- << power
- << noise



since 1990
- 3x speed increase
- 2x slice reduction

➡ 5x tube power
- 25g force

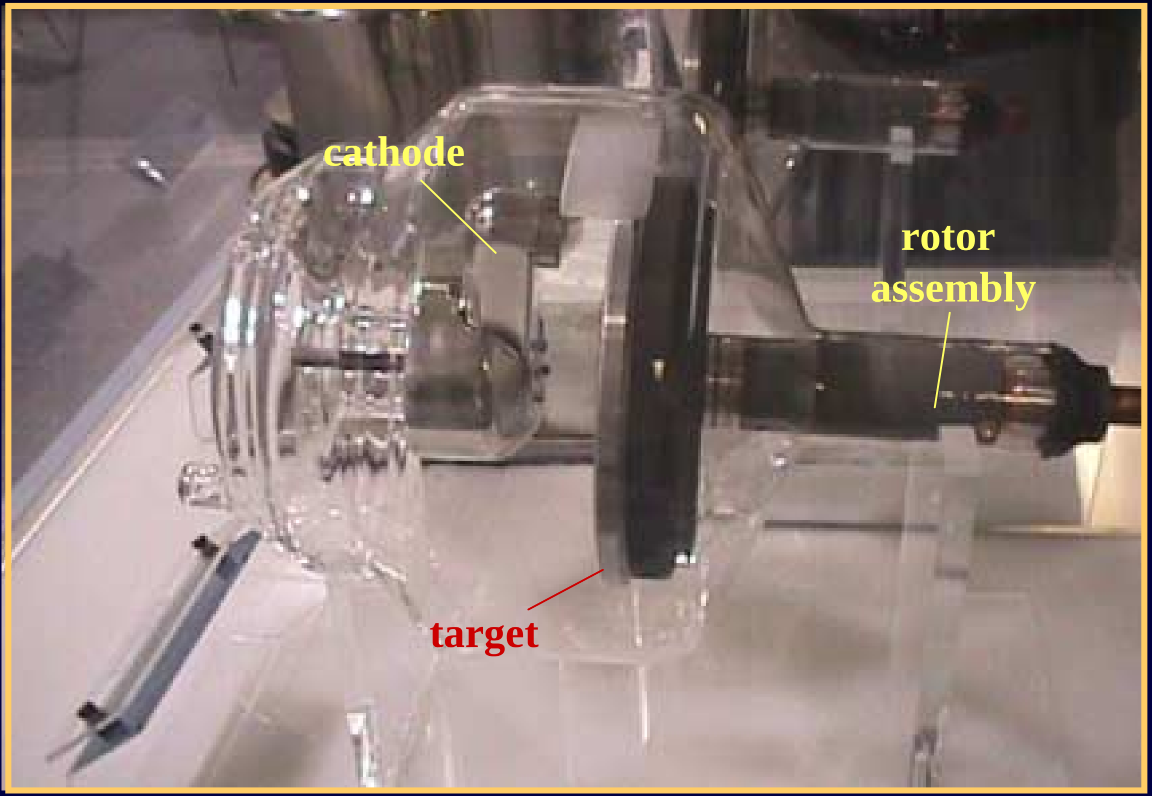
- 64000 1x1mm cells
- mm alignment

since 1990
- 3x speed increase
- 64x number slices

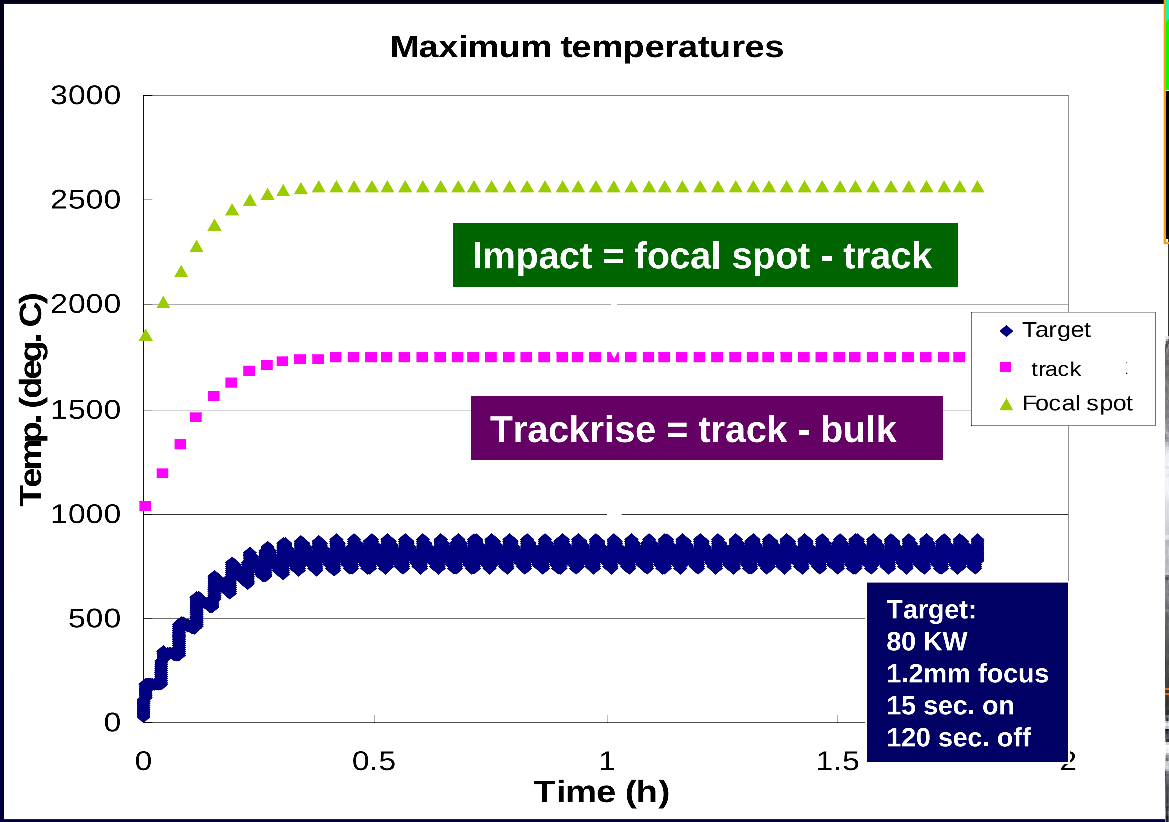
➡ 200x data rate

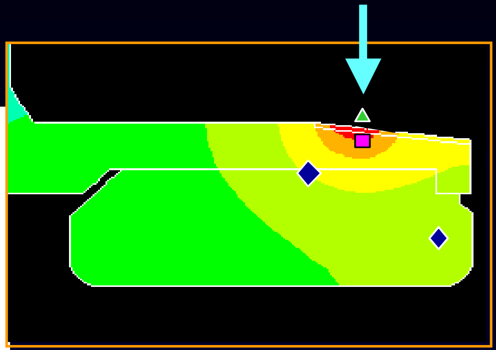
X-ray Tube

- X-ray tube is the heart of the CT system.
- One of the biggest challenges is the thermal management.



Thermal Consideration

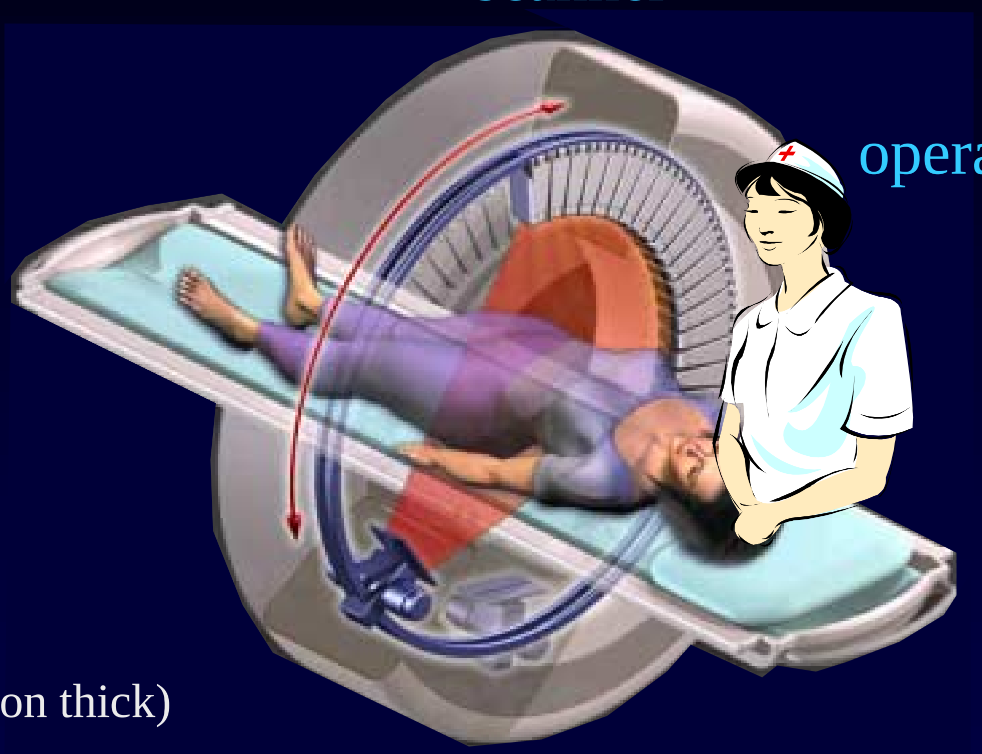




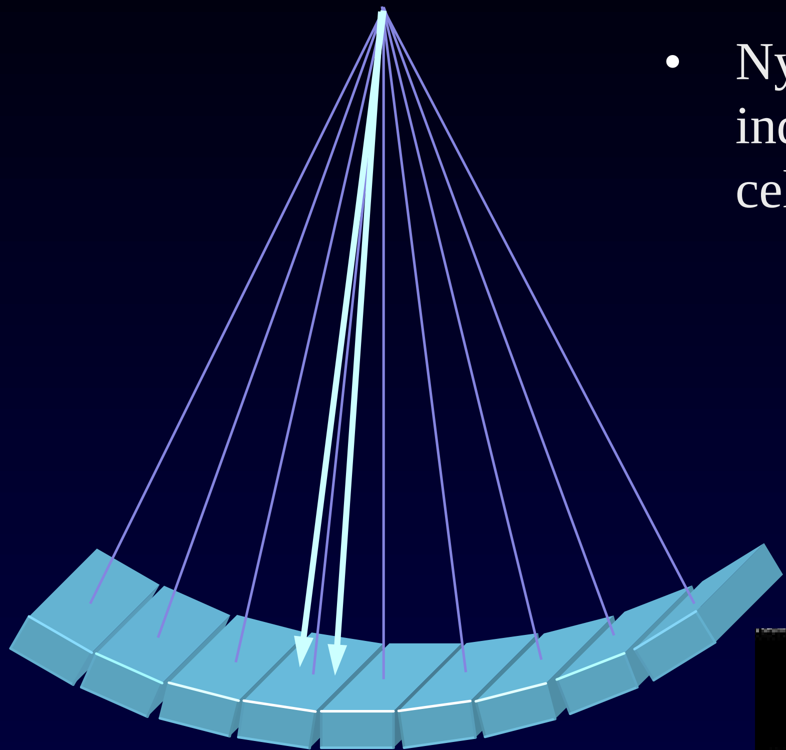
Target Thermal Gradients

Root-Causes of Artifacts

- **Nature of the X-ray Physics**
  - Beam Hardening
  - Scatter
  - Aliasing
- **New Technology**
  - Helical
  - Cone Beam
- **Patient**
  - Motion
  - Photon Starvation
- **Operator**
  - Protocols (scan thin, recon thick)
  - Partial Volume



Aliasing Artifact

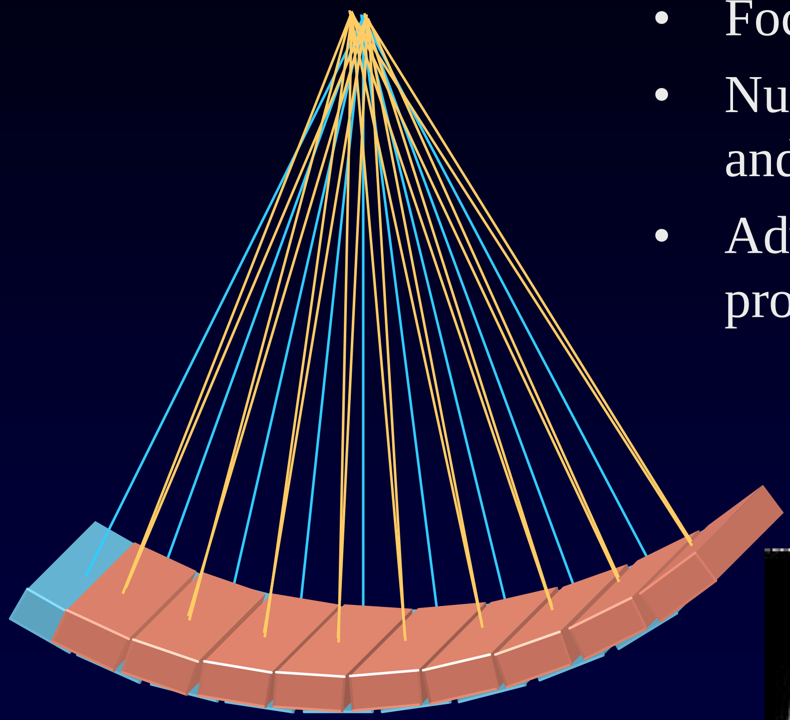


- Nyquist sampling theorem indicates that two independent samples are needed per detector cell to fully represent the projection.

Patient Scan

Animal Experiment

Dynamic Spot Control & Flying Focal Spot



- Focal spot wobble is an old technology.
- Number of views per rotation are very restrictive and are determined by the CT geometry.
- Advanced technology has been developed to provide flexibility in sampling frequency.

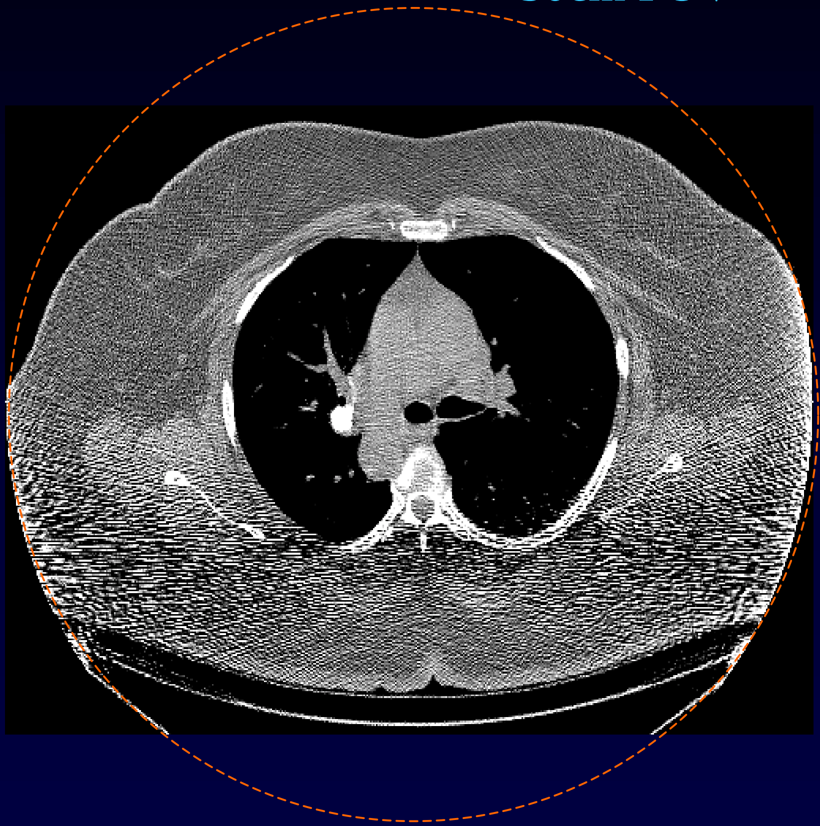
original | **dynamic control**

Photon Starvation

- Beer's law indicate that the amount of attenuation increases exponentially with path length.

$$\frac{I}{I_0} = e^{-\mu L}$$

- At low signal level, the noise in the projection is no longer dominated by the x-ray photon.
- Convolution filtering operation will further amplify the noise and streak artifacts will result.



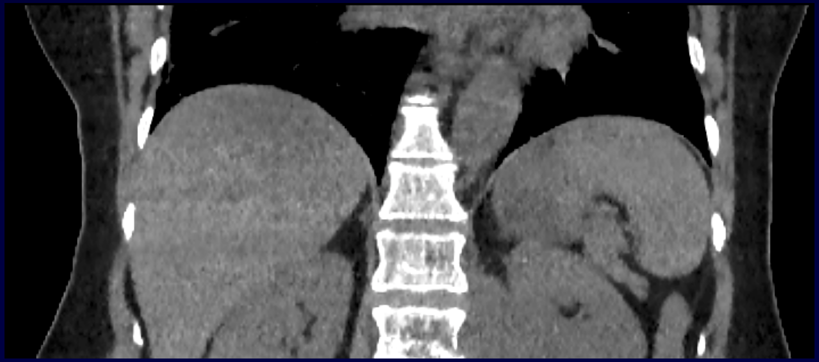
patient scan example

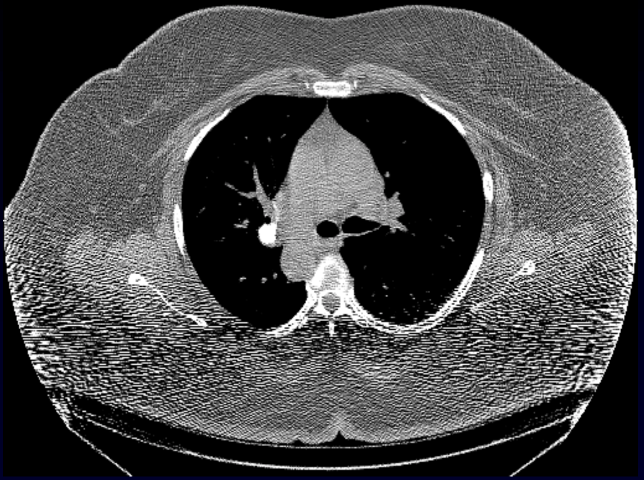
Artifact Reduction

- Algorithmic Correction
  - Adaptive filtering for streak reduction
  - Iterative reconstruction

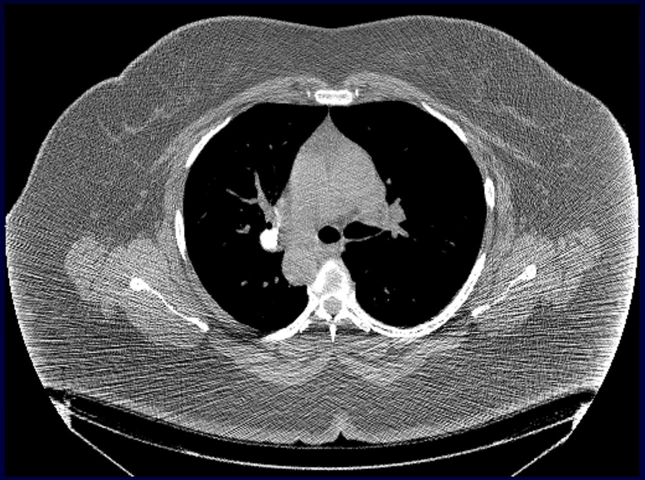
FBP

MBIR





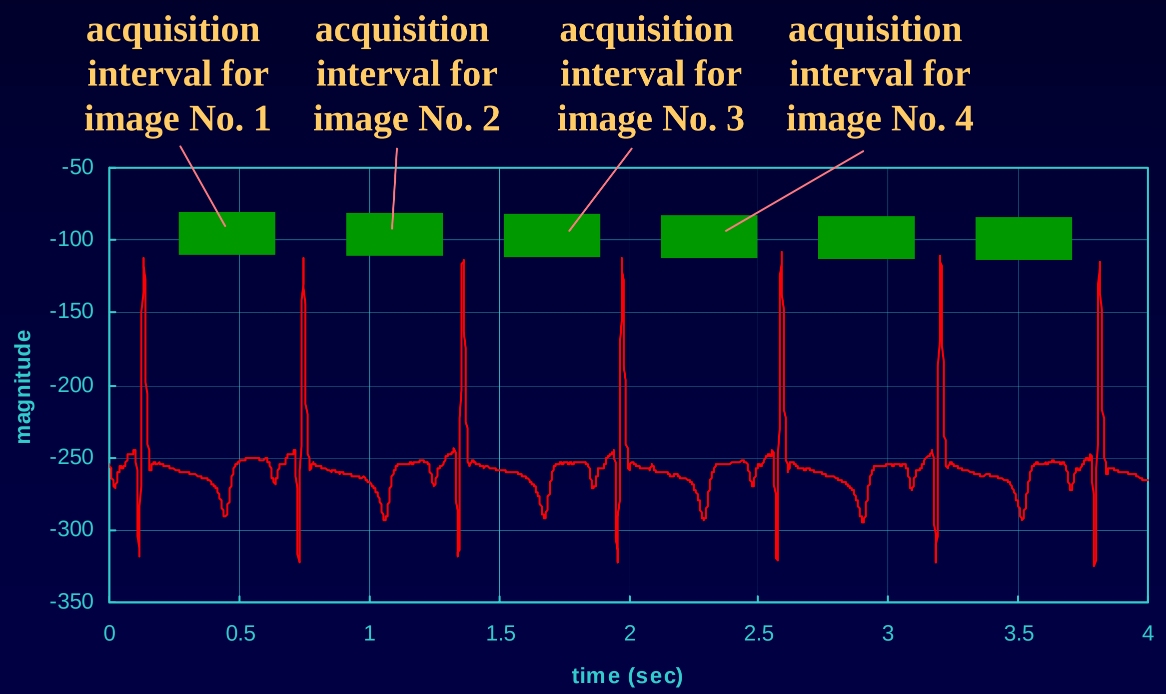
original



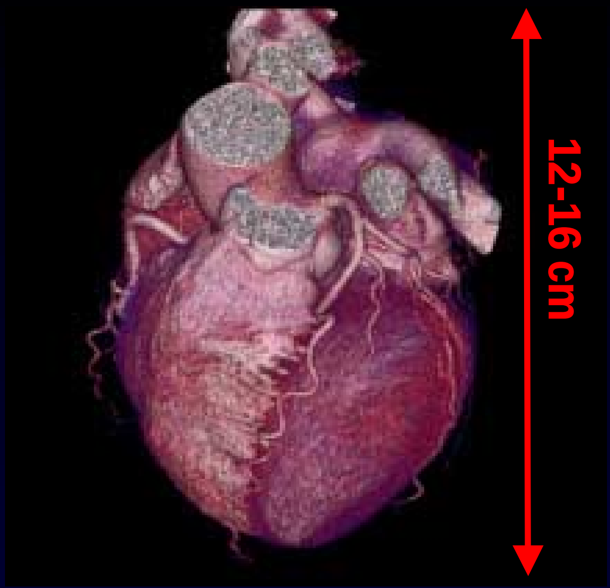
adaptively filtered

Cardiac Scans

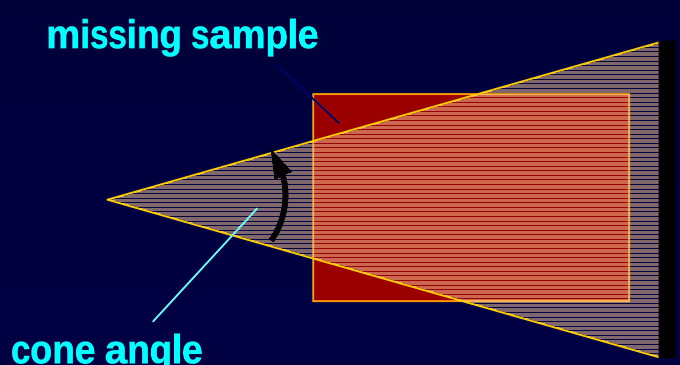
- Projection data used in the reconstruction is selected based on the EKG signal to minimize motion artifacts.

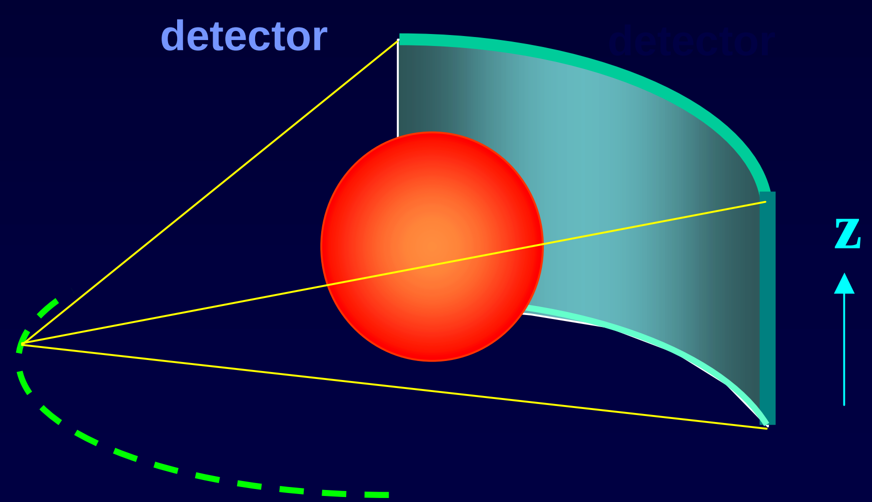


Coverage



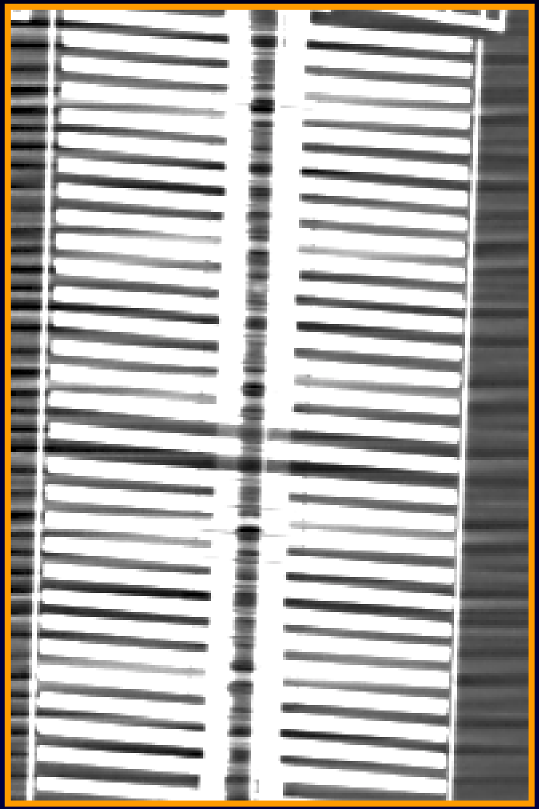
- Driven by cardiac, 4D CTA
- Pros
  - Reduce heart rate variation
  - Reduce scan time
- Cons
  - Cone beam artifact
  - Truncation





Axial Cone-beam Artifacts

coronal view

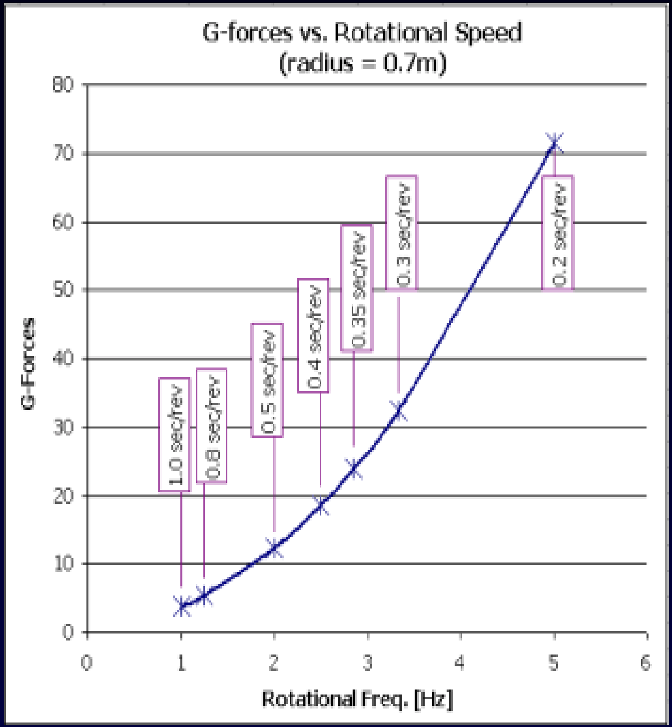


Helical Scan

Axial Scan

In-plane Temporal Resolution

0.5s gantry rotation

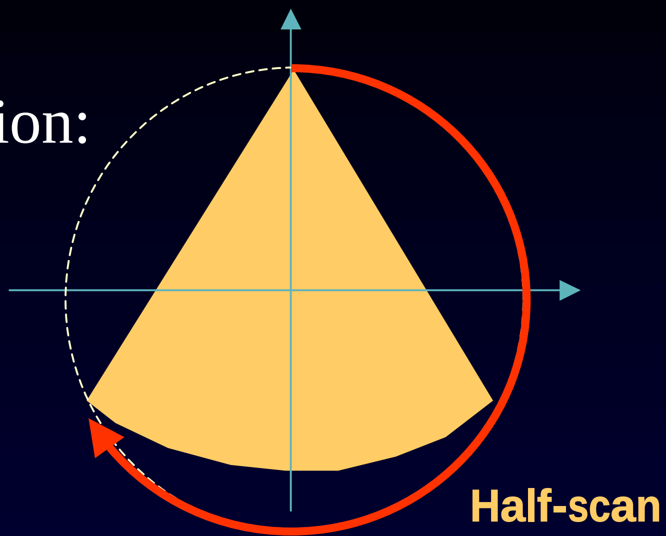


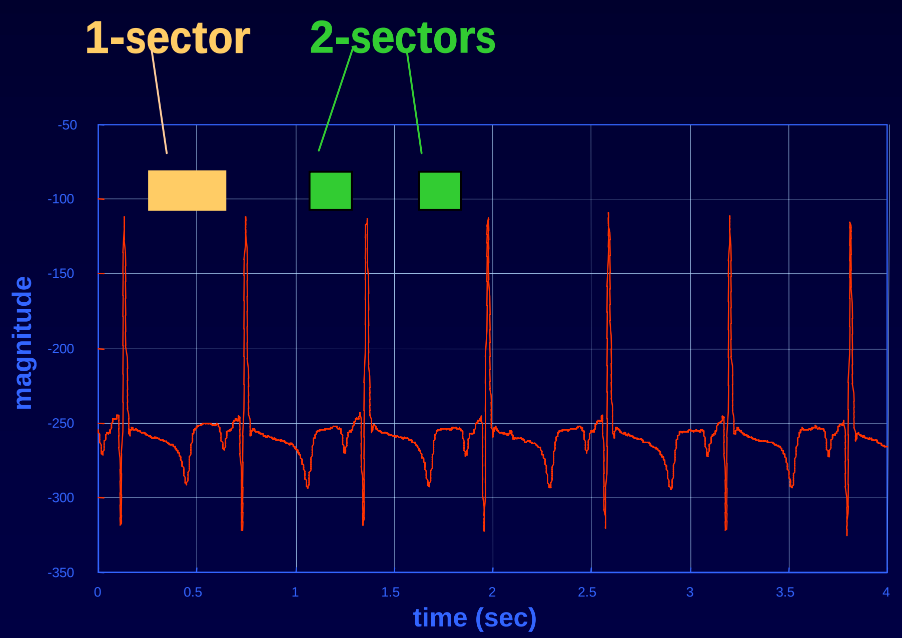
- 25 g at 0.35 s
- 8X safety margin → 200 g
- 76 g at 0.2 s
- 8X safety margin → 612 g

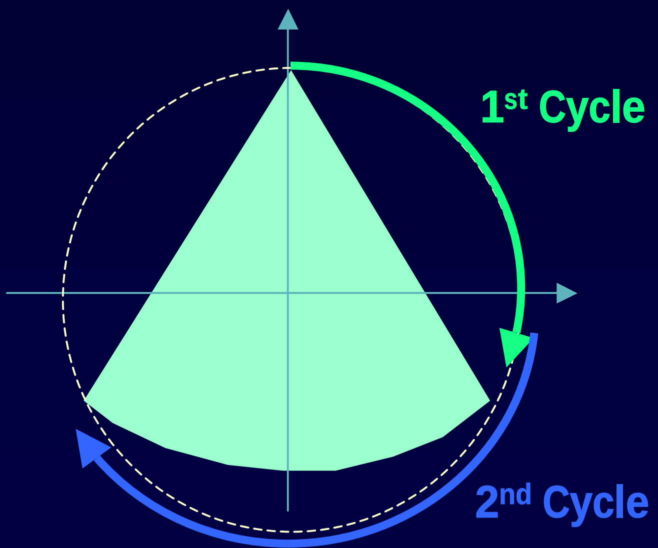
Temporal Resolution Improvement

Other methods to improve temporal resolution:

- Half-scan
  - 230°-240° rotation → 35-40% speedup
- Multi-sector recon
  - 120°-130° rotation → 45-50% speedup





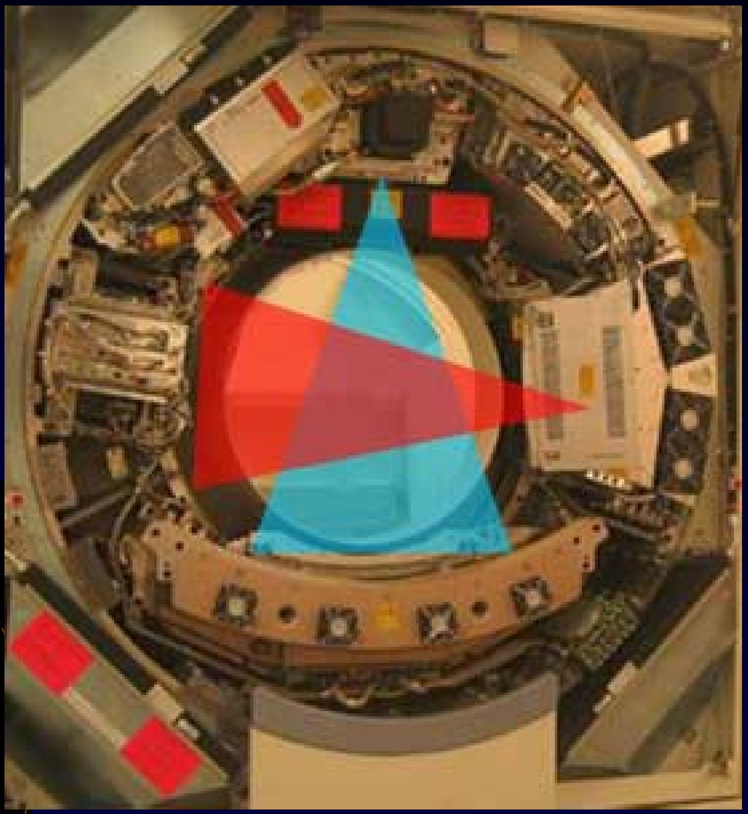


Dual Source CT

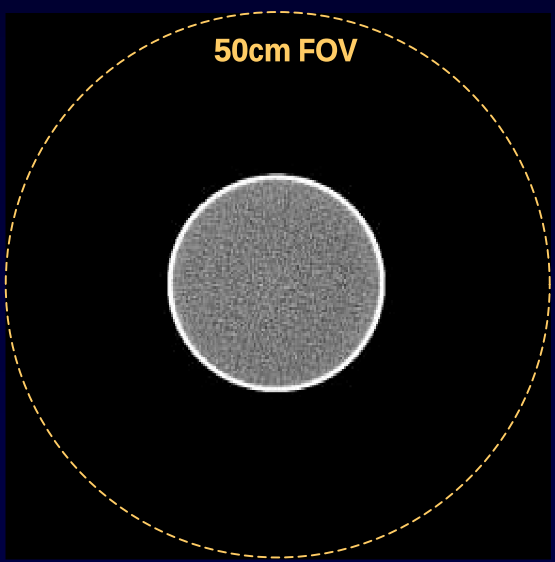
Dual Source Approach

Cons:

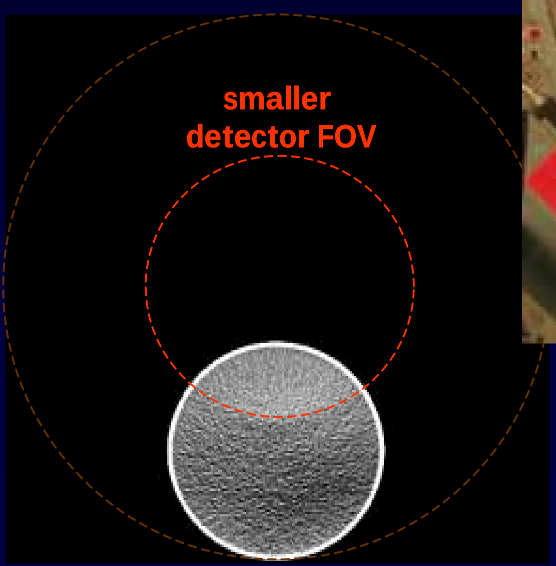
- Reduced FOV (26-33 cm)
- Scatter radiation from 2 sources



centered phantom



off-centered phantom



Prior Image Constrained Compressed Sensing (PICCS)

- Joint research with University of Wisconsin-Madison results in significant artifact reduction in animal studies.
- Redundant information present even for half-scan data acquisition.

Temporal resolution improvement using PICCS in MDCT cardiac imaging

Guang-Hong Chen[a)]
Department of Medical Physics, University of Wisconsin–Madison, Madison, Wisconsin 53705 and Department of Radiology, University of Wisconsin–Madison, Madison, Wisconsin 53705

Jie Tang
Department of Medical Physics, University of Wisconsin–Madison, Madison, Wisconsin 53705

Jiang Hsieh
GE Healthcare, 300N Grandview Boulevard, Waukesha, Wisconsin 53188

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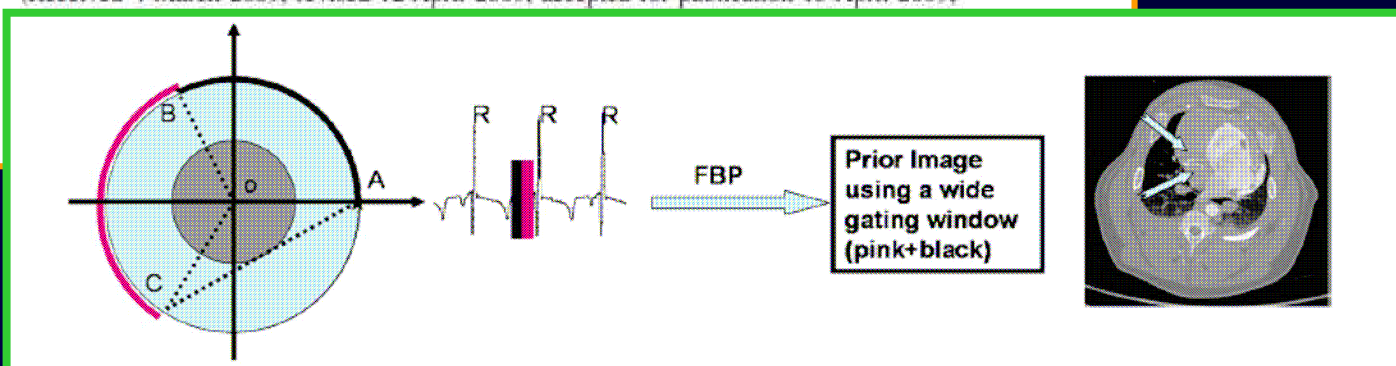
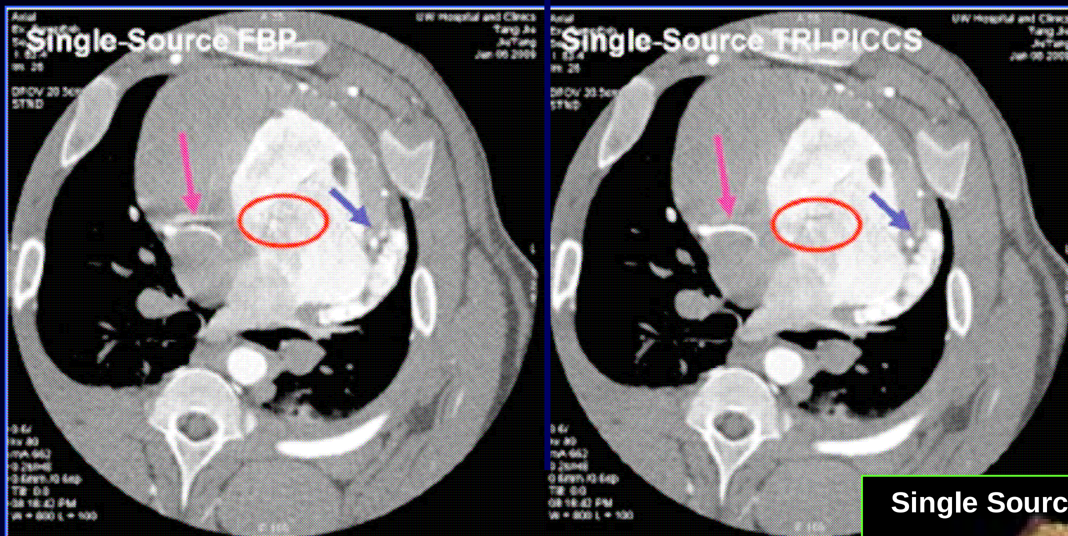


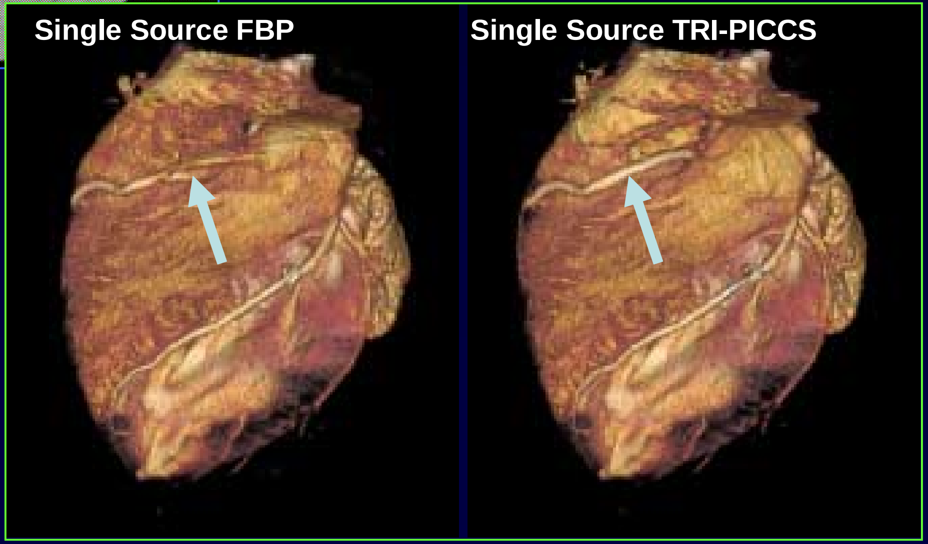
PICCS

Animal Experiment – 96+/-5bpm



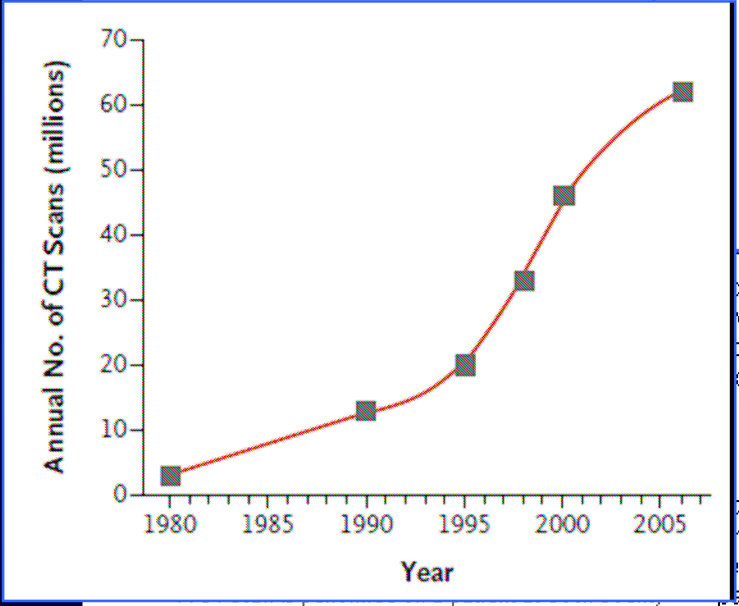
FBP PICCS

120kV
600mA
0.35s,
HR: 96+/-5bpm



FBP PICCS

X-ray CT Radiation



Stroger Hospital in Chicago in May.

The NEW ENGLAND JOURNAL of MEDICINE

REVIEW ARTICLE

CURRENT CONCEPTS

Computed Tomography — An Increasing Source of Radiation Exposure

David J. Brenner, Ph.D., D.Sc., and Eric J. Hall, D.Phil., D.Sc.

The advent of computed tomography (CT) has revolutionized diagnostic radiology. Since the inception of CT in the 1970s, its use has increased rapidly. It is estimated that more than 62 million CT scans per year are currently obtained in the United States, including at least 4 million for children.[1]

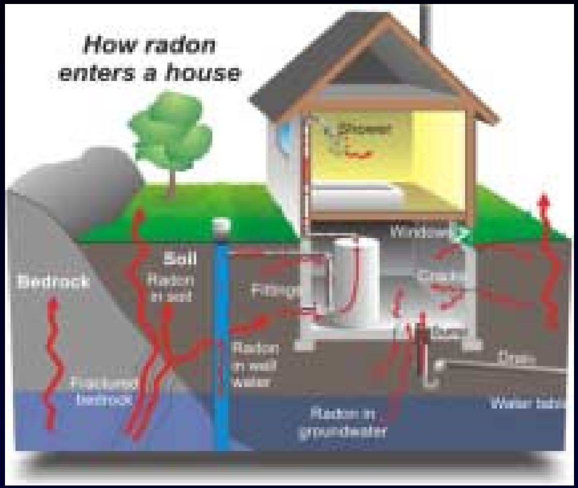
By its nature, CT involves larger radiation doses than the more common, conventional x-ray imaging procedures (Table 1). We briefly review the nature of CT

From the Center for Radiological Research, Columbia University Medical Center, New York. Address reprint requests to Dr. Brenner at the Center for Radiological Research, Columbia University Medical Center, 630 W. 168th St., New York, NY 10032, or at djb3@columbia.edu.

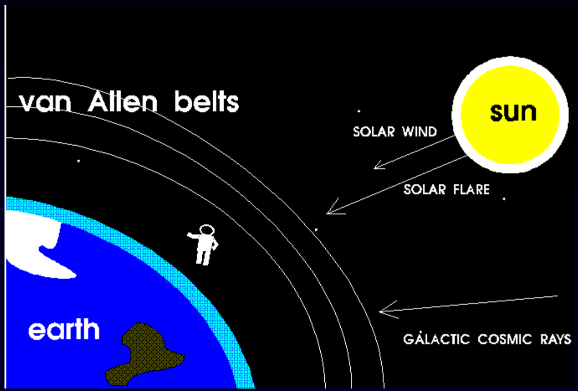
Table 1. Typical Organ Radiation Doses from Various Radiologic Studies.

Study Type	Relevant Organ	Relevant Organ Dose* (mGy or mSv)
Dental radiography	Brain	0.005
Posterior–anterior chest radiography	Lung	0.01
Lateral chest radiography	Lung	0.15
Screening mammography	Breast	3
Adult abdominal CT	Stomach	10
Barium enema	Colon	15
Neonatal abdominal CT	Stomach	20

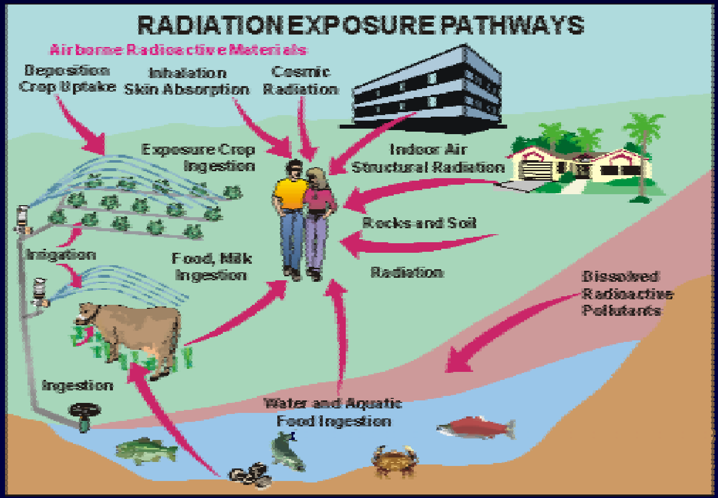
Radiation Sources



Radon Gas



Space Radiation



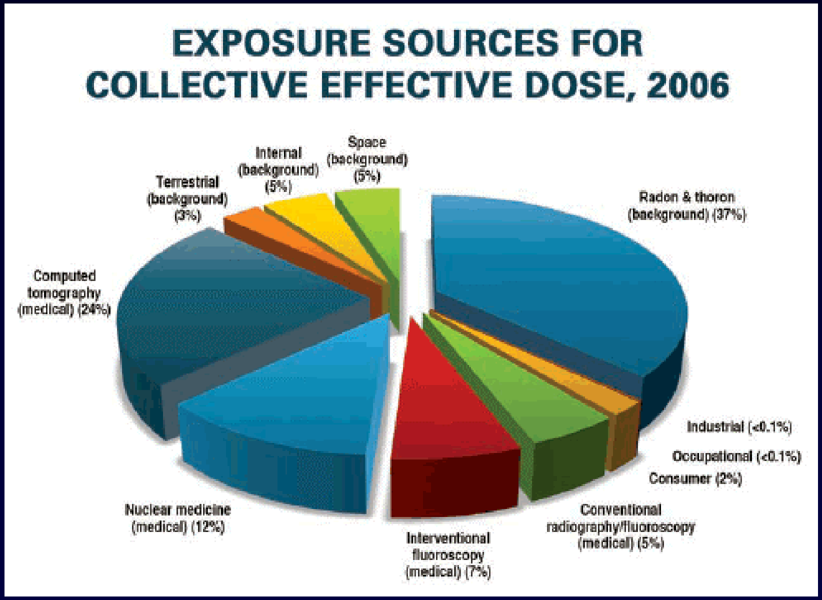
Maternity Radiation Dress

Computer Radiation Cleaner

Sources of Radiation

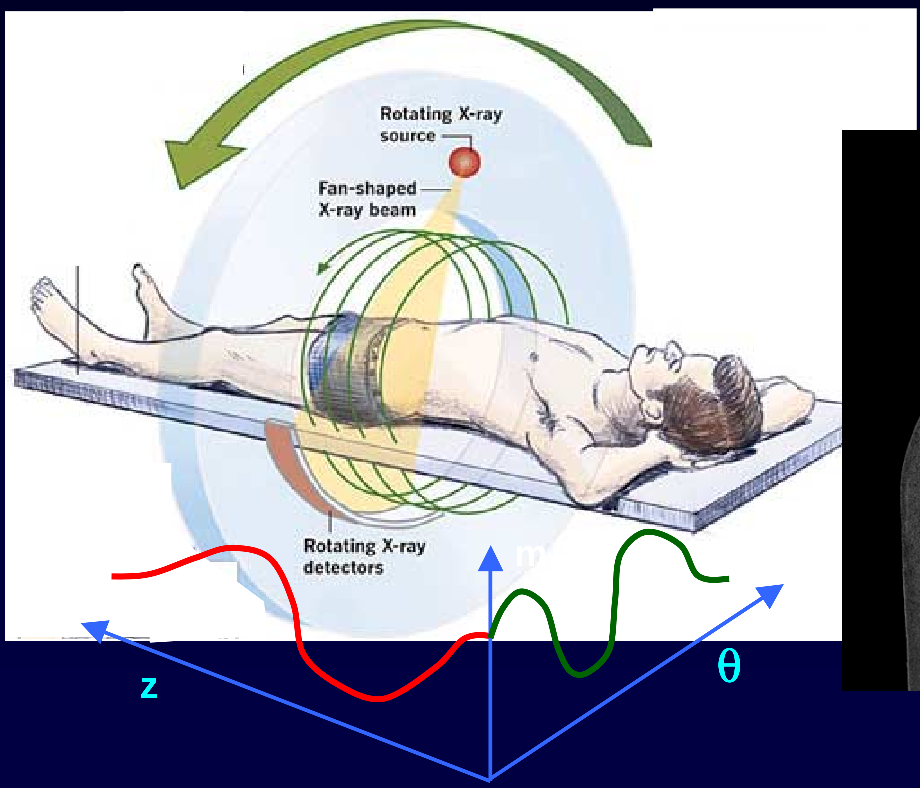
- Background radiation dose consists of the radiation doses received from natural and man-made background.

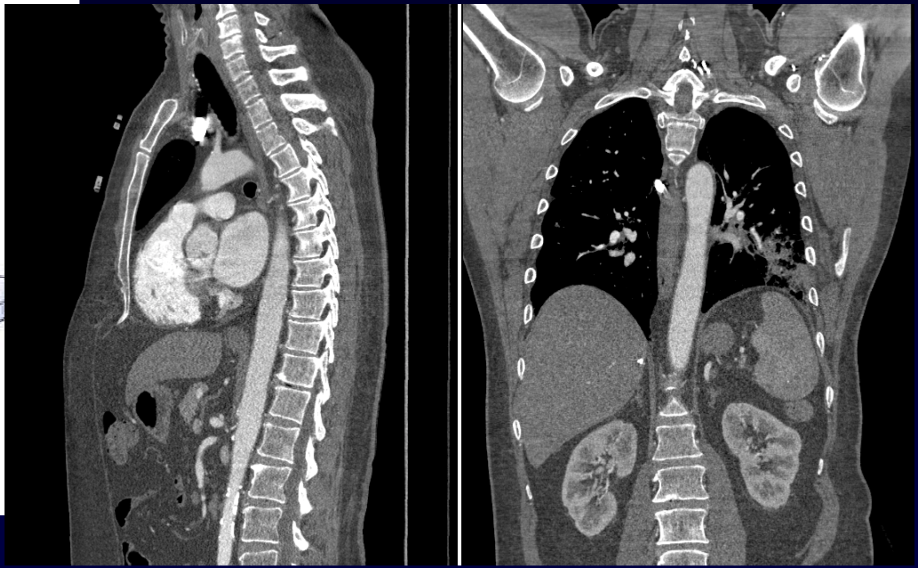
- **The annual background radiation exposure for a typical American 3.70 mSv.**
- **The average dose from watching color TV is 0.02 mSv each year.**
- **The granite from Grand Central Station exposes its employees to 1.20 mSv of radiation each year**
- **People in Denver receive 0.50 mSv more each year than those in LA because of the altitude.**
- **Medical imaging procedures contribute to nearly ½ of the total radiation.**

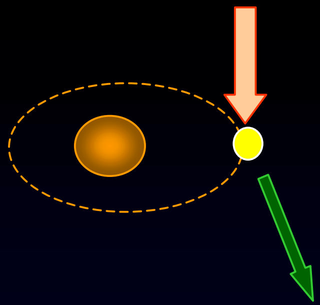


Tube Current Modulation

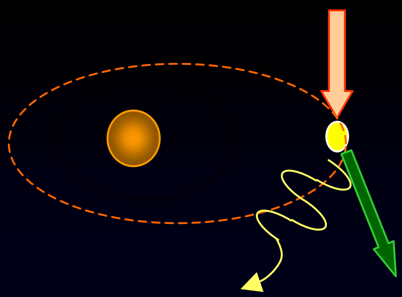
- Human bodies are not cylindrically shaped
- Attenuation to x-ray depends on the projection orientation and anatomy location
- Tube current should change based on the attenuation variation







Dual-energy Imaging

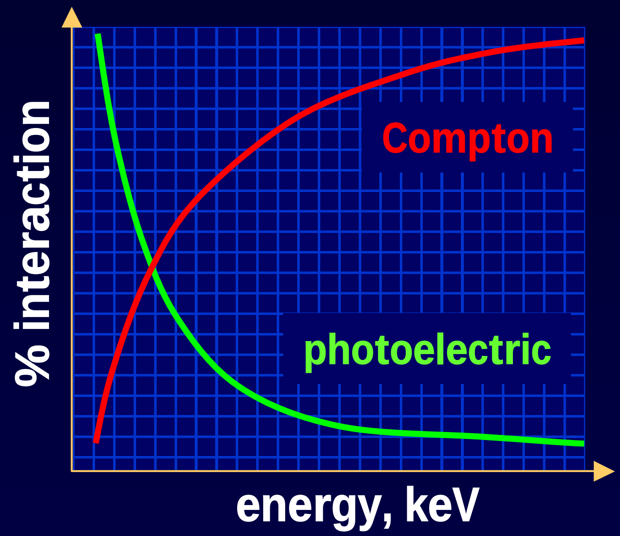


- Concept proposed in the 70's.
- Two x-ray / matter interactions: photoelectric & Compton.
- Mass attenuation coefficient can be expressed as the linear combination of the Photoelectric function, f_p, and the Compton function, f_c.

$$\left(\frac{\mu}{\rho}\right)(E) = \alpha_p f_p(E) + \alpha_c f_c(E)$$

- Also be expressed as a linear combination of the mass attenuation coefficient of two materials.

$$\left(\frac{\mu}{\rho}\right)(E) = \beta_A \left(\frac{\mu}{\rho}\right)_A (E) + \beta_B \left(\frac{\mu}{\rho}\right)_B (E)$$



Material Basis

- Measured projections from high- and low-kVp, I_L and I_H, are related to the density projections, η_A and η_B, of materials A and B:

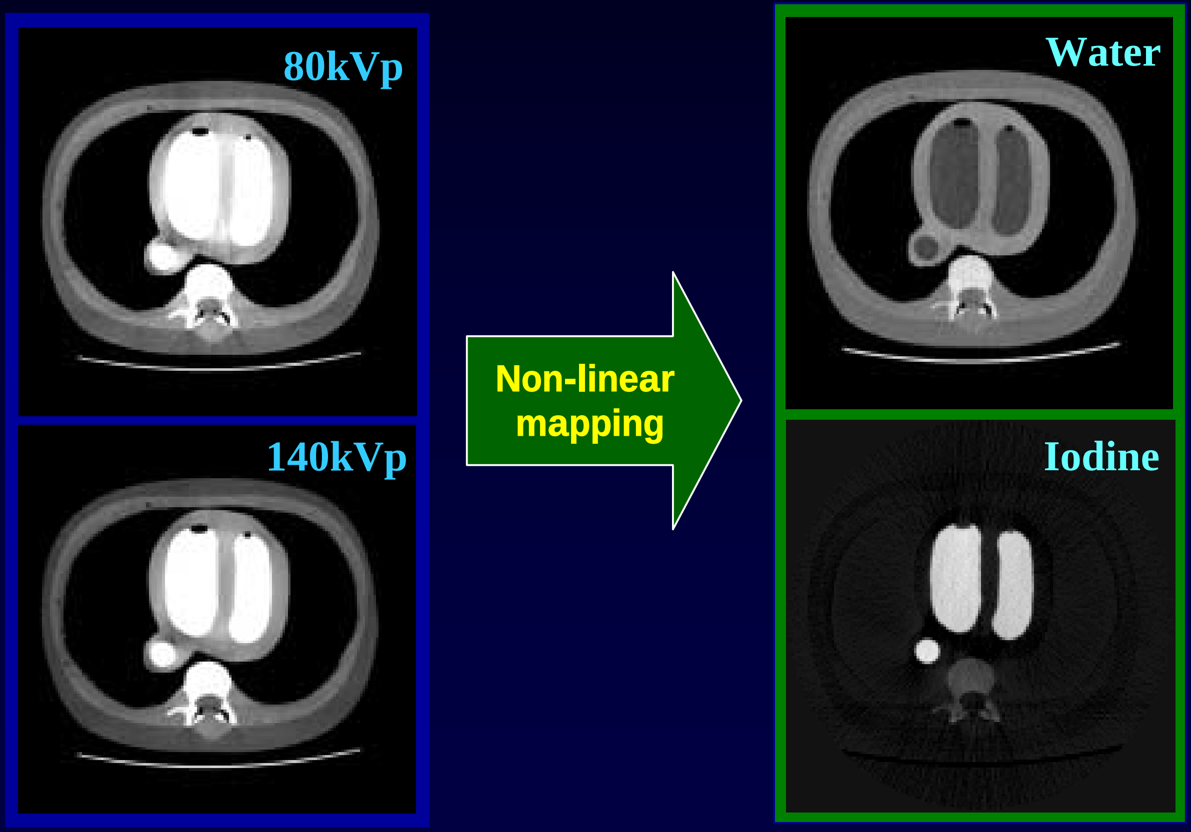
$$I_L = \int \psi_L(E) \exp\left[-\eta_A \left(\frac{\mu}{\rho} \right)_A (E) - \eta_B \left(\frac{\mu}{\rho} \right)_B (E) \right] dE$$

$$I_H = \int \psi_H(E) \exp\left[-\eta_A \left(\frac{\mu}{\rho} \right)_A (E) - \eta_B \left(\frac{\mu}{\rho} \right)_B (E) \right] dE$$

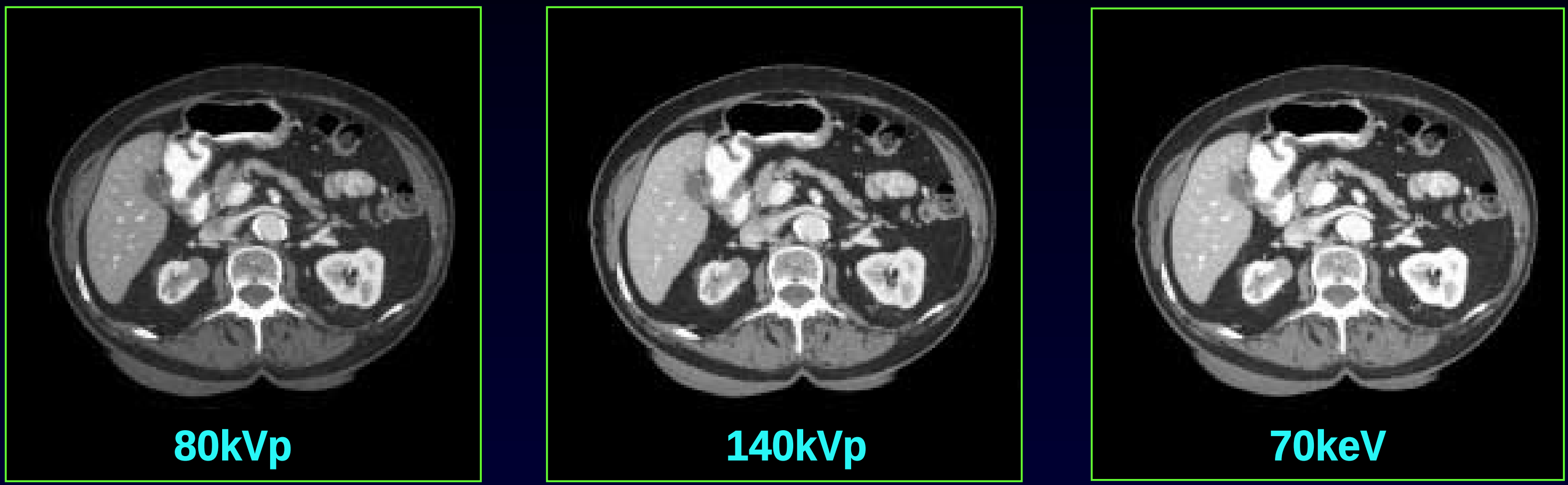
- Density projections η_A and η_B, can be solved in terms of I_L and $I_{H.}$
- Reconstruction of η_A and η_B lead to equivalent-density images of materials A and B.

Equivalent-density Images

- Non-basis materials are mapped to both.
- Equivalent-density images are not in HU, but in g/cm^3

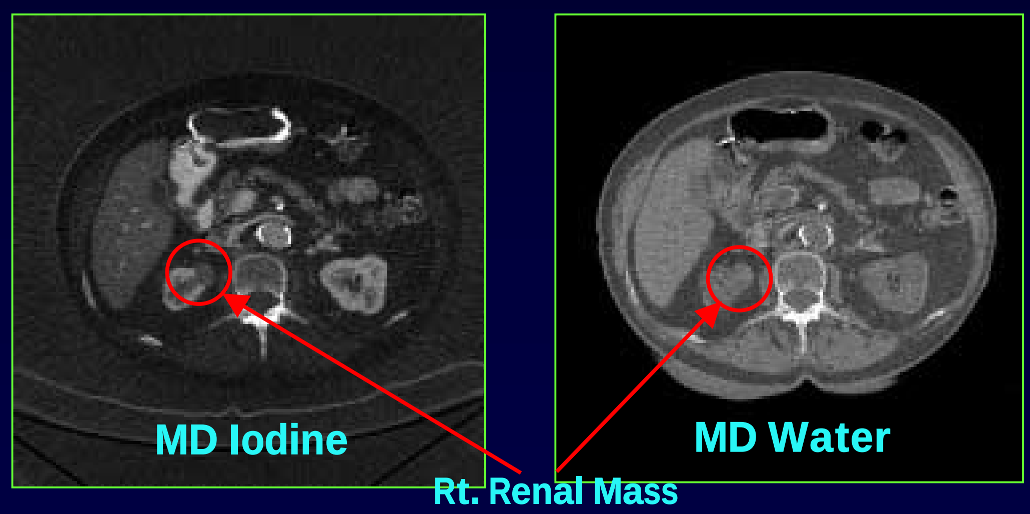


Hypodense Renal Cell Carcinoma



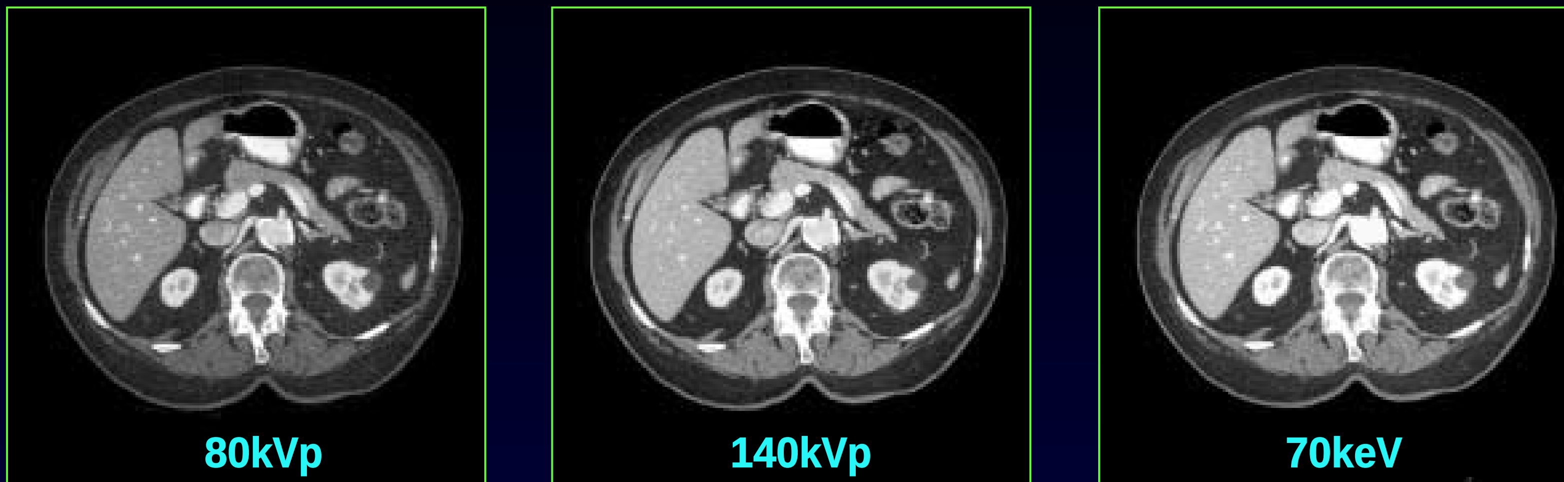
MD Iodine Image:
Shows enhancement confirming malignancy

MD Water Image:
Shows lesion is slightly hyperdense (Not a cyst)

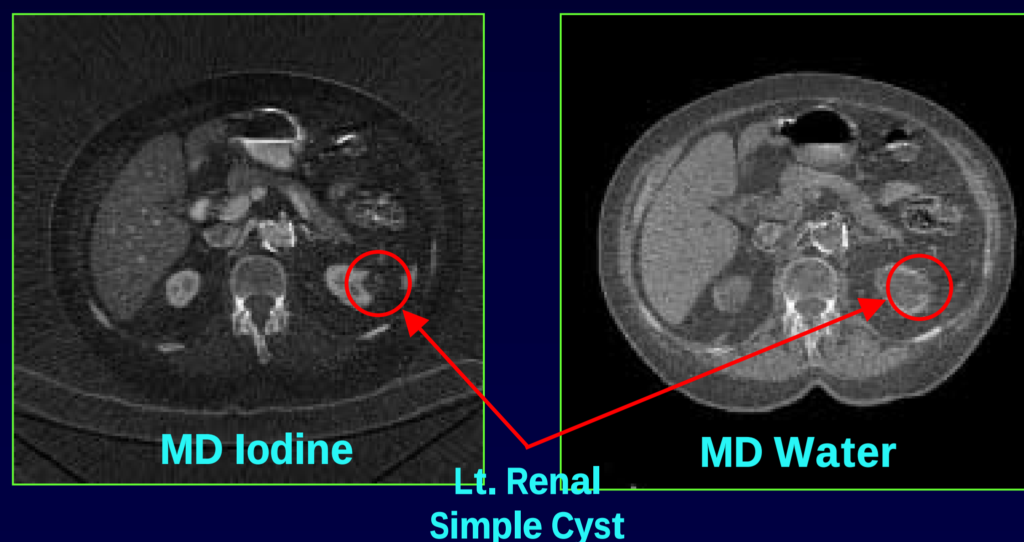


Images courtesy Mayo Clinic Scottsdale

Simple Renal Cyst



Left Renal Simple Cyst
Comparison to Rt.
Renal Carcinoma
(Previous Slide)



Images courtesy Mayo Clinic Scottsdale

Data Acquisition Approaches

Spectrum Optimization

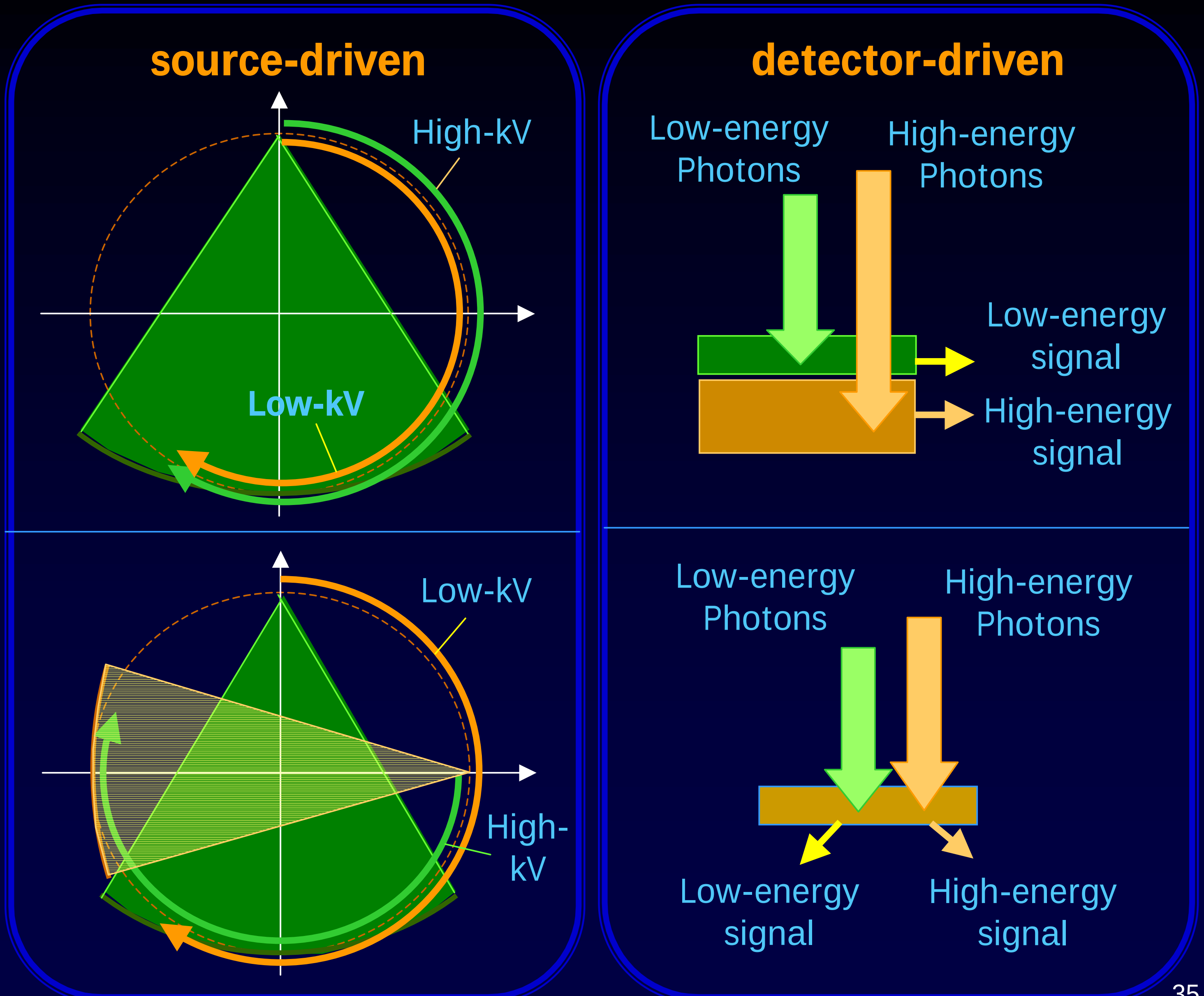
Motion

Low-high Adjustment

Coverage

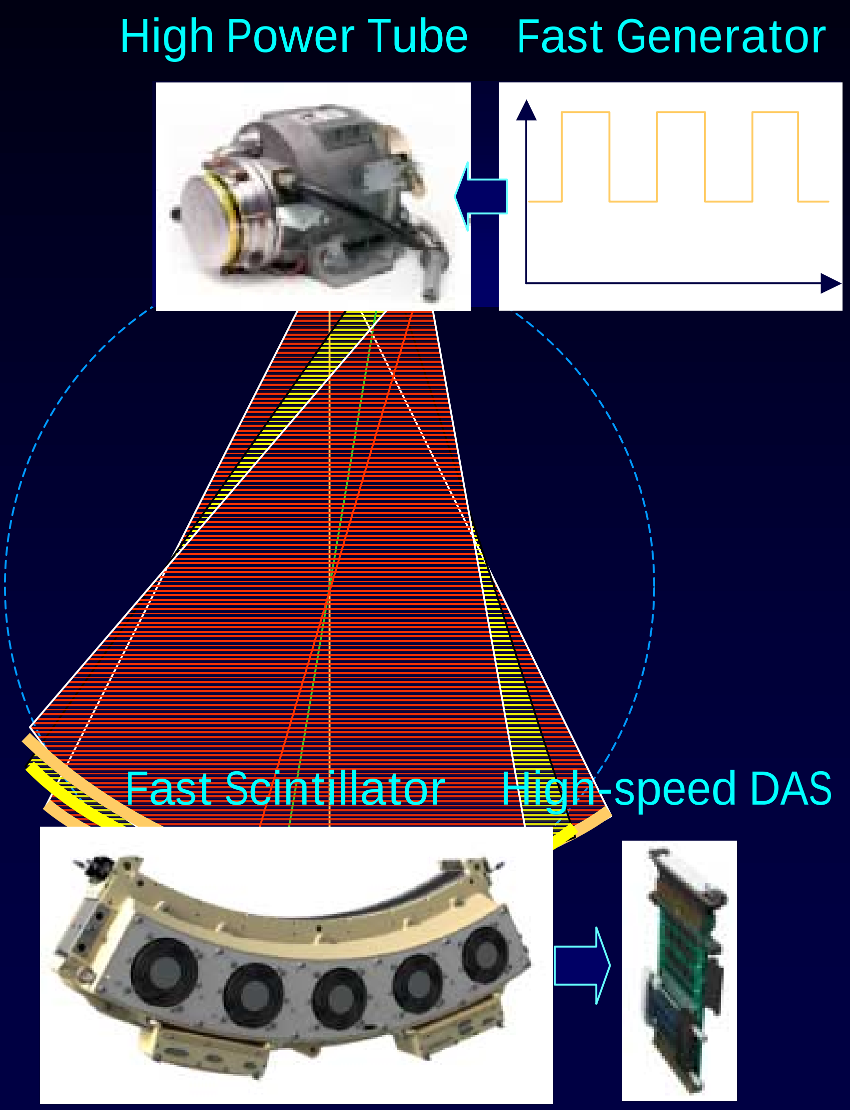
Projection vs. Image space

Complexity



Fast kV Switching

- Change kVp setting on a view by view basis.
  - High- and low-kV are toggled every view
  - Little patient motion
  - Allow projection space processing
- Require fast generator response.
- Require fast scintillator response.



Information Explosion

1998 (4-slice)

Runoff 1200 mm @ 2.5mm
Acquisition time: 65 sec
No. Images: 500-1000

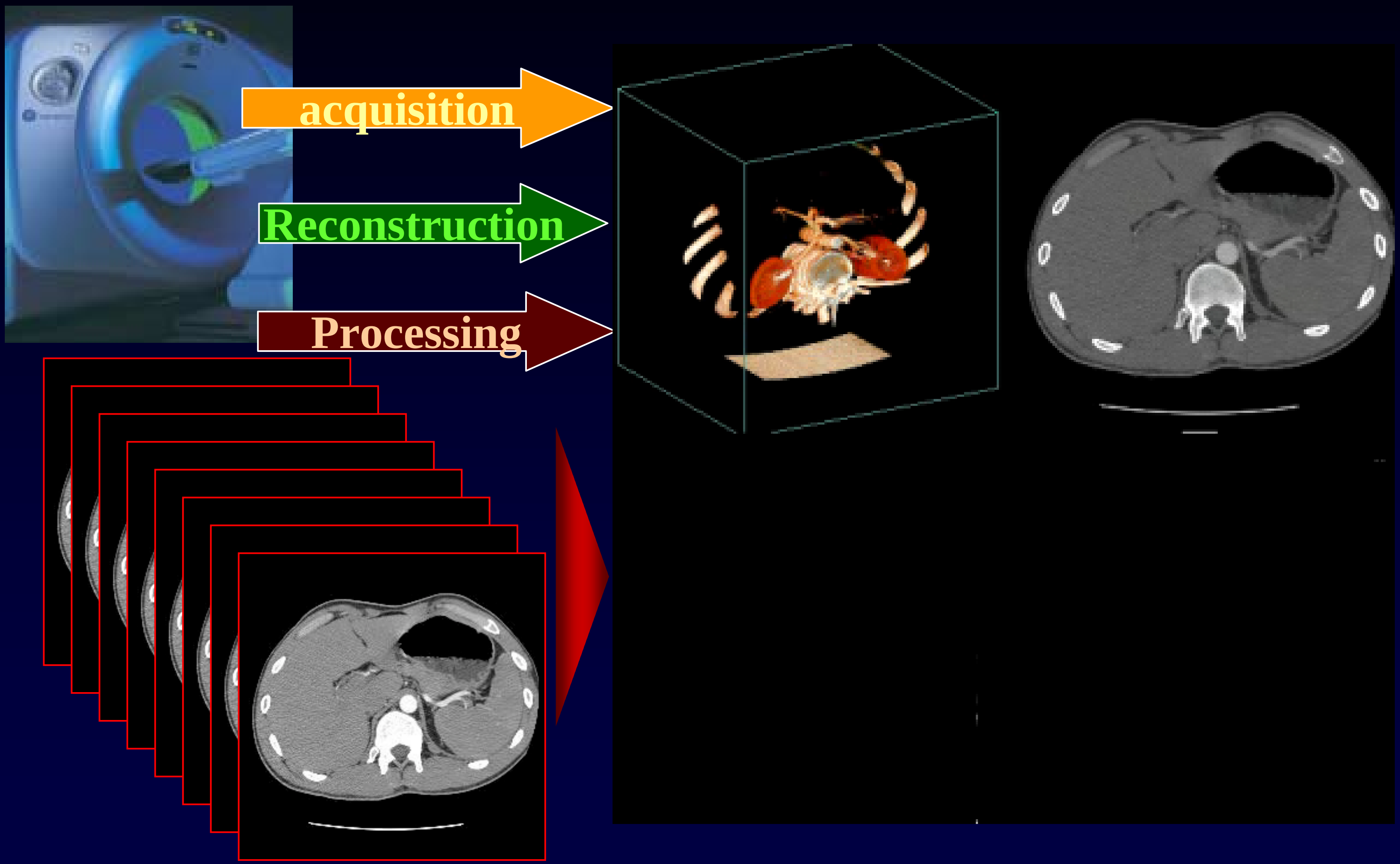
2005 (64-slice)

Runoff 1200 mm @ 0.625mm

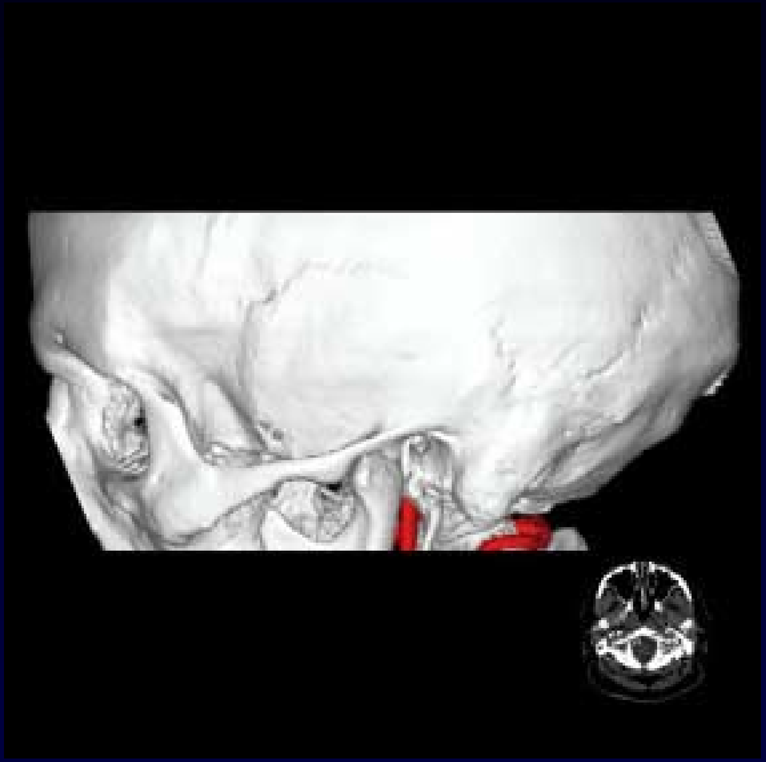
Acquisition time: 9 sec

No. Images: 2000-4000

“Real Time” Reconstruction



Automatic Bone Removal



Volume Rendered View

Thank You!