Inverse Geometry CT

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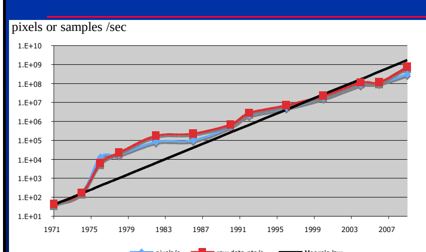
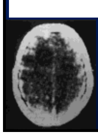
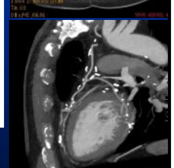
CT Speed Gains



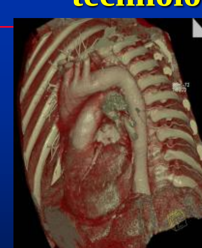


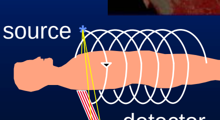
2 80x80 10-bit images
every 20 minutes
~ 100 bits/s



320 512x512 12 bit
images every ~0.28
seconds ~ 3.5 Gbit/s

Multi-slice CT has been the driving technology

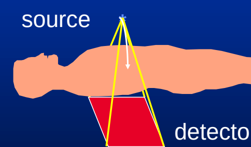




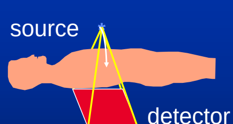
Motivation for faster volumetric coverage

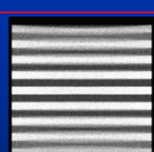
- CT angiography
- More reliable cardiac imaging
- Dynamic imaging
  perfusion imaging
- Patient comfort
- Throughput

Cone-beam CT



Cone-beam CT

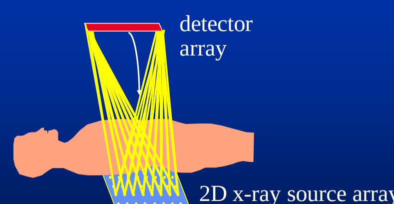




Sourbelle, et al, RSNA 2003, #183

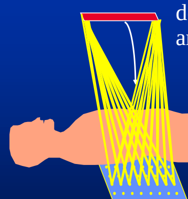
cone-beam artifacts
severity increases with increasing axial coverage

Inverse-Geometry CT



- sources energized one at a time: needs fast detector
- each source illuminates a fraction of the volume: flux challenge

Inverse-Geometry CT



source and detector have ~ same axial extent
to avoid cone beam artifacts

Potential benefits of IGCT

- reduced cone beam artifacts
- more uniform spatial resolution (less variation in apparent focal spot size)
- less scatter (smaller source solid angle)
- more efficient detectors
- virtual bowtie

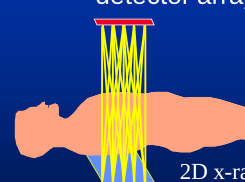
Scanned source IGCT

SBDX scanned anode x-ray source

~5x5 cm^2 photon counting array

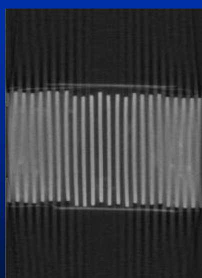
NovaRay

Reconstruction Algorithm

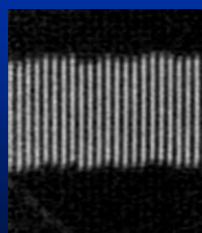


PET-type FBP reconstruction

Cone-beam artifacts

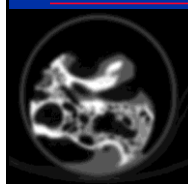


Cone-beam μCT
+/- 5 degree cone-angle

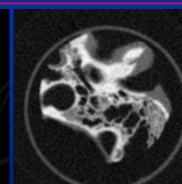


IGCT

Cadaver inner ear

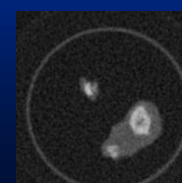


GE Lightspeed 16
axial 0.625 mm slice
200 mAs, std recon



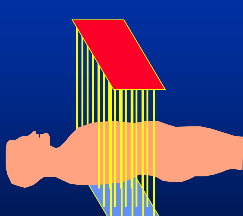
IGCT

360 slices
0.125 mm spacing
~0.25 mm thickness

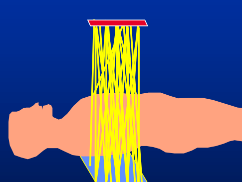


Schmidt, et al, Med Phys 33, 1867-78, 2006.

Photon Efficiency



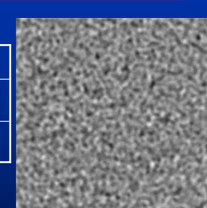
2D Parallel Ray



IGCT

Results: Photon Efficiency

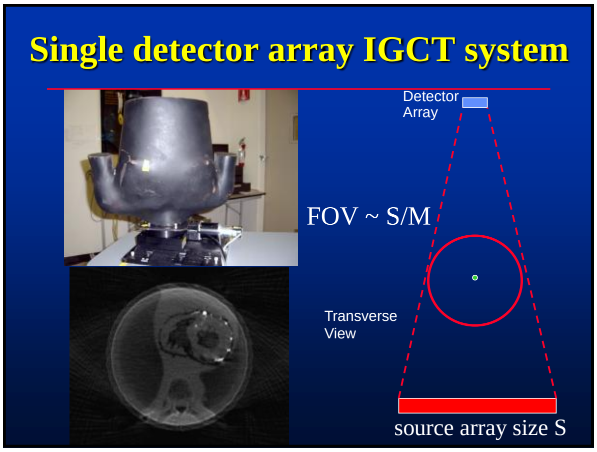
Theoretical	**σ=10 HU**
Parallel ray	**σ=10.5 +/- 1.1 HU**
IGCT	**σ=9.4 +/- 1.2 HU**

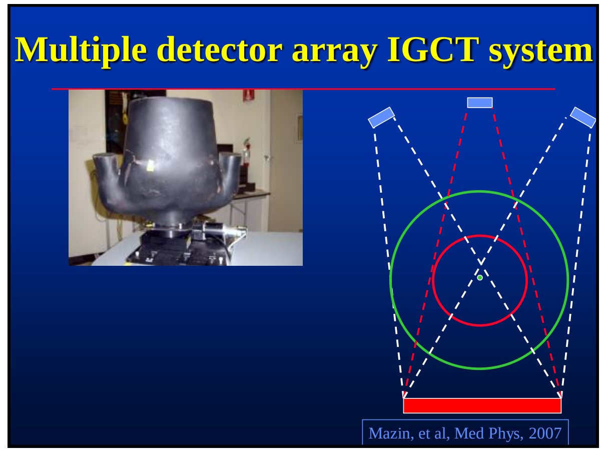


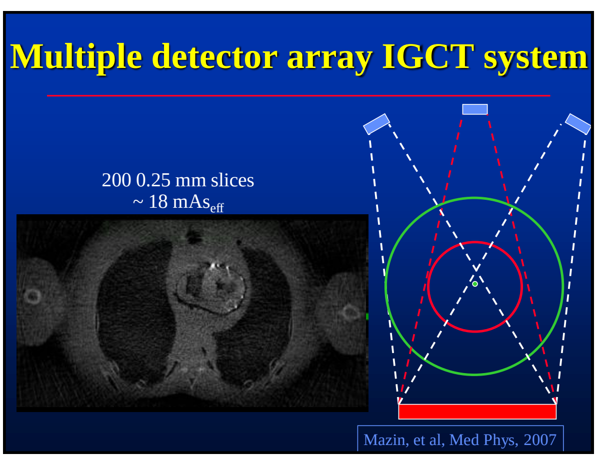
Reconstruction filter cutoff:
7.5 lp/cm

Level: -1000 HU
Window: +/- 50 HU

No noise penalty for using cross plane rays


Single detector array IGCT system
Detector Array
FOV ~ S/M
Transverse View
source array size S


Multiple detector array IGCT system
Mazin, et al, Med Phys, 2007


Multiple detector array IGCT system
200 0.25 mm slices
~ 18 mAs_{eff}
Mazin, et al, Med Phys, 2007

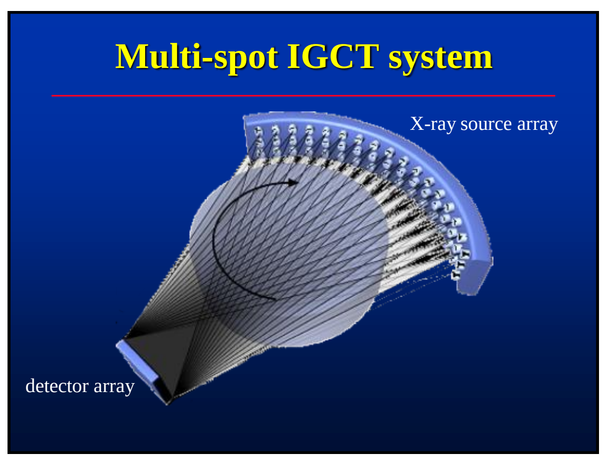
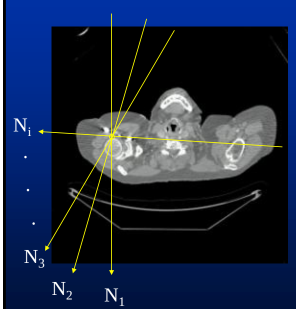

Multi-spot IGCT system
X-ray source array
detector array

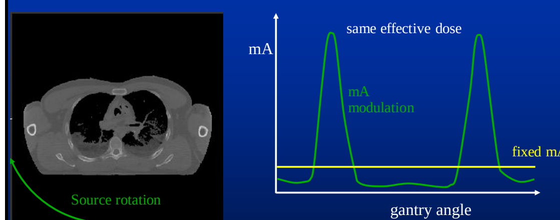
image noise



$$\sigma^2 \sim \sum_{\text{views}} \frac{1}{N_i}$$

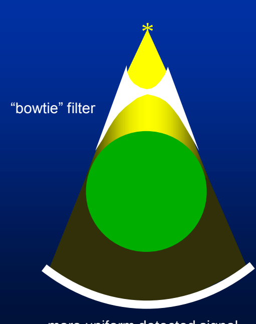
A few measurements with low intensity can dominate noise

mA modulation



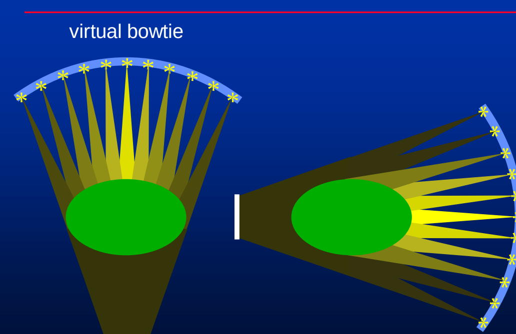
mA modulation can't change the intensity distribution across fan beam

"Bowtie" filters



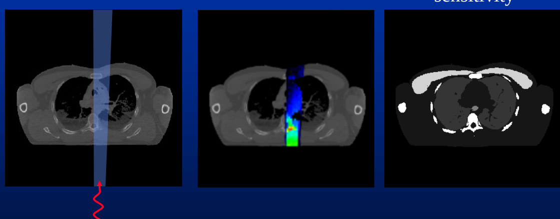
can't adapt to the object being scanned or the imaging task

"Virtual" bowtie



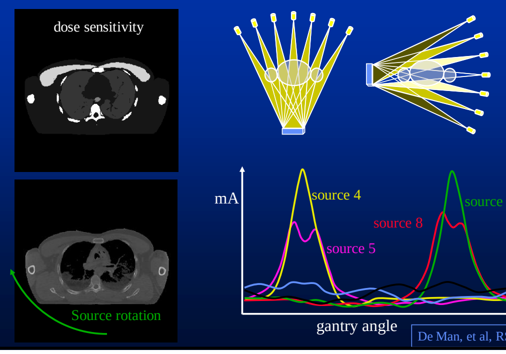
Optimization problem

each beam contributes to CNR | and delivers radiation dose | relative dose sensitivity



Solve for intensity of each source in each view to optimize noise for a given dose

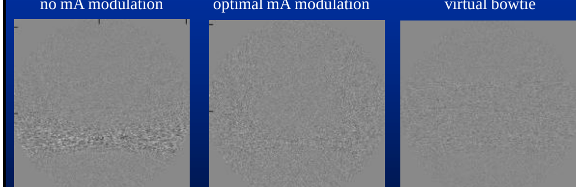
Example: 12 sources, minimize peak variance



De Man, et al, RSNA 2007

Variance maps, same effective dose

bowtie filter no mA modulation | bowtie filter optimal mA modulation | 12 source IGCT virtual bowtie



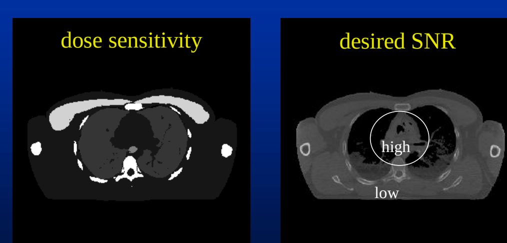
57% lower peak variance

61% lower peak variance

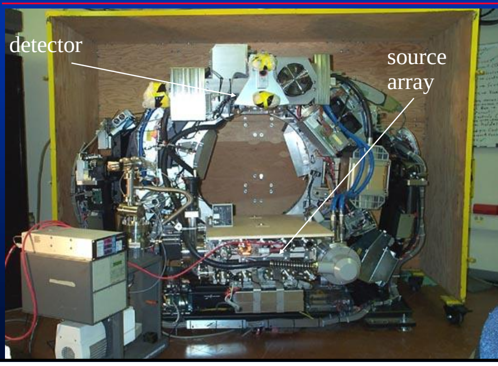
83% lower peak variance

De Man, et al, RSNA 2007

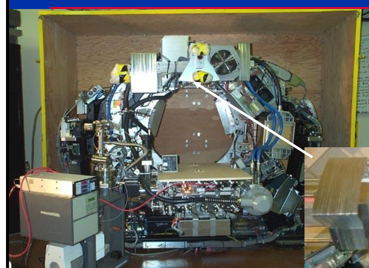
Further enhancements



IGCT gantry



Detector array

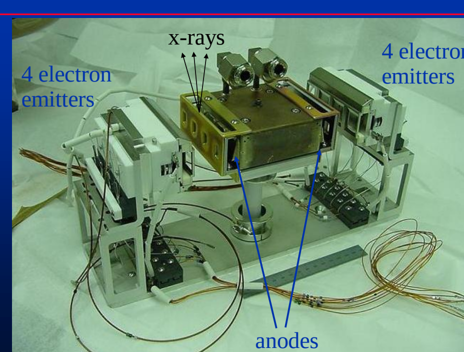


64 x 256 channels
70 mm (in x)
by 196 mm (in z)

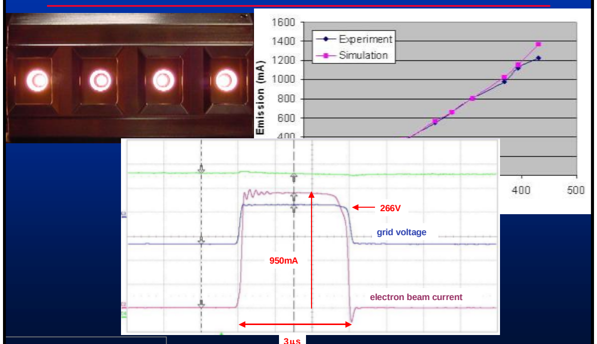
Source array

up to 4
multi-spot
modules

2x4 source module



Electron emitters



Courtesy of Kris Frusche

32 source array

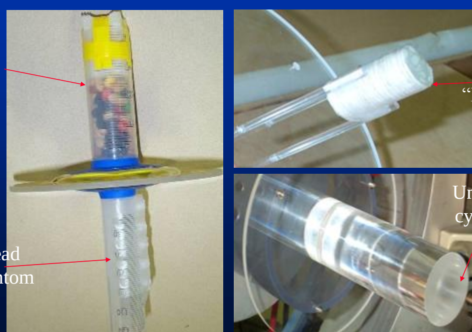
Gantry integration and balancing

Experimental parameters

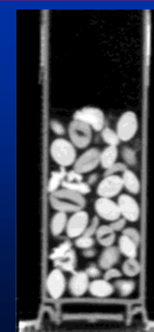
scan time	1 sec
sources	8
pulses per source	125
kVp	80
mA	125*
time per pulse	5.45 µsec

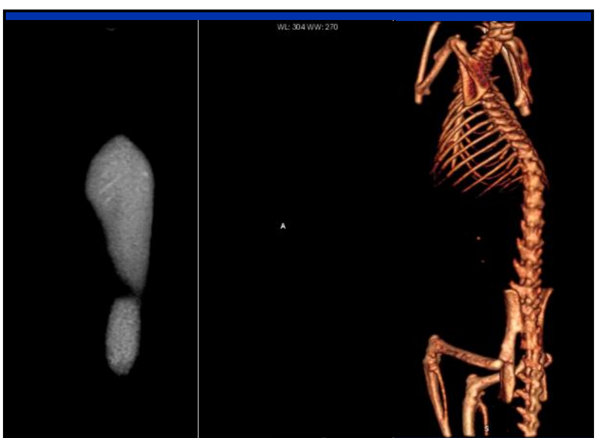
* not at the thermal limit, could be higher

Phantoms

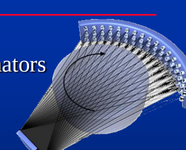


Reconstructed coronal images



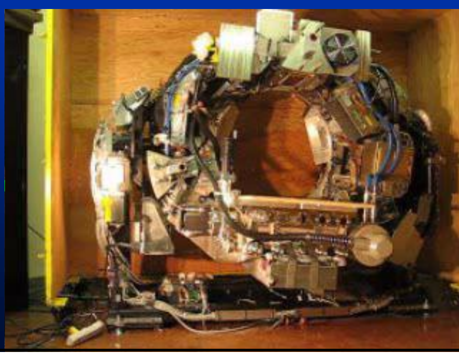


X-ray flux challenge

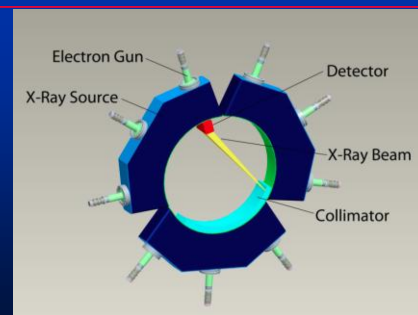


- Cause
  - More x-rays removed by our collimators
    - especially as FOV increases
  - Stationary anode x-ray sources
- Solutions
  - Higher power
    - more sources, short pulse times, lower duty cycle
  - Virtual bowtie
  - Higher efficiency detectors
  - Statistical reconstruction algorithms
- Target lower dose systems

Gantry integration and balancing



Stationary source IGCT



only collimator and detectors rotate

Summary

- IGCT potential advantages
  - Scalable axial coverage, no cone beam artifacts
  - Uniform spatial resolution across FOV
  - Better dose efficiency ("virtual bowtie")
- Disadvantages
  - No anti-scatter grid (also an advantage)
  - X-ray flux challenge
- Much work remains
  - Demonstrate virtual bowtie, IQ studies, future configurations
- … but initial results are promising

Thank you !

Stationary source IGCT

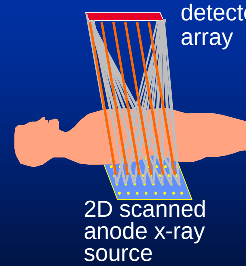
Source to isocenter distance	60 cm
Detector to isocenter distance	50 cm
Field of view (diameter)	44 cm for circular objects
Detector size	10 cm (transverse) by 15 cm (axial)
Number of detector cells	101 by 151
Number of source arrays	3
Source array length	120 cm
Gaps between source array	6 cm
Number of views (N) including missing views	1810
Missing views (in gap)	105
Firings per view (M)	21
Number of	
Total numl	
Source spo	
Move time	

parallel beam

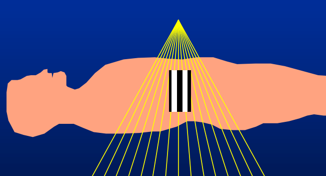
ssIGCT

Inverse-Geometry CT

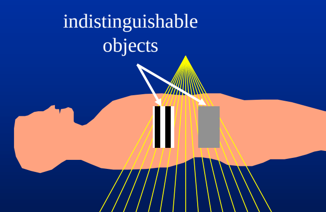
- In-plane sampling is fan-like
- no cone beam artifacts
- Extra rays reduce noise



Cone-beam artifacts



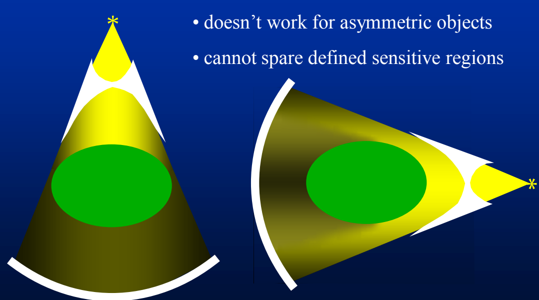
Cone-beam artifacts



Fundamental sampling problem, no correction exists

“Bowtie” filters

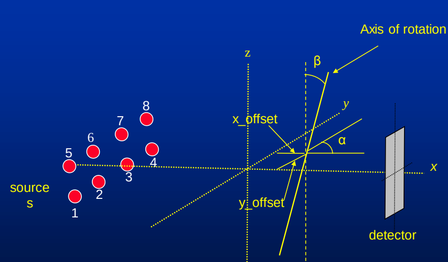
- cannot be ideal for all objects
- doesn’t work for asymmetric objects
- cannot spare defined sensitive regions



Recent progress

- First rotating gantry experiments
- Geometric calibrations
- Algorithm refinements
- Initial images

Geometric calibration



need to measure the spatial parameters of sources, axis of rotation, and detector array

“Bowtie” filters