**Physics of x-ray computed tomography**

- how to generate a CT image?
- CT scanner, now and past
- clinical applications
- image artifacts
- radiation dose

CT provides 3-D, quantitative images with high radiation dose.

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**Radiography: 2-D**

- acquire x-rays that travel through the object to form a projection image → measure photon attenuation in object
- superimpose a 3-D object onto a 2-D detector or film
  - loss of depth information
  - poor contrast
  - no quantitation

**CT: 3-D**

- x-tube and detector rotating in synchrony around the patient to acquire a series of projection images from different angles (views)
- image reconstruction from the projection images
- image display
  - transaxial, sagittal, coronal, rendering
  - display windows
CT pixel value: Hounsfield unit (HU)

- Both measure x-ray attenuation coefficients of tissues.
- For CT, \( HU = 1000 \cdot \frac{\mu_{\text{tissue}} - \mu_{\text{water}}}{\mu_{\text{water}}} \)
- HU depends on photon energy and tissue.
  e.g. @73 keV, water: 0, bone: 1000, air: -1000, muscle: 50, fat: -100, lung: -500

CT vs radiography

- Better detectable contrast (1%)
- Poorer spatial resolution (~ 0.5 mm)
- More noise
- Much higher (> 100 times) radiation dose

Data acquisition

- Count (number) on the detector depending on the position on detector (d) and projection angle (θ)
  \( P(d, \theta) = \ln(I_0/I_n)/x = \mu_1 + \mu_2 + \ldots + \mu_n \)
  \( I_0 \): initial x-ray intensity, \( I_n \): survived x-ray intensity
  \( x \): pixel size, \( \mu_i \): attenuation coefficient of the \( i \)th pixel

Axial and spiral CT

- Axial (step-and-shoot) CT: turn on x-tube, rotate x-tube and detectors to acquire data, turn off x-tube, and translate table to next slice, then repeat the steps
- Spiral CT: translate table as x-tube and detector rotate to continuously acquire data
  - Rotation speed < 3 turns/s
  - Table index (TI) > 10 mm
  - Scan time < 100 sec
  - X-tube current: 20 - 500 mA
  - Peak voltage: 120 or 140 kVp
  - Output power: 60 – 100 kW
**Advantages of spiral CT**

- Fast volumetric CT \(\rightarrow\) reduced motion artifacts, high patient throughput
- Slice thickness can be determined at will in image reconstruction, not in data acquisition.
- Equivalent image quality as axial CT
- Similar radiation dose to axial CT

**Multi-slice spiral CT**

- Single-slice
- Multi-slice
- Multiple row of detectors \(\rightarrow\) faster volumetric CT

**MSCT beam pitch**

- Beam pitch = \(TI/BW\)
- Table index (TI): Table travelling distance during one rotation (360°) of the x-ray tube
- Beam width (BW): Cover all slices
- Pitch in the range of 0.13 – 1.5
Image reconstruction

- determine the $\mu$ or HU of each voxel of a CT image using a math algorithm
- two types of algorithms:
  - filtered backprojection algorithm, fast but poor noise characteristic
  - iterative algorithm such as OSEM recently introduced to better handle noise and hence reduce radiation dose, slow but acceptable in clinic

CT: transaxial slices

A transverse slice contains only structures in the slice.

Reformatting to sagittal and coronal images

A slice of sagittal or coronal image
**Volume and surface rendering**

**Display windows**

- manipulate gray scale to optimize the display of selected tissues
- two parameters:
  - window width: a range of HU values
  - window level: center of the window

**Effects of WW and WL**

- WW determines how many tissues in gray range. A narrow WW produces better contrast but smaller gray range and worse display noise.
  - 100 HU for brain, 200 HU for liver, 500 HU for soft tissue, or 1500 HU for lung and bone
- WL determines what tissues in gray.
  - WL of lung: white liver and pelvis
  - WL of liver: white pelvis and black lungs
  - WL of pelvis: black liver and lungs

**Question 1**

The main advantage of CT over radiography is

A. less radiation dose.
B. better contrast.
C. lower noise.
D. less artifacts.
Question 2
In an MSCT study, the table movement is 10 mm per gantry rotation and the beam width is 8.3 mm. What is the pitch?
A. 1
B. 1.5
C. 1.2
D. 0.8

Question 3
Use of intravascular contrast in CT will significantly increase the
A. blood HU.
B. required kVp.
C. required mA.
D. patient dose.

Question 4
Display contrast of CT images
A. must be determined before the scan.
B. may be adjusted after the scan.
C. can be used to change the HU values.
D. does not modify the appearance of the CT image.

Question 5
The HU of a tissue is 100. The tissue
A. appears black when a display window with a width of 50 HU is centered at 50 HU.
B. changes to 50 HU if the slice thickness is reduced to one half.
C. has a density greater than 1 g/ml.
D. has smaller attenuation coefficient than water.
**Question 6**

A CT image has 64 slices and each slice is 0.625 mm. The table index is 5 cm. What is the pitch?

A. 0.73  
B. 0.86  
C. 1.15  
D. 1.25  
E. 1.50

beam width = 64 × 0.625 = 40 mm  
pitch = 5 cm / 4 cm = 1.25

**Question 7**

True or false? If a CT display is set at a window width of 100 HU and a window level of 50 HU,

A. HU value of water changes to 50.  
B. white matter looks gray.  
C. fat looks black.  
D. water looks black.  
E. bone looks white.  
F. lungs look white.

**History of CT**

- 1967: Hounsfield: initial CT idea, first CT scanner: γ-rays, 9 d acquisition and 2.5 h processing
- 1971: first clinical brain CT scanner (x-rays, 20 m acquisition)
- 1972: Cormack: filtered backprojection algorithm for CT reconstruction
- 1972-74: four generations of CT scanners

**1st, 2nd, 3rd and 4th generation CT**

- I: stationary ring detector and rotating x-ray tube
- II:  
- III:  
- IV:  

stationary ring detector and rotating x-ray tube
History of CT

- 1974: first clinical WB CT scanner
- mid-1980’s: EBCT (50 ms) for cardiac cine CT

History of CT

- 1989: spiral/helical CT (bed translates as gantry rotates) for fast volume CT
- 1998: multi-detector CT (more than one row detectors) for CT fluoro, CT angio, virtual endoscopy
- 2003: 64 slice CT
  - 64 rows of 0.625 mm detector → scan an organ within a single breath hold
  - maximum table speed: 87 mm/s
  - fastest gantry rotation: 3 rotations/sec

Major components of a CT scanner

- x-ray tube with longer operation time (upto 100 sec), more intensive current (upto 700 mA), and stable at high speed rotation (upto 3 rot/sec).
- Filtration reduces low-energy components to reduce patient dose and beam hardening effect.
- Collimation defines the shape of the x-ray beam (width and angle).
Major components of a CT scanner

- x-ray detectors: solid-state material or CsI, usually multiple rows
- A bow-tie filter is inserted between patient and detector to reduce beam hardening effect.
- slip rings (stationary and rotating) to remove electric and data cables

Slip rings

- No cables are long enough for continuous gantry rotation. How to input power and output data?
- 2 conductive rings (stationary and rotating) arranged in disk or cylinder
- electrical and data cables connected to the stationary ring
- brushes on the rotating ring gliding in contact grooves on the stationary ring to supply electricity

Slip rings

- data acquisition system (DAS)
  - through the brushes as the power supply
  - lens and photodiodes on the stationary ring to receive the CT data from the rotating detector (optoelectronic process) at a transmission speed of 5 Gbits/s

Dual source CT for cardiac CT

- 180° CT data acquisition ➔ each tube rotating 90°
- It needs only 1/4 of the scan time and 50% of the radiation dose.
- A cardiac study can be finished in a single heartbeat.
- adaptive ECG for real-time heartbeat-controlled dose modulation
- adaptive table speed for reduced acquisition time and dose at higher heart rates
Dual energy CT

- The two x-ray tubes emit x-rays at different energies, e.g. 80 kVp and 140 kVp.
- To differentiate materials, such as bone and iodine in angiography, based on different attenuation at different x-ray energies.
  - e.g. at 80 kVp
    - \( \mu_{\text{bone}} = 550 \text{ HU} \)
    - \( \mu_{i} = 425 \text{ HU} \)
  - at 140 kVp
    - \( \mu_{\text{bone}} = 400 \text{ HU} \)
    - \( \mu_{i} = 250 \text{ HU} \)

256 or more slice CT

- e.g. Toshiba AquilionOne: 320 0.5 mm rows, Philips Brilliance iCT: 256 slices, 0.25 s/360°
- \( z > 13 \text{ cm} \) → One rotation covers a heart or brain without table translation.
- Lower radiation dose thanks to no helical overlap.
- Mainly for cardiac CTA.
- In principle, large angle cone-beam acquisition w/o helical scan cannot provide sufficient data for image reconstruction.

Flat panel CT detector

- \( 40 \times 30 \text{ cm aSi-CsI detector with } 2000 \times 1500 \text{ pixels and } \sim 200 \mu\text{m pixel size} \)
- FOV: 14 cm (A) × 45 cm (T) → One rotation covers the whole heart.
- 12 sec/rotation
- Modest contrast-to-noise ratio due to the scatter → degraded low-contrast detectability (5 HU)
- Poor susceptibility to radiation damage

New clinical applications of 64 slice CT

- Cardiac CT
  - Ca scoring
  - Coronary CT angiography
- Brain perfusion CT
- CT pulmonary angiography
- CT scanning of multiple body segments with a single bolus injection of contrast for multiple trauma patients
- WB CT angiography
**Cardiac CT**

- acquire data as the heart motion is minimal (usually the end of diastole) to minimize heart motion effect
- submillimeter spatial resolution to see coronary arteries
- fast scan to cover whole heart in one breath hold
- 64 slice CT with 1/3 second rotation time and special protocols

**Prospective ECG-triggering for Ca scoring**

- low mAs axial CT with ECG monitoring
- The minimum time for one CT rotation (360°) is 333 ms while complete CT data can be acquired in an angular range of 180° + ϕ that takes about 200 ms.
- Slice thickness depends on the distance of each table translation and in turn, on the heart rate.

**Prospective ECG-triggering for Ca scoring**

- low radiation dose
- noisy image due to short acquisition and low mAs
- good spatial resolution with thin slices
- poor temporal resolution
Retrospective ECG-gating for coronary artery

- helical CT with ECG signals
- retrospectively select data acquired from diastole period for image reconstruction and discard other data
- data selected from one heart cycle when diastole period is 200 ms or longer
- data selected from multiple heart cycles when the heart beat is too fast that diastole period is shorter than 200 ms and then sum the segmented data to form complete data

Partial data are discarded which wastes dose.

To obtain adequate spatial resolution, the segmented reconstruction requires
- good periodicity of the heart beats
- small pitch (substantial beam overlap) to prevent spatial gaps between the cycles.
- high radiation dose
- good temporal resolution

Brain perfusion CT

- purpose: to evaluate cerebral blood flow (CBF) for suspected ischemia and stroke
- repeated CT scans with 80 kVp, 200 mA and 1 sec/rotation to acquire the time-concentration curve of a contrast for each voxel of the brain
- From the CT data, MTT (mean transit time) and CBV (cerebral blood volume) are derived using deconvolution.
- $\text{CBF} = \frac{\text{CBV}}{\text{MTT}}$ for each voxel
Brain perfusion CT

CT → CBV → MTT → CBF

Image artifacts

discrepancies between the reconstructed image and patient, caused by
- metal implants
- beam hardening
- patient motion
- image truncation
- partial volume effect
- image processing
- low- and high-contrast resolution

metal: shading and streaks

metal: streaks and shading
corrected
**Beam hardening for bremsstrahlung x-rays**

- Low-energy x-rays are absorbed more than high-energy x-rays → higher average energy → smaller HU
- Deeper in the body → more beam hardening

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**Reduction of beam hardening**

- A bowtie filter inserted between patient and detector
- Heavier beam filtering
- Software

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**Beam hardening:**
- Dark center for a uniform object (cupping)
- A band between high HU structures
beam hardening: bands and streaks between high HU structures

beam hardening: a band

beam hardening: streaks starting from a high HU structure

patient motion: a line \perp motion direction
patient motion: parallel streaks

patient motion: streaks and shading

truncation: streaks and incomplete contour

truncation: streaks
**Partial volume**

averaging when a voxel contains more than one type of tissue → streaks, shading, etc

- reducing pixel size and slice thickness
- software

partial volume: shading and distorted shape

partial volume: streaks starting from the tissue boundary, difficult to distinguish from beam hardening

corrected

aliasing artifacts due to too few views: streaks
defected detector unit: ring

too few lateral counts: increased mAs

standard filter  bone filter  lung filter

better resolution → worse noise

Images of a 15-mm polyp

colonoscopy  CT 100 mAs

filtered CT 25 mAs  filtered CT 6.3 mAs
**Low contrast resolution**

- For low contrast structures (e.g. several HU higher than the bkg), noise is the dominant factor for the resolution. A low noise level results in better resolution.
- It also depends on the size of the structure. A larger structure is easier to be resolved.

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All spheres have 6 HU above the background but different radii.

240 mAs, 5 mm 240 mAs, 2.5 mm 80 mAs, 5 mm

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200 mAs 400 mAs 600 mAs 1000 mAs

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25 mAs 640 mAs \(\rightarrow\) noise down to 1/5
**Noise level and contrast-to-noise ratio**

- For Poison distribution (e.g. CT), standard deviation of a voxel \( SD = \sqrt{\text{count}} \).
- A noise level is measured by the relative standard deviation.
  \[ \% SD = \frac{SD}{\text{counts}} \times 100\% = \frac{1}{\sqrt{\text{count}}} \times 100\% \]
- Low contrast resolution depends on contrast-to-noise ratio (CNR). A larger CNR means better low contrast resolution.
  \[ \text{CNR} = \frac{(HU_1 - HU_2)}{SD} \]
  e.g. for \( HU_1 = 82 \), \( HU_2 = 68 \), and \( SD = 3.5 \)
  \[ \text{CNR} = \frac{82 - 68}{3.5} = 4 \]

**Noise also depends on noise power spectrum**

- same %SD but different NPS
- more noise at low f
- more noise at high f

**Low contrast resolution depends on NPS**

- same %SD
- different NPS: more noise at high f → better resolution

**High contrast resolution**

Because the object contrast is fairly high, spatial resolution dominates the imaging resolution that is determined by

- focal spot size
- detector width
- slice thickness
- pixel size
- reconstruction filter
Question 8

Heavier filtration of the pre-patient x-ray beam may
A. increase patient dose.
B. increase image contrast.
C. reduce partial volume effect.
D. reduce beam hardening effect.

Question 9

The visibility of low-contrast structures in CT images may improve with increase of
A. filtration.
B. mAs.
C. kVp.
D. image matrix size.
**Question 10**

Ring artifacts in CT are most likely caused by

A. kVp drift.
B. a faulty detector.
C. patient motion.
D. varying mA.

**Question 11**

The HU depends on all the following EXCEPT

A. beam hardening.
B. mA.
C. partial volume effects.
D. kVp.

**Question 12**

The visibility of high-contrast structures in CT images most likely improves with increases of

A. mAs.
B. pitch.
C. image matrix size.
D. slice thickness.
E. kVp.

**Question 13**

Partial volume effects in CT can be reduced if

A. slice thickness increases.
B. scan time increases.
C. image matrix size increases.
D. focal spot size increases.
**Question 14**

In helical CT, all of the following apply EXCEPT

A. slip ring is required.
B. helical CT cannot be performed with bowtie filters.
C. higher x-ray tube heat capacity is needed.
D. partial volume effects will increase.

**Question 15**

Anode heat loading on a CT x-ray tube increases with all of the following EXCEPT

A. kVp.
B. mA.
C. scan time.
D. slice thickness.

**Question 16**

CT transaxial (in-plane or x-y plane) spatial resolution improves with increase of

A. focal spot size.
B. kV and mA.
C. scan time.
D. image matrix size.

**Question 17**

Which of the following is correct for retrospective ECG-triggering CT cardiac studies?

A. Relatively large pitch number is used.
B. Temporal resolution is relatively good.
C. Only the data acquired during one heart cycle can be used in image reconstruction.
D. Patient dose is relatively low.
**Question 18**

Which CT reconstruction parameters result in the best quality for reformation to sagittal and coronal images?

A. 10 mm thick slice, 10 mm increment  
B. 3 mm thick slice, 3 mm increment  
C. 3 mm thick slice, 2 mm increment  
D. 0.5 mm thick slice, 0.5 mm increment  
E. 0.5 mm thick slice, 0.25 mm increment

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**Radiation dose**

- absorbed dose: energy deposited on a unit mass of absorber. This is a pure physical concept  
  \[ 1 \text{ rad} = 100 \text{ erg/g}, \ 1 \text{ Gy} = 1 \text{ joule/kg} \]  
  \[ \Rightarrow 1 \text{ Gy} = 100 \text{ rad} \]
- dose-equivalent: effectiveness in damaging cells (physical + biological effects)  
  \[ 1 \text{ Sv} = 100 \text{ rem} \]
- for x, y, c: 1 rem = 1 rad, 1 Sv = 1 Gy

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**Radiation exposure**

- total charge deposited on a unit mass of air, mainly used for x-rays  
- can be directly measured using an ion chamber  
- 1 R = 2.58 \times 10^{-4} \text{ C/kg}  
- 1 R is converted to 0.869 rad or 8.69 mGy of skin dose.

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**Sources of radiation (1982): 4.6 mSv/person/yr**

- natural background: 3.6 mSv/person/yr (78% of all dose)  
  - radon gas at home: 2.6 mSv/person/yr  
  - U-238, Th-232 and others present in soil and construction materials  
  - cosmic radiation from the sky  
  - internal radiation from 40K, 14C and 3H
- man-made: ~ 1 mSv/person/yr (22%)  
  - medical: ~ 0.54 mSv/person/yr (12%)  
  - fallout and consumer products
Sources of radiation (2006): 6.7 mSv/person/yr

- Radon and other natural doses remain the same but drop from 78% down to 52% of all dose.
- Dramatic increase of medical dose from 0.54 to 3.2 mSv/person/yr (48% of all dose)
  - CT dose: ~1.6 mSv/person/yr (50% of medical dose)
  - NM dose: ~0.8 mSv/person/yr (25% of medical dose)
- Dramatic increase of number of CT studies
  
<table>
<thead>
<tr>
<th>Year</th>
<th>NM (M)</th>
<th>CT (M)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1984</td>
<td>6</td>
<td>3</td>
</tr>
<tr>
<td>2006</td>
<td>18</td>
<td>67</td>
</tr>
</tbody>
</table>

Average dose per CT study keeps decreasing.

How to measure CT dose?

- CT dose index (CTDI): the dose for a slice of the phantom
- Dose line product (DLP): the total dose for the scanned volume
- Individual organ dose
- Effective dose for the whole body

CTDI and DLP

- CTDI: dose to a slice of the standard uniform cylinder (d = 16 cm or d = 32 cm)
- DLP: dose to the scanned volume,
  \[ \text{DLP} = \text{CTDI} \times \text{scan length} \]
- These numbers are shown on the CT console, but are NOT the actual patient dose.

Why using the cylinders?

- To mimic the patient
  - d = 16 cm cylinder for children and adult head
  - d = 32 cm cylinder for adult body
- Easy to compare the dose of different kVp, mAs, pitch, and patient size
  - e.g. with 120 kVp, 260 mAs, and pitch 1
  - \[ \text{CTDI}_{\text{vol}} = 43.59 \text{ mGy} \text{ for } d = 16 \text{ cm} \]
  - \[ \text{CTDI}_{\text{vol}} = 22.27 \text{ mGy} \text{ for } d = 32 \text{ cm} \]
- Easy to evaluate a CT scanner
  - e.g. \[ \text{CTDI}_{\text{vol}} < 25 \text{ mGy} \text{ required in ACR accreditation for pediatric abdomen CT} \]
dose of a single slice contributed from the fixed 100 mm interval around the slice
\[ CTDI_{100} = \frac{1}{nW} \int_{-50 \text{mm}}^{50 \text{mm}} D(z)dz = k \cdot Q/nW \]

charge \( Q \) from the ion chamber

\[ CTDI_w \] (weighted dose in a transverse slice)
\[ = 1/3 \text{ of central } CTDI_{100} \]
\[ + 2/3 \text{ of average peripheral } CTDI_{100} \]

\[ CTDI_{vol} \] (average single-slice dose over the total scanned volume of a uniform phantom)
\[ = CTDI_w / \text{pitch} \]

Exposure to primary and scattered x-rays

◆ typical skin dose in CT (exposure to the primary x-ray beam): 30 mGy
◆ typical dose for a person standing 1 m from a CT scanner: 30 \( \mu \)Gy. This is because almost all primary x-rays are absorbed by the detector and successive absorbers and the parson receives only scatter x-rays.
**Dose to individual organs**

The average organ dose is measured using a series of anthropomorphic phantoms composed of tissue-equivalent materials. The results are:

- **40 - 60 mGy** to head in brain CT (critical organ: thyroid)
- **250 mGy** to head in brain perfusion CT
- **10 - 35 mGy** to stomach in abdomen CT
- **5 - 10 mGy** to lungs in chest CT
- **60 mGy** to heart in coronary CTA
- **10 - 30 mGy** to a fetus in mother’s abdominal CT

**Effective dose for whole body**

- **measure the overall radiation effect**
  - effective dose = \( \sum_i (\text{organ dose} \times \text{organ factor}) \)

<table>
<thead>
<tr>
<th>Organ</th>
<th>WE (Sv)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gonads</td>
<td>0.30</td>
</tr>
<tr>
<td>Red marrow</td>
<td>0.32</td>
</tr>
<tr>
<td>Colon</td>
<td>0.12</td>
</tr>
<tr>
<td>Lungs</td>
<td>0.12</td>
</tr>
<tr>
<td>Stomach</td>
<td>0.12</td>
</tr>
<tr>
<td>Bladder</td>
<td>0.05</td>
</tr>
<tr>
<td>Bones</td>
<td>0.05</td>
</tr>
<tr>
<td>Liver</td>
<td>0.05</td>
</tr>
<tr>
<td>Esophagus</td>
<td>0.05</td>
</tr>
<tr>
<td>Thyroid</td>
<td>0.01</td>
</tr>
<tr>
<td>Skin</td>
<td>0.01</td>
</tr>
<tr>
<td>Bone surfaces</td>
<td>0.01</td>
</tr>
<tr>
<td>Remainder</td>
<td>0.05 split equally between adrenal, brain, upper large intestine, small intestine, kidneys, muscle, pancreas, spleen, thymus, and uterus</td>
</tr>
</tbody>
</table>

- **In practice, it is estimated using DLP.**

**From DLP to effective dose (Radiology 2008, 248:995-1003)**

- scan a cylinder (d = 16 or 32 cm) with 120 kVp, 100 mA, 1 sec, pitch=1, 16×1.25 mm
- calculate ED using 3 software (ImPact, CT-Expo, ImpactDose)
- calculate ED/DLP (\( \mu \text{Sv/mGy·cm} \))

<table>
<thead>
<tr>
<th></th>
<th>Head</th>
<th>Cerv. spine</th>
<th>Chest</th>
<th>Abdomen</th>
<th>Pelvic</th>
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<tbody>
<tr>
<td>LightSpeed</td>
<td>2.2</td>
<td>5.6</td>
<td>17</td>
<td>16</td>
<td>19</td>
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<td>Brilliance</td>
<td>2.2</td>
<td>5.1</td>
<td>18</td>
<td>17</td>
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<tr>
<td>Senation</td>
<td>2.2</td>
<td>5.4</td>
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<td>16</td>
<td>18</td>
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<tr>
<td>Acqilion</td>
<td>2.1</td>
<td>5.3</td>
<td>17</td>
<td>16</td>
<td>18</td>
</tr>
<tr>
<td>Average ± s.d.</td>
<td>2.2 ± 0.1</td>
<td>5.4 ± 0.2</td>
<td>17 ± 0.5</td>
<td>16 ± 0.5</td>
<td>19 ± 1</td>
</tr>
</tbody>
</table>

**From DLP to effective dose (Radiology 2008, 248:995-1003)**

- Ed/DLP independent of kVp in head studies but increasing with kVp (80 – 140 kVp) up to 25% in body studies
- good agreement generally between the 3 software

<table>
<thead>
<tr>
<th>Examination</th>
<th>Average (mSv)</th>
<th>Range (mSv)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head</td>
<td>2</td>
<td>0.9-4.0</td>
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<tr>
<td>Neck</td>
<td>3</td>
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<tr>
<td>Chest</td>
<td>7 (1 for low-dose scan)</td>
<td>4.0-18.0</td>
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<td>Chest for PE</td>
<td>15</td>
<td>13-40</td>
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<tr>
<td>Abdomen</td>
<td>8</td>
<td>3.5-25</td>
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<td>Pelvis</td>
<td>6</td>
<td>3.3-10</td>
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<td>Spine</td>
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<td>1.5-10</td>
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<td>Calcium scoring</td>
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<td>1.0-12</td>
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<td>Coronary angio</td>
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<td>5.0-32</td>
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<tr>
<td>3-phase liver</td>
<td>15</td>
<td>…</td>
</tr>
<tr>
<td>Virtual colonoscopy</td>
<td>10</td>
<td>4.0-13.2</td>
</tr>
</tbody>
</table>

**Effective dose of chest radiography**

0.05 – 0.1 mSv

**What factors determine CT dose?**

- kVp (peak voltage of x-ray tube)
- mAs (mA × scan time)
- pitch (= table index/beam width)
- combine mAs and pitch together → effective mAs = mAs/pitch
- slice thickness and pixel size
- patient size and composition
- image reconstruction algorithm

**kVp and dose**

- Dose is proportional to kVp², e.g. when kVp increases from 100 to 120, the dose increases to \((120/100)^2 = 1.44\). Thus it is a magnified effect.
- Lower kVp results in lower dose and better contrast, but markedly higher noise level.
- To partially reduce the noise, mAs is increased, resulting in increased dose. Optimization of kVp and mAs is performed according to the study type, body size, and composition.
- Lower kVp is particularly useful for CTA due to markedly increase of contrast of iodine.
Automated selection for kVp and mAs

- Manual optimization is impractical.
- Automatic optimization of kVp and mAs is available.
- Or you may follow a simple rule, i.e. 100 kVp for chest CT with BMI < 30 or 100 kVp for abdominal CT with BMI < 25.
- An optimized CNR can maintain the diagnostic quality with 25% less dose.

mAs and dose

- Dose is linearly proportional to mAs.
- Counts are also proportional to mAs.
- Noise level is proportional to $1/\sqrt{\text{counts}}$ and hence to $1/\sqrt{\text{mAs}}$.
- E.g. when mAs increases from 200 to 250, the dose increases to $250/200 = 1.25$ but the noise is down to $\sqrt{200/250} = 0.89$.

Automatic tube current modulation

- Set a noise level and then mA varies automatically with body shape and composition (geometry modulation) or cardiac cycle (time modulation).
- It can reduce dose by 40% to 50%.

If mA doubles, dose will double but noise will decrease by 30% ($1/\sqrt{2}$).

50 mA, 18% 100 mA, 13% 160 mA, 11%
**Body shape modulation**

- z modulation from scout scan, e.g. chest vs pelvis
- angular or x-y modulation, e.g. AP vs lateral directions of the shoulder

**Time modulation using ECG**

- full mA during diastole which is most likely to produce best image quality due to the relatively slow motion of the heart
- reduced mA (~ 50% of the full mA) during systole because the data are most likely to discard

**Pitch and dose**

- Dose is inversely proportional to pitch.
- Larger pitch means faster scan and hence results in lower dose but more noise. E.g. as pitch increases from 1 to 1.3, the dose is down to $1/1.3 = 0.77$.
- shorter scan length $\rightarrow$ less dose

**Slice thickness and dose**

For thinner slices, you have 2 options.

- fixed noise level
  It requires more total counts (due to more slices) and hence higher mAs is needed, resulting in higher dose.
- fixed dose
  Same mAs, hence same total counts, are maintained, resulting in more noise due to fewer counts in each slice.
- As slice thickness is cut to $\frac{1}{2}$, (A) same noise but double dose or (B) same dose but more (1.41) noise
Dose is doubled for the 2.5 mm slice.

$\text{Pixel size and dose}$

For smaller pixels, you have 2 options.

- fixed noise level
  It requires more total counts (due to more pixels) and hence higher mAs is needed, resulting in higher dose.

- fixed dose
  Same mAs, hence same total counts, are maintained, resulting in more noise due to fewer counts in each pixel.

- As pixel size is cut to \( \frac{1}{2} \), (A) same noise level but 4 times dose or (B) same dose but double noise level
**Patient size and dose**

- 2 options: (1) fixed techniques (kVp, mAs, pitch etc.) or (2) fixed noise level
- Same techniques emit same amount of x-rays $N_0$. The count increases a lot for a small patient which is clinically unnecessary but the dose remains almost same. **This is a bad choice.**

  e.g. 80 keV x-rays irradiate two water cylinders with the same $N_0$. The phantom absorbs $N_a$ and allows $N$ to penetrate. Thus $N_a$ is proportional to the dose and $N$ is the counts.

  for $d = 16$ cm, $N_a(16) = 0.947 N_0$ and $N(16) = 0.0527 N_0$
  for $d = 32$ cm, $N_a(32) = 0.9972 N_0$ and $N(32) = 0.00278 N_0$
  Thus $N(16)/N(32) = 19$ while $N_a(16)/N_a(32) = 95\%$.

**Patient size and dose**

- Fixed noise level requires same counts $N$. The dose decreases a lot for a small patient. **This is a good choice.**

  e.g. 80 keV x-rays irradiate two water cylinders with different $N_0$ but same $N$.
  for $d = 16$ cm, $N = 0.0527 N_0(16)$
  for $d = 32$ cm, $N = 0.00278 N_0(32)$
  $N_0(16) = 0.0528 N_0(32)$
  $N_a(16) = 0.947 N_0(32)$ and $N_a(32) = 0.9972 N_0(32)$
  Thus $N_a(16)/N_a(32) = 5\%$

**Factors for image noise**

- % SD proportional to $1/\sqrt{\text{mAs}}$
- % SD proportional to $1/\text{kVp}$
- % SD proportional to $1/\text{slice thickness}$
- % SD proportional to $1/\sqrt{\text{dose}}$
- ACR CT accreditation requires $-7 \text{ HU} < \text{uniform water} < 7 \text{ HU}$.

**Radiation dose in pediatric CT**

- Decreased radiation dose for children is mandated because of
  - more radiosensitive due to more dividing cells
  - longer lifetime risk for radiation induced cancers
- same image quality achievable with much less radiation dose
- Overexposure carries penalty in film radiography but provides better image quality in CT with little diagnostic gain or poor benefit-risk ratio.
**Weight-based mA**

<table>
<thead>
<tr>
<th>weight (lb)</th>
<th>mA for chest</th>
<th>mA for abdomen</th>
</tr>
</thead>
<tbody>
<tr>
<td>10-19</td>
<td>40</td>
<td>60</td>
</tr>
<tr>
<td>20-39</td>
<td>50</td>
<td>70</td>
</tr>
<tr>
<td>40-59</td>
<td>60</td>
<td>80</td>
</tr>
<tr>
<td>60-79</td>
<td>70</td>
<td>100</td>
</tr>
<tr>
<td>80-99</td>
<td>80</td>
<td>120</td>
</tr>
<tr>
<td>100-150</td>
<td>100-120</td>
<td>140-150</td>
</tr>
<tr>
<td>&gt;150</td>
<td>&gt;140</td>
<td>&gt;170</td>
</tr>
</tbody>
</table>

**Diameter-based mAs**

<table>
<thead>
<tr>
<th>abdominal diameter (cm)</th>
<th>mAs reduction factor</th>
<th>% dose reduction</th>
</tr>
</thead>
<tbody>
<tr>
<td>12</td>
<td>0.028</td>
<td>95</td>
</tr>
<tr>
<td>14</td>
<td>0.043</td>
<td>93</td>
</tr>
<tr>
<td>16</td>
<td>0.072</td>
<td>88</td>
</tr>
<tr>
<td>18</td>
<td>0.124</td>
<td>81</td>
</tr>
<tr>
<td>20</td>
<td>0.196</td>
<td>72</td>
</tr>
<tr>
<td>22</td>
<td>0.304</td>
<td>60</td>
</tr>
<tr>
<td>24</td>
<td>0.457</td>
<td>44</td>
</tr>
<tr>
<td>26</td>
<td>0.678</td>
<td>26</td>
</tr>
<tr>
<td>28</td>
<td>1.0</td>
<td></td>
</tr>
</tbody>
</table>

**Fetal dose**

The policy statement of American College of Ob/Gyn:

“Women should be counseled that x-ray exposure from a single diagnostic procedure does not result in harmful fetal effects. Specifically, exposure to less than 50 mGy has not been associated with an increase in fetal anomalies or pregnancy loss.”

**CT dose to fetus**

<table>
<thead>
<tr>
<th>head CT</th>
<th>0 (mSv)</th>
</tr>
</thead>
<tbody>
<tr>
<td>chest CT</td>
<td></td>
</tr>
<tr>
<td>routine standard</td>
<td>0.2</td>
</tr>
<tr>
<td>pulmonary embolus standard</td>
<td>0.2</td>
</tr>
<tr>
<td>CT cardiac angiography</td>
<td>0.1</td>
</tr>
<tr>
<td>routine abdomen</td>
<td>4</td>
</tr>
<tr>
<td>routine abdomen/pelvis</td>
<td>25</td>
</tr>
<tr>
<td>CT angiography of aorta</td>
<td>34</td>
</tr>
<tr>
<td>abdomen/pelvis (stone protocol)</td>
<td>10</td>
</tr>
</tbody>
</table>
Iterative reconstruction and dose

- Two reconstruction algorithm in CT:
  1. filtered backprojection algorithm (FBP)
  2. iterative algorithm
- FBP is fast but approximate. Only when the acquired data are noise-free and have unlimited resolution, FBP produces a perfect image. It produces a lot of artifacts as noise level is high.
- Iterative algorithm can reduce noise by 40% so the dose is reduced by 40%. It was too slow but becomes fast enough to be clinically acceptable.

Iterative reconstruction and dose

- Iterative reconstruction uses ‘trial-and-error’ algorithm with a ‘correction loop’.
- It contains several iterations. The initially guessed image is compared to the acquired data and is then corrected. The updated image is compared to the acquired data again and is corrected. After several repetition, the image is gradually close to the true image.

Question 19

Which CT study has the highest individual organ dose?
A. adult head cerebrum with and without contrast
B. adult routine abdomen
C. adult routine chest
D. coronary CTA

Radiation effect is cumulated so the number of scans for a patient matters.
Question 20
Which CT study has the highest effective dose?
A. adult head cerebrum
B. adult routine abdomen
C. adult routine chest
D. coronary CTA
E. calcium scoring

Question 21
If the pixel size reduces to half but the noise level remains the same in a CT study,
A. patient dose will decrease to half.
B. patient dose will decrease to ¼.
C. patient dose will remain the same
D. patient dose will double.
E. patient dose will increase to 4 times.

Question 22
When kVp increases from 120 to 140, mAs decreases from 300 to 220 and pitch increases from 0.75 to 1, how do dose and noise level change?
1. dose proportional to kVp^2 and mAs/pitch \( \Rightarrow \) 
   \[
   \text{dose}_{\text{new}} / \text{dose}_{\text{old}} = \left(\frac{140}{120}\right)^2 \cdot \left(\frac{220}{300}\right) \cdot (0.75/1) 
   \]
   \[
   = 1.36/1.36 \cdot 0.75 = 0.75
   \]
2. %SD proportional to 1/kVp and sqrt(pitch/mAs) \( \Rightarrow \) 
   \[
   \%\text{SD}_{\text{new}} / \%\text{SD}_{\text{old}} = \left(\frac{120}{140}\right) \cdot \sqrt{\frac{300}{(220 \cdot 0.75)}} 
   \]
   \[
   = 0.86 \cdot 1.35 = 1.16
   \]
3. Dose decreases to 0.75 and noise increases to 1.16 which satisfies 0.75 = 1/1.16^2.

Question 23
A radiologist would like to decrease the slice thickness for an exam from 5 mm to 2.5 mm while maintaining same noise level. This will cause what change in radiation dose?
A. Radiation dose is doubled.
B. Radiation dose is cut to half.
C. Radiation dose remains same.
D. Radiation dose increases by 40%.
Question 24

<table>
<thead>
<tr>
<th>Beam Width</th>
<th>Slice Thickness</th>
<th>Noise Level</th>
<th>Dose (mGy)</th>
<th>Or</th>
<th>Dose (mGy)</th>
<th>Noise Level</th>
</tr>
</thead>
<tbody>
<tr>
<td>4 \times 2.5 mm</td>
<td>3.2 mm</td>
<td>1.00</td>
<td>11.2</td>
<td>1.00</td>
<td>11.2</td>
<td>1.00</td>
</tr>
<tr>
<td>8 \times 1.25 mm</td>
<td>1.6 mm</td>
<td>1.01</td>
<td>23.8</td>
<td>1.46</td>
<td>11.1</td>
<td>1.98</td>
</tr>
<tr>
<td>16 \times 0.625 mm</td>
<td>0.85 mm</td>
<td>1.02</td>
<td>45.6</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Question 25

A radiologist would like to increase the pixel size from 0.5 mm to 1 mm while maintaining the same radiation dose. How does the noise level change?

A. Noise level will remain the same.
B. Noise level will be cut to half.
C. Noise level will be cut to 1/4.
D. Noise level will increase to 4 times.

Question 26

Which CT scan will deliver the highest radiation dose?

A. 120 kVp, 200 mA, 0.5 s, pitch 0.8
B. 120 kVp, 150 mA, 1.0 s, pitch 1.0
C. 90 kVp, 150 mA, 1.5 s, pitch 1.5
D. 140 kVp, 200 mA, 0.5 s, pitch 1.5
E. 80 kVp, 200 mA, 0.5 s, pitch 1.5

Dose \approx kVp^2 \times mAs/pitch

140^2 \cdot 100/1.5 = 1370000, \ 120^2 \cdot 150 = 2160000

Question 27

Assume that a scan of one CT slice produces 30 mGy to a slice of tissue. Which of the following statements is wrong when 10 contiguous slices are scanned with the same technique?

A. The risk to patient increases.
B. Scatter contributed to the dose.
C. The radiation dose to this slice of tissue increases approximately by 10 times.
D. The total dose increases approximately by 10 times.
**Question 28**

When the pitch increases in helical CT, which of the following is correct?

- A. improved axial resolution
- B. reduced image noise
- C. reduced beam hardening effect
- D. reduced patient dose

**Question 29**

Image noise in CT will be reduced when

- A. the patient size increases.
- B. the slice thickness decreases.
- C. the pixel size increases.
- D. the scan time decreases.