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Master of Science EPF-ETH degree in Nuclear Engineering
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7 CT

Acknowledgements: Sue Edyvean (HPA – UK) for many of her slides
Radiography

• Origin of the signal?
  – What are
    • The relevant parameters
    • The major problems

• What kind of information can be detected?
  – Typical pixel size?

• What kind of risk?
Objectives

• Explain the principle of CT image acquisition

• Describe the challenges of image quality assessment when dealing with CT imaging

• Describe the principle of DECT
Development of the CT Scanner

- 1st clinical CT scanner
  - developed at EMI Medical by Godfrey Hounsfield
  - prototype installed at Atkinson Morley’s Hospital, Wimbledon, London
  - 1st clinical scan 1st October 1971
What is CT?

• Computed Tomography
  – Computed – determined by mathematical methods
  – Tomos – section, graphia – to write or draw

• An x-ray device capable of cross sectional imaging
  – creates images of ‘slices’ through the patient
What challenges?
Modern multi-slice scanners

- 1998 (4 slice); 2004 (64 slice)
- 2007 (320 detector rows)
- 2012 Sub-mSv chest acquisitions
Construction of a CT scanner

• What’s under the covers?
The CT scanner

- **Cables**
- **Tube**
- **X-ray fan beam**
- **Aperture / bore**
- **Detectors ~ 1000**
- **Flat filter**
- **'Bow-tie' filter**
CT scanner

- The X-ray beam is often referred to as a fan beam where the beam width along the longitudinal axis is small.
- For multi-slice scanners where the longitudinal beam width is no longer small, the X-ray beam is often referred to as ‘cone beam’.
The CT scanner - demonstrating a fan beam

X-ray fan beam

1 - ~ 4 slice
10 to 20 mm beam

Picture courtesy of K. Gelijns, Leiden
The CT scanner – Multi-slice

X - ray fan beam → cone beam

Wider beam and more slices

Z- axis length typically ~ 40 mm
(from 10 mm up to 160 mm)

Picture courtesy of K. Gelijns, Leiden
The CT scanner – Multi-slice

Aquilion 64
64 X 0.5 mm

1, 4, 16, 64, 128, 320 detector rows
min size of detector element ~ 0.5, 0.6 mm

Picture courtesy of K. Gelijns, Leiden
Axial scanning – ‘step and shoot’

– Also known as sequential scanning
X-ray attenuation

- Each detector cell measures: $\frac{N}{N_0}$
Acquisition of the data
Acquisition of the data

Profile as a function of X-ray tube rotation angle

Sinogram

X-ray tube angle

Attenuation profile
Simple back-projection

• Problem

We have the raw data: sinogram

We are interested in $\mu$ values of each voxel
Image reconstruction

• Problem

We have the raw data: 
*sinogram*

We are interested in \( \mu \) values of each voxel

Analytical methods are not feasible
The standard way is to use the Filtered Back Projection (FBP) method
Iterative reconstructions are now standard options
FBP

Each line is an attenuation profile at a given X-ray tube angle.

original

One attenuation profile is considered
original

sinogram

5° are considered
FBP

original

sinogram

60° are considered
FBP

original

sinogram

180° are considered
Reconstruction filter (kernel)
Effect of the reconstruction filter

- Influence on
  - Spatial resolution and image noise

Smooth

Bone
Image representation

• Hounsfield units
  – Linear attenuation coefficient relative to water

\[
NCT(\text{tissue}) = 1000 \frac{\mu_{\text{tissue}} - \mu_{\text{water}}}{\mu_{\text{water}}} \quad \text{(unit: } / . \text{)}
\]

– Typical values
  • NCT(water) = 0 \text{ (by definition)}
  • NCT(air) = -1'000
  • NCT(bone) = 250 to 2'000
Influence of beam energy on CT

Ratio of attenuation coefficients

Energy (keV)

Soft tiss / $\mu_{\text{wat}}$
Fat / $\mu_{\text{wat}}$
Bone / $\mu_{\text{wat}}$
Density and CT #

<table>
<thead>
<tr>
<th>Tissu</th>
<th>Graisse</th>
<th>Sang</th>
<th>Cerveau</th>
<th>Coeur</th>
<th>Rein</th>
<th>Muscle</th>
<th>Pancréas</th>
</tr>
</thead>
<tbody>
<tr>
<td>Densité électronique relative</td>
<td>0.925</td>
<td>1.054</td>
<td>1.033</td>
<td>1.024</td>
<td>1.045</td>
<td>1.054</td>
<td>1.048</td>
</tr>
</tbody>
</table>
Exercices

• The CT number of a muscle is 40 HU. Considering that at the effective energy used the linear attenuation coefficient of water is equal to : 0.19 cm⁻¹

• Calculate the linear attenuation coefficient of the muscle

• How would change the CT number of the muscle if we would reduce the X-ray beam energy by 20 % ?
  – Justify why
Image representation

- Windowing
- Eye dynamic range
  - env. 200 grey levels can be resolved
  - Impossible to resolve more than 4000 grey levels available
- Limitation of the range of grey level one analyses

CT #

C/W : 50, 400

Mediastinal window

These pixels are white

These pixels are black
Image representation

• Windowing

• Eye dynamic range
  – env. 200 grey levels can be resolved
  – Impossible to resolve more than 4000 grey levels available

• Limitation of the range of grey level one analyses
Helical (spiral) scanning

- Continuous gantry rotation + continuous table feed
- Scan data traces a helical path - or ‘spiral’ - around patient
  - data used to form axial images (interpolation of attenuation profiles)
Pitch definition

\[ \text{Pitch} = \frac{\text{Table advance per tube rotation}}{\text{X-ray beam nominal collimation}} \]

- \( p = 1 \) : exact coverage
- \( p < 1 \) : over coverage
- \( p > 1 \) : partial coverage

(loss of axial resolution, not that much of a problem with MDCT)
Advantages and limits of helical SDCT

• Helical single detector row CT advantage
  – Volume acquisition
  – One can get slice at all position along the examined region

• Longitudinal resolution and acquisition speed tradeoff
  – The higher the pitch the poorer the sampling of the data
  – Pitch values in general not above 1.5

  – Example
    • Pitch of 1.5, rotation time 1 s, nominal slice thickness 7 mm
      – In 1 s one scans : 7 x 1.5 mm → 10.5 mm – in 15 s : 157 mm
        » If rotation time = 0.5 s → 315 mm
        » Actual slice thickness : about 9 mm

    • Pitch of 1.0, rotation time 1 s, nominal slice thickness 2.5 mm
      – In 1 s one scans : 2.5 x 1.0 mm → 2.5 mm – in 15 s : 37.5 mm
        » If rotation time = 0.5 s → 75 mm
        » Actual slice thickness : about 2.5 mm
MDCT : Multi-detector row CT

- **Isotropic volume acquisition**
  - Voxels of cubic shape (sub-millimeters)
  - Slice reformatting (MPR Multi-Plane Reformatting)
    - From transverse slice
      - Sagittal, coronal
  - Large volume acquisition in a single breath hold (< 15 s) with an excellent longitudinal resolution
Clinical performance of four-section CT in sequential scan mode. Follow-up images in a patient after surgical removal of pituitary tumor. Left: 4-mm-thick image with standard head kernel for soft-tissue evaluation. Right: 1-mm-thick image with bone kernel for bone evaluation. Both images were generated from the same scan data (four sections at 1-mm collimation).
Exercise

• The acquisition parameters for a chest acquisition are the following:
  – Detector configuration: 64x0.625
  – Rotation time: 0.6 s
  – Pitch factor: 1.4
  – Scan length: 35 cm

What is the acquisition time?
Parameters and image quality

• In plane spatial resolution
  – Pixel size
    • Image 512x512 en general
      – Depends on the reconstructed field of view
  – Reconstruction filter

• Longitudinal spatial resolution
  – Reconstructed slice thickness
  – Detector configuration
    • 16x1.25 16x0.625 for example
  – Pitch value
Parameters and image quality

- How would you increase the contrast of tissues in CT?

- How would you increase the in-plane spatial resolution?
  - What would be the impact?

- How would you increase the longitudinal spatial resolution?
  - What would be the impact?
Exercices

- Two categories of material (classes A et B) are scanned at 80 kV and then at 140 kV. How would you analyze the following behavior?
Exercices

• What is roughly the pixel size of a radiograph ?

• What is roughly the pixel size of a CT image?
  – What is the higher frequency one can expect to get?

• Propose a solution to increase the data sampling in CT

• How would you reduce the acquisition time
  – For cardiac application for example
Dose distribution in Scan Plane

• In CT whole body irradiated

Projection radiography

X-ray CT
Dose distribution in CT

Dose profile from one slice
Dose indicators in CT: CTDI and DLP

• Computed Tomography Dose Index
  – Introduced in the seventies by the FDA
    • Can be measured in air or in phantom
      – Normalized Computed Tomography Dose Index in air ($n_{CTDI_{air}}$)
        » Tube output in mGy per mAs at isocentre

The radiation dose profile along a line perpendicular to the plane of a single axial CT scan shows a peak where the primary beam slices through the CTDI phantom. The tails of the dose profile are caused by scattered radiation. The integral of the area under the curve is normalized to the nominal beam width $NT$ to determine the CTDI. A $CTDI_{100}$ value is obtained if integration limits of ±50 mm are used.

CTDI evaluation

Equipment typically used to measure CTDI\textsubscript{100} includes an integrating electrometer (black arrow), a 100-mm-long CTDI ionization chamber (white arrow), and a CTDI phantom made of polymethylmethacrylate (arrowhead).

All CTDI measurements made with a pencil ionization chamber are performed with a stationary patient table.

\[ nCTDI_w = \frac{1}{3} nCTDI_{centre} + \frac{2}{3} nCTDI_{periphery} \]

\[ nCTDI_{vol} = nCTDI_w / \text{pitch} \]

Dose length product (DLP)

- Dose descriptor used to indicate overall exposure for CT
- DLP = CTDI$_{vol}$ × scan length
- Relates to risk
### How to calculate effective dose (2)

#### DLP conversion to effective dose (Adults)

<table>
<thead>
<tr>
<th>Region of body</th>
<th>Normalized effective dose $E_{\text{dlp}}$ (mSv mGy$^{-1}$ cm$^{-1}$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head</td>
<td>0.0023</td>
</tr>
<tr>
<td>Neck</td>
<td>0.0054</td>
</tr>
<tr>
<td><strong>Chest</strong></td>
<td><strong>0.017</strong></td>
</tr>
<tr>
<td>Abdomen</td>
<td>0.015</td>
</tr>
<tr>
<td>Pelvis</td>
<td>0.019</td>
</tr>
</tbody>
</table>
Mass Attenuation Coefficients as a Function of Energy

<table>
<thead>
<tr>
<th>Energy (keV)</th>
<th>Mass attenuation coefficient (cm²/g)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Water</td>
</tr>
<tr>
<td>30</td>
<td>0.371</td>
</tr>
<tr>
<td>40</td>
<td>0.267</td>
</tr>
<tr>
<td>60</td>
<td>0.205</td>
</tr>
<tr>
<td>80</td>
<td>0.183</td>
</tr>
<tr>
<td>100</td>
<td>0.171</td>
</tr>
<tr>
<td>150</td>
<td>0.150</td>
</tr>
</tbody>
</table>

Willi A. Kalender, Ph.D.
New Horizons Session NH8, ECR, Vienna, March 4, 2011
DECT - intuition

It should be possible to use how a CT number changes with kVp to learn more about the pixel composition.
Fig. 4—Clinical example dataset obtained on dual-source CT scanner using 0.4-mm stannum filter at 140 kVp and 71 mAs and 100 kVp and 69 mAs with overall CT dose index of 5.7 mGy. Images were generated with Syngo dual-energy software (version VE32B, Siemens Healthcare) of 72-year-old woman with liver metastasis from colorectal cancer.

A. Image acquired at 140 kVp using stannum filter.
B. Image acquired at 100 kVp.
C. Quasi monoenergetic image extrapolated to 140 kEV.
D. Optimum contrast image after “sigmoidal blending.”
E. Algorithm differentiates iodine (blue) from calcium (red).
F. Angiographic image after bone removal.
G. Algorithm quantifies iodine by color-coding iodine in orange.
H. Virtual unenhanced image after iodine subtraction.
I. Fusion of color-coded iodine image and unenhanced image.
Conclusion

• Present situation of CT
  – Sub-millimeter 3D imaging procedure
  – Cardiac → between 70 and 140 ms acquisition time

• 10% radiological examinations → 70 % exposure dose

• FBP → Iterative reconstruction
  – Non linear process → image quality assessment becomes challenging

• From detection to characterization
  – DECT is still finding its clinical applications