Lecture 10:
Computerised X-Ray Tomography II

Contents

Visualisation of CT Data

After measuring CT data and reconstruction (e.g. by filtered back-projection) an approximate density function u inside the investigated volume has been obtained. The function u is scaled in Hounsfield units (HF).

To display these data for inspection, two basic approaches are possible:

- Arbitrary 2-D sections (also tilted ones) can be displayed.
- 3-D representations can be computed.

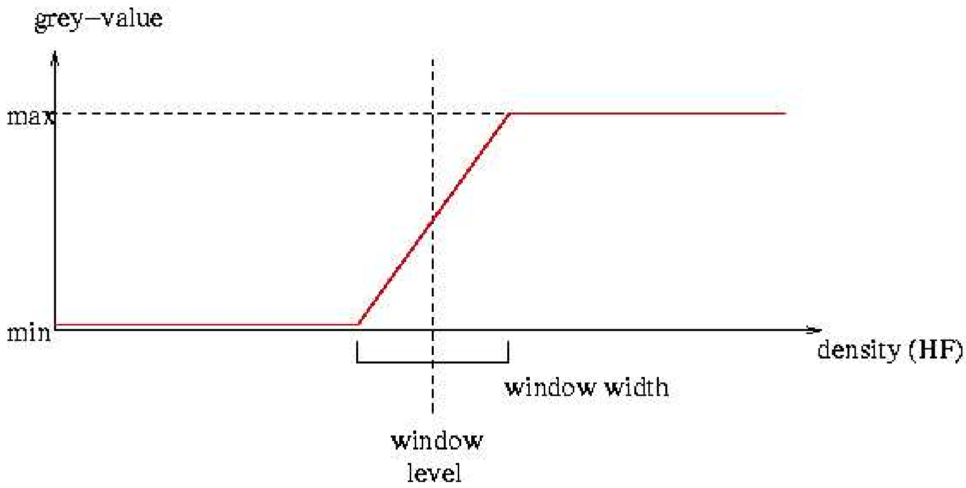
Visualisation of CT Data (2)

2-D Visualisation

Often, 2-D visualisations will display the entire reconstructed data array as a stack of 2-D sections. Moreover, sections along arbitrary planes can be displayed.

Windowing: To improve the contrast for the interesting structures, usually not the whole range of possible densities is represented but only a subinterval (*window*).

A window is characterised by its central point (*window level*) and interval length (*window width*), both in HF.



Visualisation of CT Data (3)

3-D Visualisation

- Advantage: Spatial situation can be represented adequately.
- Disadvantage: Occlusions and inclusions of structures prevent simultaneous visualisation of the entire data array.

A typical approach to 3-D visualisation is to choose a threshold, and consider voxels with densities above the threshold as opaque, such with lower densities as transparent. One therefore has to render *level surfaces* of the reconstructed density function.

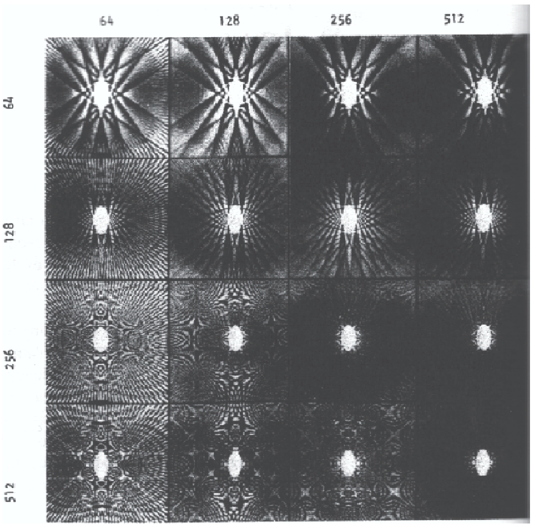
Often it will be necessary to perform some denoising before using a 3-D visualisation. Denoising is a topic of image processing lectures.

Artifacts and Noise in CT

Discretisation Artifacts

We have described the basic principles of tomographic reconstruction in the continuous case. In reality, only finitely many data values are measured (discretisation).

Discretisation leads to *aliasing* and *ghost* artifacts in reconstructed images.



Aliasing artifacts in reconstructing an elliptical structure. Angular resolution (number of projection) increases from left to right, pixel resolution of projections from top to bottom. *(Kak/Slaney 2001)*

Ghosts

- Ghosts are density functions for which all projections measured within the CT setting used are zero.

Aliasing

- By measuring a signal with finite resolution (*sampling*), only a limited range of frequencies can adequately be mapped (*sampling theorem*).
- Frequencies to be represented adequately have to be less or equal to *Nyquist's frequency*, i.e. half the sampling frequency.
- Aliasing means that higher frequencies are "wrapped" into the low-frequency range by sampling.

Aliasing in Projections

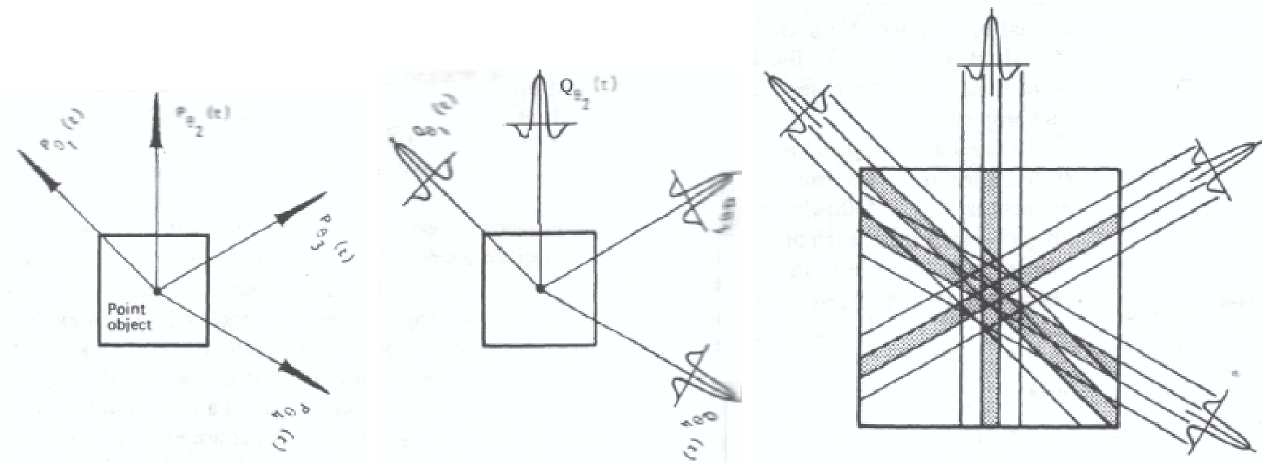
- Structures measured in CT in fact contain higher frequencies. Since projections are measured with limited resolution, they already contain aliasing.
- This is the case particularly in the vicinity of sharp edges. By aliasing, oscillatory artifacts appear.

Artifacts and Noise in CT (3)

Aliasing in Back-Projection

- Because of the finite number of projections, back-projection leads to star-shaped patterns. These can be considered as an angular aliasing effect.

A rule of thumb says that the number of projections and the number of rays in each projection should be roughly equal.



Right: Angular aliasing leads to star-shaped artifacts. *(Kak/Slaney 2001)*

Artifacts and Noise in CT (4)

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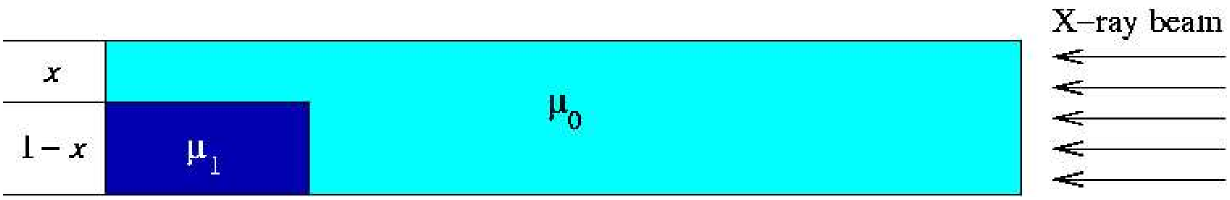
Partial Volume Effect



- By now we assumed that X-rays used for measurement were sharp beams.
- Real X-ray beams have finite width.
- Partial volume effects occur if a beam hits some structure with part of its width.
- In this case, the exponential attenuation law does not hold for the entire beam. Consequentially, the linearity assumption underlying the tomographic reconstruction is violated. Streak-like artifacts are observed.

Top: Partial voluming in a stripe-like beam. *(After P. Joseph.)* **Bottom:** Artifacts due to partial voluming close to the edge of contrast medium in patient's stomach. *(P. Joseph, copied from Epstein 2003)*

Artifacts and Noise in CT (5)

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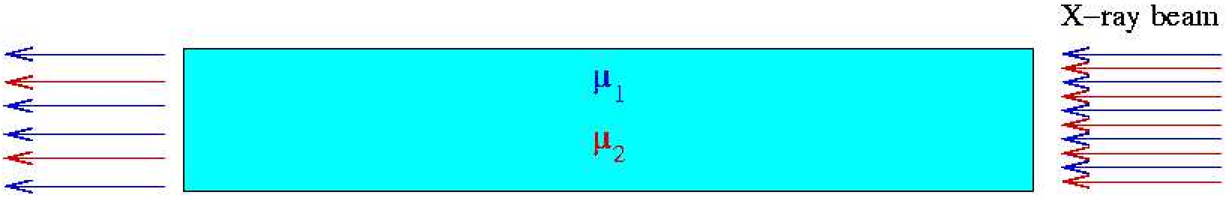
Beam Hardening

- The X-rays emitted by an X-ray tube are not monochromatic but have a continuous spectrum (compare Lecture 7).
- Attenuation coefficients of materials change with wavelengths.
  Typically, attenuation is smaller for shorter wavelengths ("harder" radiation).
  The spectrum of the attenuated beam is therefore shifted towards harder radiation.
- The attenuation of the entire beam does not exactly obey Beer's law:
  Instead, different exponential attenuations take place for different wavelengths.
  This leads again to violations of the linearity assumption.



Artifacts and Noise in CT (6)

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Noise in CT images

Two main sources of noise in CT measurements are

- *Photon Noise*
  - only small number of photons contributes to each single measured value
  - creates stochastic perturbations (Poisson noise)
- *Bad Rays and Bad Views*
  - calibration errors or malfunctions of the X-ray tube or detectors
  - lead to isolated erroneous measurements
  - either bad rays or bad views
  - creates impulse noise
    (for higher generation scanners also more complicated noise types)

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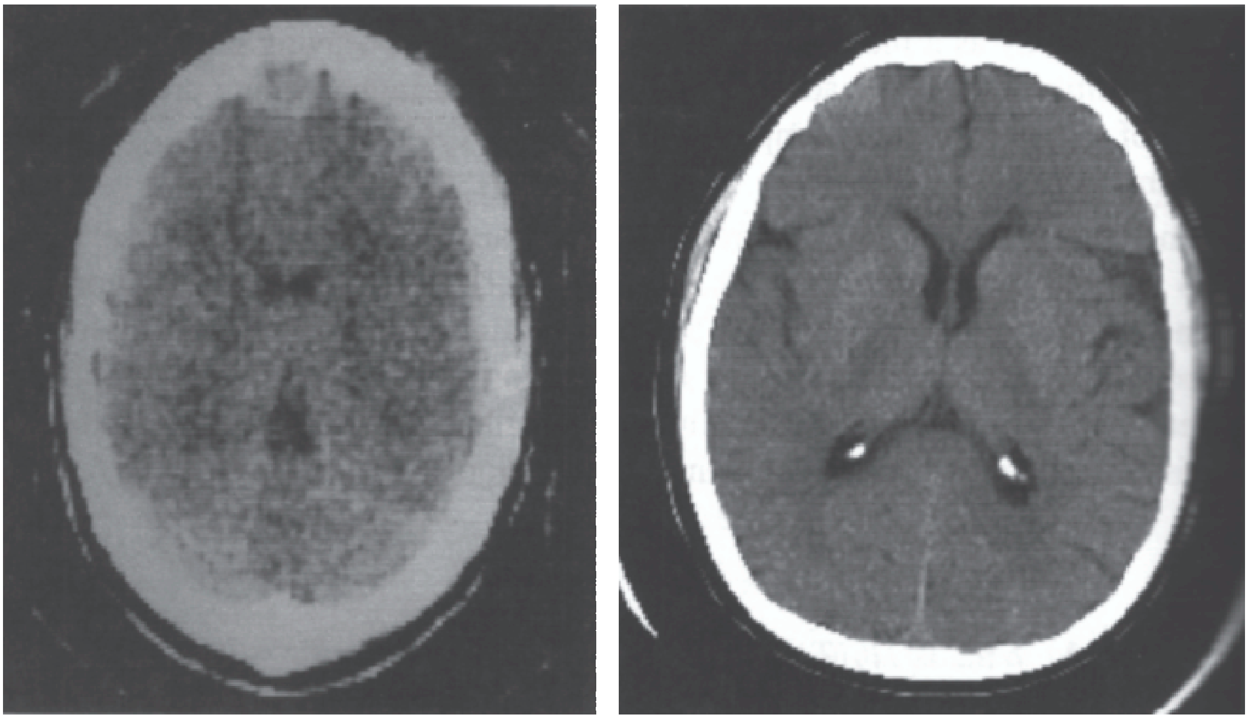
Image Quality in CT



Left: Brain section CT, 1976, reconstructed by filtered backprojection. **Right:** Brain section CT, 2002, reconstructed by filtered backprojection. *(Left image by P. Joseph, right image by D. Hackney, both copied from Epstein 2003)*

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Computerised Emission Tomography

Like in conventional emission radiography (see Lecture 7), also in emission tomography the object itself becomes a source of gamma rays which are detected outside the object.

There are two main methods used today:

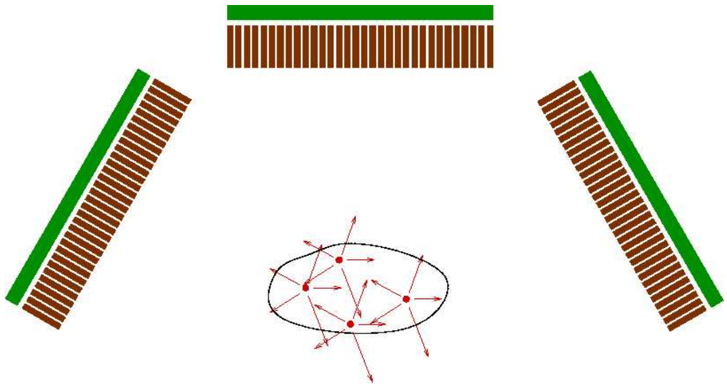
- *Single Photon Emission Computed Tomography (SPECT).*
  Single gamma photons emitted by injected radionuclides are detected.
- *Positron Emission Tomography (PET).*
  Pairs of gamma photons which are simultaneously emitted are detected.

Single Photon Emission Computed Tomography (SPECT)

SPECT is the straightforward tomographic extension of conventional emission radiography (scintigraphy).

A patient is injected radioactive isotopes which propagate in the body and emit gamma radiation. A gamma camera is moved around the body to measure emitted radiation in different directions.

- If the gamma camera uses a parallel-hole collimator, it measures directly projections of the gamma source (thus, radionuclide) distribution.
- Since gamma rays are very hard radiation, attenuation is reduced. However, for improved reconstruction, it has to be taken into account, necessitating complicated attenuation correction methods.



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- The tomographic reconstruction process is analogous to that in transmission CT.
- Image quality is considerably lower than that in transmission tomography:
  - lower resolution
  - stronger influence of scattered photons (simultaneous emission in the entire volume).

Rotating gamma camera for SPECT scanning. *(Webb 1988)*

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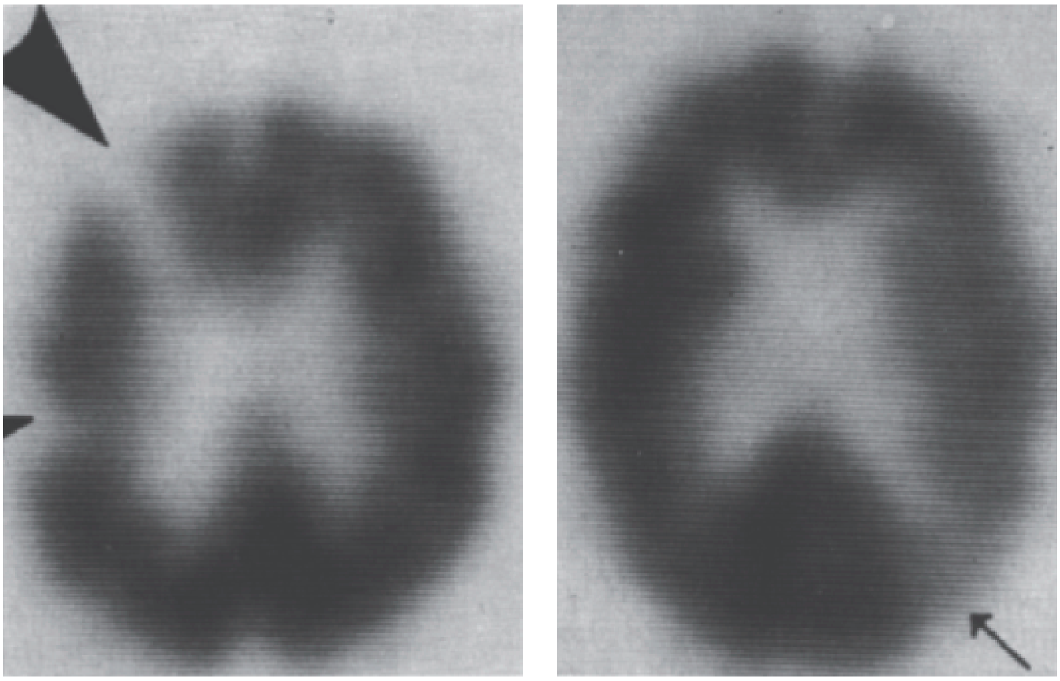
Single Photon Emission Computed Tomography (3)

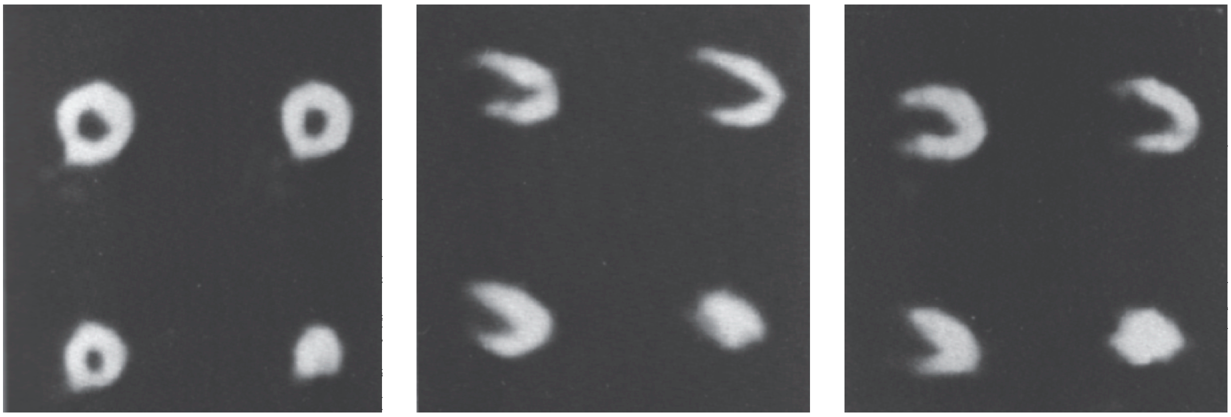
SPECT Examples



SPECT images of human brain (transaxial sections) acquired using Technetium-99. This technique allows to image cortical blood flow. Arrows indicate regions with reduced cortical blood flow, resulting from stroke (left) and Alzheimer's disease (right), resp. *(P. J. Ell, copied from Webb 1988)*

Single Photon Emission Computed Tomography (4)

SPECT Examples



SPECT images of human myocard (sections across three different axes) acquired using Tellur-201 chloride, showing heart-muscle activity. *(P. J. Ell, copied from Webb 1988)*

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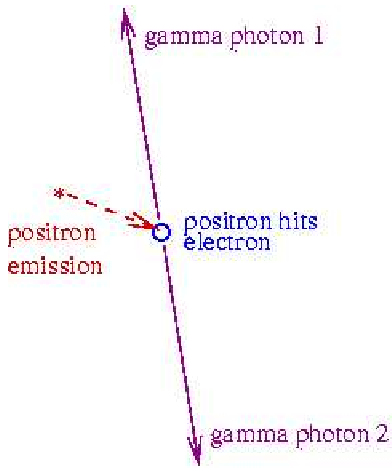
Positron Emission Tomography (PET)

- In positron emission tomography, a different type of radionuclides is injected into the patient's body. These isotopes decay emitting *beta rays* consisting of *positrons*, the positively charged anti-particles of electrons.
- In dense matter, each positron will meet an electron after a short distance (millimetres to centimetres).
- As anti-matter particles, positrons *annihilate* with their anti-particles as soon as they meet. That is, both particles are destroyed, and a pair of gamma photons is emitted in opposite directions.



- Detecting both gamma photons simultaneously allows to localise the beta decay event to a line, with unsharpness resulting from the positrons free movement.

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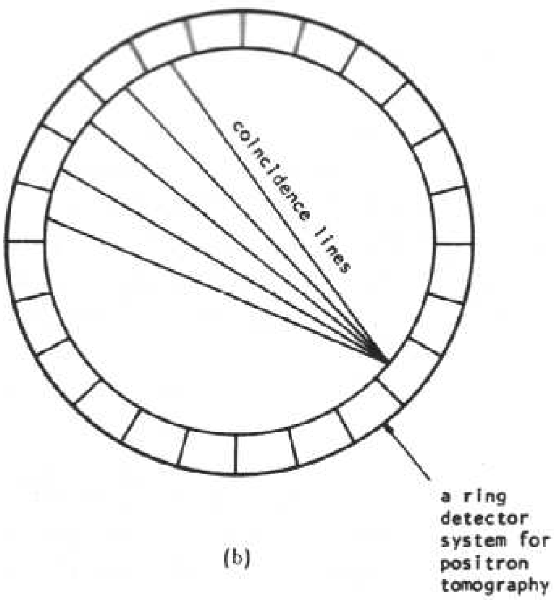
- PET is particularly valuable for its ability to map metabolism activity by measuring the oxygen consumption in tissue.
- Detector systems consist of X-/gamma ray detectors and additional electronic circuits to detect coincidence of detected photons.
- Different detector configurations are possible. One of them is a ring of detectors. For each pair of detectors, coincidence detection takes place. This allows to measure densities of positron annihilation events along each line connecting two detectors.



Detector ring for PET. *(Kak/Slaney 2001)*

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Attenuation Compensation for PET

Assume we have a line segment between two detectors D_1 and D_2. For simplicity, use 1-D coordinates with D_1 at 0 and D_2 at X. Assume an annihilation event on this line, at coordinate x_0, emits antiparallel gamma photons along this line, i.e. directed to D_1 and D_2.

- The probabilities of the event to be detected by D_1 and D_2 are proportional to

$$\exp\left(-\int_0^{x_0} \mu(x)\,\mathrm{d}x\right) , \qquad \exp\left(-\int_{x_0}^{X} \mu(x)\,\mathrm{d}x\right) .$$

- Attenuation on both line segments is independent such that the probability for simultaneous detection is

$$\exp\left(-\int_0^{x_0} \mu(x)\,\mathrm{d}x\right) \times \exp\left(-\int_{x_0}^{X} \mu(x)\,\mathrm{d}x\right) = \exp\left(-\int_0^{X} \mu(x)\,\mathrm{d}x\right)$$

  which is independent on x_0 and identical with the attenuation factor for X-ray transmission along the same line.

- An X-ray transmission scan in the same energy range (without CT reconstruction) can thus be used for attenuation compensation in PET.

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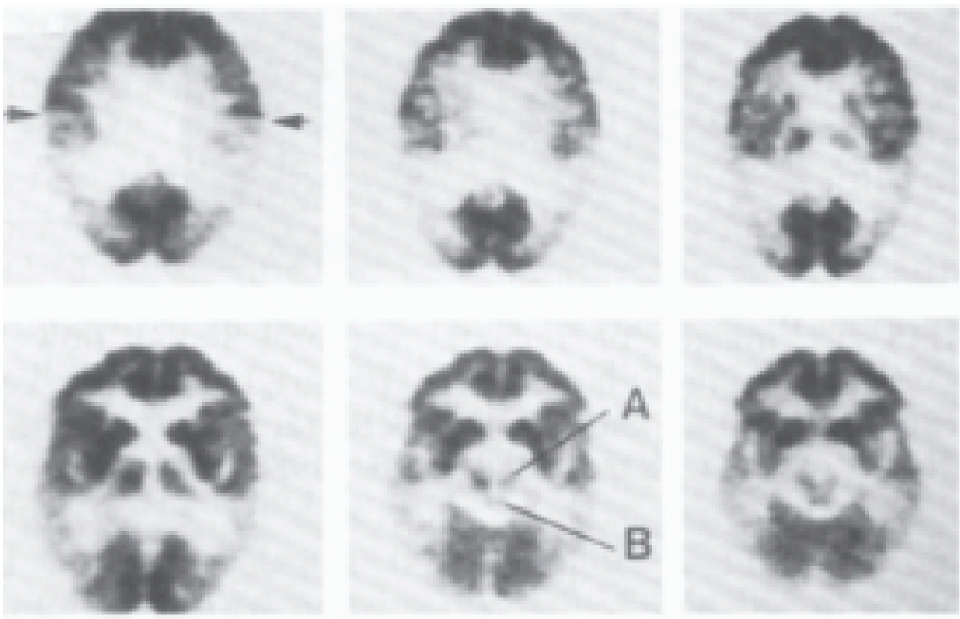
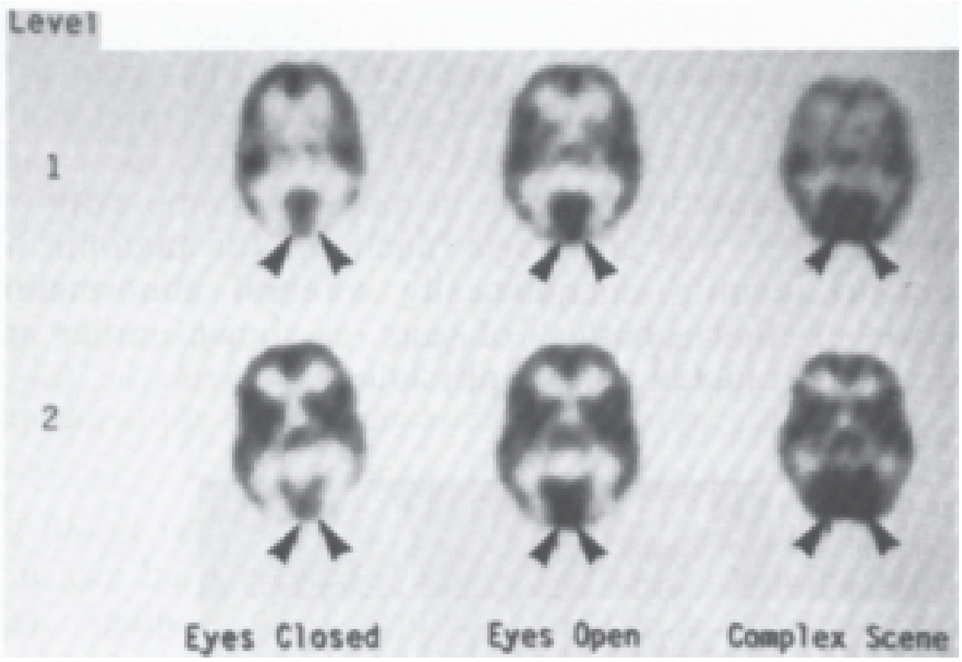
PET Examples



PET images of human brain (six levels) using a glucose derivate marked with a radioactive Fluor isotope. Arrows indicate regions with reduced metabolistic activity caused by Alzheimer's disease. A and B point to specific fine structures. *(Phelps 1986, reproduced from Webb 1988)*

PET Examples



PET images of human brain (two levels) using a glucose derivate marked with a radioactive Fluor isotope. The activity level in the visual cortex (lower part of the brain slices) can be seen, dependent on the visual activity. *(Phelps 1986, reproduced from Webb 1988)*

Summary

- Aliasing and ghosting are typical discretisation artifacts in CT.
  The number of projections and the number of rays per projection should be roughly equal.
- Partial volume effects due to the beam width may create streak-like artifact at object boundaries.
- Beers law may be violated due to wavelength-specific absorption.
- Photon noise in CT is of Poisson type.
- SPECT and PET are two types of computerised emission tomography.
  Both measure activity and have fairly low resolution.
- SPECT measures gamma rays and is the tomographic extension of scintigraphy.
- PET measures two opposite gamma photons caused by an annihilation of a positron with an electron.

References

- S. Webb, *The Physics of Medical Imaging.* Institute of Physics Publishing, Bristol 1988.
- A. C. Kak, M. Slaney, *Principles of Computerized Tomographic Imaging.* SIAM, Philadephia 2001.
- C. L. Epstein, *Introduction to the Mathematics of Medical Imaging.* Pearson, Upper Saddle River 2003.