Nuclear Medical Imaging assignment

BIOM9027 Medical Imaging

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Contents

Introduction

Because the references are used throughout an piece, for convenience, the references appears at the end of the section where they are used.

1 Question 1

1.1 a)

An isotope is defined as a version of an chemical element. The isotope has a different mass number than the element. Radioisotope's are radioactive isotopes of an element, i.e atoms that contain an unstable combination of neutrons and protons. [4], [7]

1.2 b)

The radioisotopes produced by a cyclotron accelerating protons have a nucleus with additions of protons, i.e there is a deficiency of neutrons (proton rich). The radioisotopes produced by neutron activation in a neutron reactor on the other hand have an nucleus with additions of neutrons, i.e there is a excess of neutrons (neutron rich). [1]

1.3 c)

Beta emission

In beta emission an unstable nucleus either converts a proton into a neutron or a neutron into a proton. The processes results in the creation of an positron and an electron neutrino or an electron and an electron antineutrino, respectively. [4]

Electron capture

In electron capture an electron is captured by a proton in the nucleus. This triggers a proton to neutron transformation and an ejection of an electronic neutrino. [4]

Positron emission

Positron emission is a type of beta emission (β^+-decay) where the proton rich nucleus converts an proton in to an neutron, a *positron*, and an electron neutrino. [4]

Isomeric transition

A decay mode may leave the nucleus in an excited state. The nucleus will reach its ground state through one of the two *isomeric transitions*:

i Emitting the excitation energy in the form of one or more γ-photons.

ii Transferring the excitation energy to an orbital electron (internal conversion). The vacancy left by the ejected electron is filled by a transition from a higher atomic shell. This results in X-rays (and/or auger electrons).

[4], [8]

1.4 d)

Nuclear medicine record radiation emitting from within the body rather than radiation created from external sources, you have to be able to detect the radiation externally. Hence the radioactive decays prefered for diagnostic imaging is positron emission (produces two 511 keV photons) and isomeric transition (produces γ- and X-rays) since this radiation can be detected. This is indirectly (and directly) Beta emissions. [4]

2 Question 2

2.1 a)

The energy of the scatter photon from an single Compton scattering is given by Equation 1.

$$E_{scatter} = \frac{E_0}{1 + \frac{E_0(1-cos(\theta))}{511}} \tag{1}$$

Where $E_{scatter}$ and E_0 are given in keV and θ is the scatter angle. For a initial photon energy of $E_0 = 140$ keV and a scatter angle $\theta = 30°$ the energy of the scattered photon becomes $E_{scatter} \approx 135$ keV.

2.2 b)

Compton scatter is the interaction between a photon and a weakly bound electron in the sense that the energy of the electron is much smaller than the photon. The photon loses a part of its energy to the recoil electron and is scattered as a photon of lower energy. In photoelectric interaction the photon interacts with tightly bound electrons. The photon is absorbed by the electron which then is ejected from the atom. [4]

2.3 c)

The probability of a photon undergoing photoelectric attenuation is proportional to the inverse if the energy cubed. So if the gamma rays originate from an part of the body with great depth realtive to the detector they will undergo greater attenuation (e.g compton scattering). Hence the probability of photoelectric interaction increases and will result in a absorption, leading to less rays reaching the detector. Note that even if the photons reach the detectors they may have

lost so much energy that they get rejected due to the energy discrimination. This will wrongfully describe the activity distribution of the object at the detectors and contribute to that the image is minimised.

3 Question 3

3.1 a)

Deterministic effects only occur after a threshold of exposure has been reached. The severity of the effects is then increased with an increased dose. Stochastic effects does not have a threshold. Instead the probability for an effect to occur is increased with an increased dose. Since you can control the deterministic effects and not (to the same extent given the the random nature) the stochastic effects, the latter is predominantly what need to be considered as potential side effects. [4]

3.2 b)

The diagnostic reference level is a level which apply to an easily measured and quantity (usually the absorbed dose). The purpose of the level is to help avoid excessive radiation dose that does not add clinical information. In nuclear medicine the reference level is in activity (Mbq) - Actvity reference level. [3], [2]

4 Question 4

The (modern) gamma camera consists of four main components. The collimator, the scintillator, photomultiplier tubes and the computer. The collimator "blocks out/absorbs" the gamma rays which origin cant be determined. When the gamma rays hit the scintillator (e.g NaI(Tl)) the gamma energy is absorbed and re-emitted as visible light. The number of light photons is proportional to the energy of the gamma-ray. The photomultiplier tubes detect and amplifies the incoming light photons, the photons are converted into an electrical pulse. The signal is proportional to the incident photon energy. The signal is converted to x,y-positions and a discrete z-signal indicating that the energy is within pulse height analyser window. The x and y signals are converted to digital form and are in some way stored (matrix or list mode) and processed by the computer. The image is then displayed on a standard computer video display. [4]

5 Question 5

The code for generating the plots in question 5 is displayed in Section Matlab.

5.1 a)

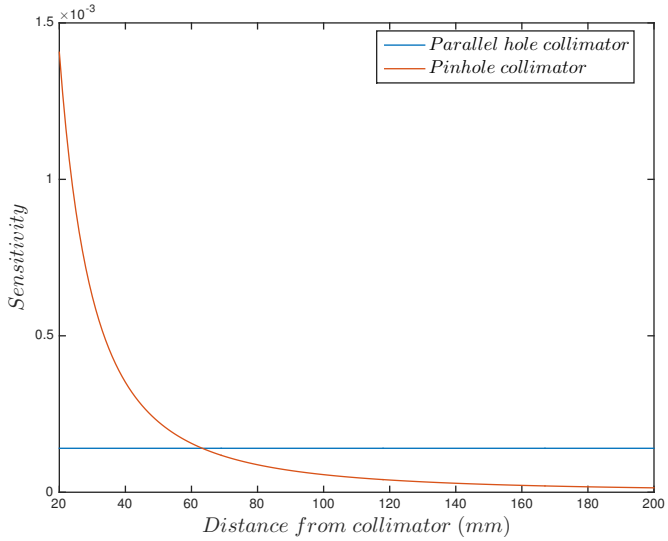

Figure 1: Collimator sensitivity.

5.2 b)

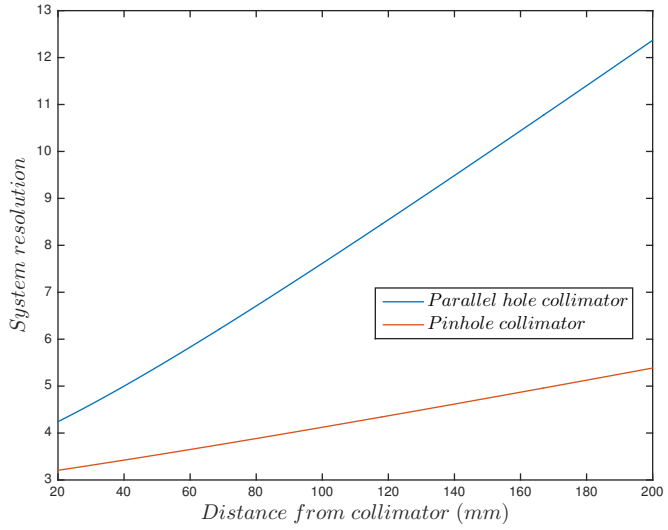

Figure 2: System resolution.

5.3 c)

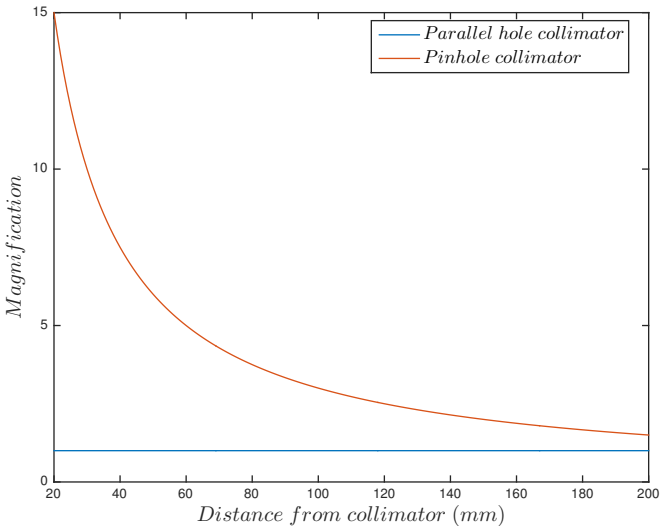

Figure 3: Magnification.

6 Question 6

6.1 a)

SPECT (Single-photon emission computed tomography) is a imaging technique using gamma rays. Hence it requires the delivery of a gamma-emitting radioisotope to the patient. SPECT imaging is performed by a gamma camera taking multiple 2D-images (planar images) of the 3D-distribution of a radioisotope. The projections are taken at different angles around the patient and the corresponding images can be reconstructed by algorithms, for example: Linear superposition of back projections (LSBP) and Filtered back projection.

6.2 b)

The advantages of SPECT over planar imaging include, since you have more information, better separation of underlying structures. Increased contrast due to removal of background. All in all, if the appropriate corrections are performed, you get a more informative picture. Some of the disadvantages are reconstruction artefacts as hot sources and changing activity distribution. Scattered photons causes apparent build up of activity towards the centre. The limited resolution can result in partial volume artifacts. Also, planar non-uniformities are enhanced by reconstruction. Another thing to consider is the possible offset

of the centre of rotation. This can cause, for example, collimator defects and spatial non-linearities. [5]

6.3 c)

A gamma-camera non-uniformity will lead to ring artefacts in the reconstructed SPECT image. The size of the non-uniformity artefact depends on the magnitude of the non-uniformity in the planar image and the distance of the non-uniformity from the centre of rotation. Especially the reconstructed non-uniformity is inverse proportional to the squared root of the distance from the centre of rotation.

7 Question 7

7.1 a)

The resolution of PET is significantly better than in SPECT. Also the resolution of SPECT degrades rapidly with distance. What that means is that you get different resolutions of an object depending on the angle, this gives raise to image distortion and non-uniformity. The collimator in SPECT is positioned directly in front of the detectors, this reduces sensitivity since many of the photon is absorbed by the collimator. In PET this is not necessary and results in a higher sensitivity. The sensitivity gain translates to a shorter image acquisition time. PET images can be quantified in terms of local activity concentration. Many PET scanners are full-ring devices that simultaneously measure projections, in SPECT the camera have to be rotated around the patient. In addition PET have less scatter and enables exact attenuation correction. [6]

7.2 b)

Several factors degrade the PET data acquired, to compensate for them one applies different corrections. [4]

Attenuation correction

Only a small fraction of the of the photon pairs escapes the body without interaction. These interactions can result in scattering photons and give raise to scattered coincidence events. The interactions can also attenuate the photons at such a great extent that they doesn't get detected. To account for the underestimation of true counts there are different methods of attenuation correction:

i Calculated attenuation correction

ii Measured attenuation correction

iii Segmented attenuation correction

iv CT/MRI-based attenuation correction

[4]

Randoms correction

Random coincidences accounts for a large part of the meased coincidence events. If not corrected it will contribute to loss of contrast and quantitative accuracy. One correction method involves estimating the random contributions, this is the most widely adopted correction method. The method uses an extra coincidence circuit (delaying the pulse from one of the detectors) which only detects random events (coincidence event between corresponding annihilation photons can't be detected). The random events count can be used as a on-line randoms correction by subtracting the number of random events from the total number of events. Alternatively the data from the random events can be stored and to later be used in a image reconstruction algorithm. Another method is to estimate the randoms from the Singles Method, given by Equation 2.

$$N_{randoms} = 2\tau N_{detector1} N_{detector2} \tag{2}$$

Where 2τ is the coincidence timing window and $N_{detector1}$ and $N_{detector2}$ are the singles rates at two opposing detectors. [4]

Scatter correction

Scattered photons can only be partially rejected using energy discrimination. Hence the uncorrected scatter forms a background in the reconstructed images which reduces contrast and degrades quantitative accuracy. There are several methods of scatter correction. E.g the scatter can be estimated by fitting a smooth analytical function to the scatter tails. [4]

Dead time correction

When a detector is processing a photon at the same time as another photon is incident, the second photon may be lost. At high count rates the likelihood of this occurring is high and results in a loss in sensitivity. This kind of loss is called dead time. The photons could also "pile up" resulting in that the detector registers the two photons as one higher energy photon. Dead time corrections are usually based on experimental measurements. A analytical model for the scanners count rate response can be derived from the measurements and used for dead time correction. [4]

7.3 c)

A hybrid PET scanner is besides the PET scanner an additional scanner. The images are acquired from both devices in the same session and can be combined into a single image. PET is functional imaging which shows the distribution of metabolic activity, this together with a scanner with anatomic imaging (e.g CT) can be used in such a way that the metabolic activity is more precisely

aligned. This also enables an effective attenuation correction by using the separate measurement of tissue attenuation. Hence the PET images will be free from attenuation artifacts and and precision of anatomic localization is added to functional imaging. [4]

8 References

[1] Australian nuclear science and technology organisation. (n.d.). *Cyclotrons* [Online]. Available: http://www.ansto.gov.au/NuclearFacts/AboutNuclearScience/ReactorsandAccelerators/Cyclotrons/

[2] Australian radiation protection and nuclear safety agency. (2013, Oct). *National diagnostic reference levels factsheet* [Online]. Available: http://www.arpansa.gov.au/pubs/Services/NDRL/NDRLfactsheet.pdf

[3] Cynthia H. McCollough. (2010, Nov). *Diagnostic Reference Levels* [Online]. Available: http://www.imagewisely.org/~/media/ImageWisely-Files/Medical-Physicist-Articles/IW-McCullough-Diagnostic-Reference-Levels.pdf

[4] D.L. Bailey et al. *Nuclear Medicine Physics - A Handbook for Teachers and Students*, International Atomic Energy Agency.

[5] Gustav Konrad von Schulthess. *Molecular Anatomic Imaging: PET-CT and SPECT-CT Integrated Modality Imaging*, Lippincott Williams & Wilkins, 2007.

[6] Martin A. Lodge et al. (2013, Oct). *Developments in Nuclear Cardiology: Transition from Single Photon Emission Computed Tomography to Positron Emission Tomography/Computed Tomography.*

[7] Wikipedia. (2016, Sep 7). *Isotope* [Online]. Available: https://en.wikipedia.org/wiki/Isotope#Radioactive.2C_primordial.2C_and_stable_isotopes

[8] Wikipedia. (2016, Aug 31). *Nuclear isomer* [Online]. Available: https://en.wikipedia.org/wiki/Nuclear_isomer#Decay_processes

9 Matlab

Listing 1: Code for question 5.

```
%% Nuclear imaging assignment.
clear all
close all
clc
clf

%Parallel
Dparr = 2; %mm
T = 0.2; %mm
C = 0.068; %hex hole constant
Le = 40; %mm

%Pinhole
L = 300; %mm
Deff = 3; %mm
alpha = 0; %degrees

% a
H = 20:1:200; %mm

Sensparr = (C*Dparr^4)/((Le*(Dparr+T))^2) + 0*H;
Senspin = (Deff^2)./(16*H.^2);

plot(H,Sensparr)
hold on
plot(H,Senspin)
hold on
ylabel('$Sensitivity$','interpreter','latex','fontsize',16)
xlabel('$Distance \; from \; collimator \; (mm)$','interpreter','latex','fontsize',16)
h = legend('$Parallel \; hole \; collimator$','$Pinhole \; collimator$','Location','Best');
set(h,'Interpreter','latex','fontsize',14);
saveas(gca, '5a.eps','epsc');

%b
clc
clf
close all

FWHMparr = Dparr*(Le+H)/Le;
FWHMpin = Deff*(L+H)/L;
Intres = 3; %mm
Mparr = 1 + 0*H;
Mpin = L./H;

Rparr = sqrt(((Intres^2)./(Mparr.^2)) + FWHMparr.^2);
Rpin = sqrt(((Intres^2)./(Mpin.^2)) + FWHMpin.^2);

plot(H,Rparr);
hold on
plot(H,Rpin);ylabel('$System \; resolution$','interpreter','latex','fontsize',16)
xlabel('$Distance \; from \; collimator \; (mm)$','interpreter','latex','fontsize',16)
```

```
h = legend('$Parallel \; hole \; collimator$','$Pinhole \; collimator$','Location','Best');
set(h,'Interpreter','latex','fontsize',14);
saveas(gca, '5b.eps','epsc');

%c
clc
clf
close all
plot(H,Mparr);
hold on
plot(H,Mpin);
ylabel('$Magnification$','interpreter','latex','fontsize',16)
xlabel('$Distance \; from \; collimator \; (mm)$','interpreter','latex','fontsize',16)
h = legend('$Parallel \; hole \; collimator$','$Pinhole \; collimator$','Location','Best');
set(h,'Interpreter','latex','fontsize',14);
saveas(gca, '5c.eps','epsc');
```