Detectors and Nuclear Electronics in Medical Imaging
BMD

August 2006
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Detectors and nuclear electronics

The human sense is unable to react to ionizing radiation therefore dedicated instruments have to be used to detect its presence. With these instruments it is possible to determine the type of radiation, measure its energy and register other parameters.

All instruments use the fact that the radiation interacts with matter, mostly via ionization. The detector converts the deposited energy to registable signals. These are sometimes detectable by the human eye (like X-ray photographic plates) or are electronic signals.

There are dedicated detectors for different type of radiation but they all rely on fundamental interaction processes. In this laboratory work some basic type of detectors for registering radiation is shown and the necessary signal handling technique is described.

The most common types of the ionizing radiation is the well known rays alpha, beta and gamma. This notation is a reminiscence from early days of nuclear physics and denote the charged helium core, the charged electron (or positron) and the neutral electromagnetic radiation respectively. All these types of radiation are present in our environment.

Detecting charged particles

To detect radiation one utilizes the interaction process with matter where the interacting medium converts the invisible radiation to detectable signals. If the radiation consists of charged heavy particles such as alphas, or light ones, like electrons or positrons, the electromagnetic interaction create charges which can be collected and detected. It can also initiate further processes, which can give rise to registable signals in the detector medium.

Detecting gamma radiation and other neutrals

The neutral gamma radiation interacts with matter with electromagnetic processes and transfer part or all its energy to electrons. These processes are the well-known Compton- and photo-effect, respectively. For photon energies above 1.022 MeV electron-positron pair creation is possible. To register other neutral particles, like neutrons, muons etc one has to rely on similar two (or more) step processes. The radiation has to interact with matter and transfer its energy to charged particles (electrons) and than the detection process mentioned above is used.
Ionizing radiation for medical imaging

Three different techniques can be used in medical imaging when nuclear processes are involved. In transmission technique like in the traditional X-ray imaging an external electromagnetic source is used to “shine through” the patients body and the attenuation information of the electromagnetic radiation is used to create an image. The picture obtained is thus dependent on the differences in absorption of the radiation and on the spatial resolution of the detector. Furthermore, the traditional X-ray radiography is a projection image, meaning that the X-ray attenuation information concerning the three-dimensional anatomy of the patient is projected on the two dimensions of the radiograph. Still, after hundred years, the technique is not far from what was used by its inventor, Röntgen, a metal excited by energetic electrons re-arranges its atomic shell and emit X-rays. The radiation penetrates the examined body and the degree of absorption is recorded on a planar sheet of radiation sensitive film.

In computed X-ray tomography, CT, technique a three-dimensional image (or rather a set of two-dimensional slices) of the object is reconstructed from a set of X-ray attenuation data. The method involves a traditional X-ray source, however the detector provides the transmission data as electronic signals to a computer.

In the nuclear radiomedicine technique, the activity is injected, inhaled or orally taken into the body of the patient and the organ to be examined is the emitter of the radiation. The diagnostic picture is constructed by recording the radiation. The method allows a detection of time dependent biological processes and is often applied for functional studies. A necessary prerequisite is the use of radio-pharmaceuticals. Although the radioactive imaging suffers from relatively poor spatial resolution it provides information about the physiological functions of the patient that is difficult to obtain from other imaging modalities.

The third imaging technique involving nuclear processes is when the organ to be examined via external stimuli is forced to emit detectable signals. The nuclear magnetic resonance imaging technique, MR, is an illustrative example of the success of the method. Here atomic nuclei with an odd number of nucleons (to have a non-zero magnetic moment, mostly protons) are stimulated by radio-frequent radiation. If an outer magnetic field with right strength is applied a small fraction of the protons will emit resonant radio-frequent radiation and thus reveal their positions. A delicate and elaborated technology is used to reconstruct a diagnostic picture and is one of the fastest developing fields of medical imaging. Although the method is based on fundamental nuclear processes the technique uses non-ionizing radiation (radio- signals in the tenths of MHz range) and therefore the detectors involved are not discussed in here.
Detectors

In transition medical imaging and in nuclear medical examinations the ionizing radiation is of electromagnetic type. The corresponding detector should therefore be able to register the charged secondary products, the electrons, of the radiation. In the following section some of the major detector groups are described. The common requirement for all detectors is good spatial and energy resolution. In addition many of the techniques requires precise time information.

Basic interaction processes of the electromagnetic radiation (like X- and gamma rays) are:

- **Photoeffect**
- **Compton scattering**
- **Pair production** (above the energy threshold 1022 keV, the mass of the electron-positron pair)

In all three processes the secondary charged products which carries the energy information are electrons (and positrons). To detect these electrons and measure their energies the ionization property is used. It can result in chemical processes, blackening of the photosensitive film, it can create vacancies in atoms resulting in light emission in scintillators or creating charges (electrons and holes) in semiconductors. The latter two types of detectors are also able to give the often-necessary information about the exact time of the interaction.
Gaseous type of ionization detectors

The most widely used type of detectors is based on the effects produced when a charged particle passes through a gas. The charged particle, the electron originally produced in one of the processes mentioned above, would along its track ionize the gas molecules. The energy needed to create an electron ion pair in gases is in the order of 30 eV and depends on the type of the molecule. If an electric field is applied the created charge will be collected on the electrodes resulting in an electric pulse which contains the total collected charge and thus the absorbed energy of the electron.

Depending on the voltage applied across a gas-filled chamber, the free electrons created might give rise to secondary ionization. The counter will work as an electron multiplier resulting in much larger output signals.

One of the advantages of the gas filled counters is the relatively simple and cheap construction. It permits also to manufacture large area detectors with multiple wires having position sensitive readout. In the laboratory exercise, X-ray imaging with wire chamber, the properties and the mode of application for imaging of a multi-wire proportional chamber is

The pulse height produced by different gas-filled counters as a function of the applied voltage, for two different radiations differing in energy by a factor of 2. In the Geiger-Müller region, all radiations give the same output pulse-height. In the other regions, the output pulse height is proportional to the energy deposited by the radiation through primary ionization.
At substantially higher electric fields electron avalanches are created as a result of the original ionization. Under proper conditions one avalanche can trigger another one at a different position in the tube, now working as a Geiger Müller counter. A typical pulse from the Geiger Müller tube represents an unusually large number of ions, (up to $10^{10}$ charges), giving rise to output pulses of identical amplitude and in the order of some volts. The electronic circuits for pulse treatment and counting are therefore greatly simplified. The GM tube is mostly used as a radiation survey monitor.

![Diagram of electron avalanches](image)

*The mechanism by which additional avalanches are triggered in a Geiger-Müller discharge.*

**Scintillators**

A scintillator is a transparent material that converts the energy lost by ionization into pulses of light. The technique is one of the oldest one and still is the most commonly used when the energy of the ionizing radiation is recorded. As mentioned in the previous section the ionization is due to charged particles. For detecting gamma radiation this will be the electrons (or positrons) from one of the three interaction processes (photo-effect, Compton scattering and creation of electron-positron pairs). In the scintillation process the kinetic energy is then converted into light in the visible wavelength region by exciting the atoms of the scintillator. At de-excitation of the atomic states the intensity of the emitted light is proportional to the ionizing particle, the electron. The conversion is linear over a wide energy range. The scintillating medium is transparent with a refractive index not too far from the construction material of the light detecting device (glass or plastic).
There are a large number of scintillator materials, with a variety of dedicated properties depending on their application. For detecting electromagnetic radiation, both organic and inorganic scintillating crystals are used. To obtain a high conversion property of the ionizing radiation to light, the scintillating material should have high density (large Z). This is the case for the most commonly used inorganic crystals such as NaI, CsI. Even higher density is obtained in bismuthgermanat, BGO, an expensive but useful scintillating material, essential in position sensing systems, such as the Positron Emission Tomograph, PET. Another property of the detector material is its speed to convert the ionizing radiation to light. One of the fastest scintillator material is BaF\(_2\), which is used in Time-of-Flight, TOF-PET, technique where the time difference of the two positron-electron annihilation quanta is measured. A laboratory exercise on our TOF-PET system is devoted to a detailed study of this imaging technique.

Properties of some of the inorganic scintillators commonly used are listed below.

<table>
<thead>
<tr>
<th>Scint.</th>
<th>Density (g/cm(^3))</th>
<th>Wavelength (\lambda\text{max}) (nm)</th>
<th>Decay time (ns)</th>
<th>Light yield (quanta/MeV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>NaI(Tl)</td>
<td>3.67</td>
<td>415</td>
<td>230</td>
<td>38000</td>
</tr>
<tr>
<td>CsI(Tl)</td>
<td>4.51</td>
<td>540</td>
<td>1000</td>
<td>52000</td>
</tr>
<tr>
<td>BGO</td>
<td>7.15</td>
<td>505</td>
<td>300</td>
<td>8200</td>
</tr>
<tr>
<td>BaF(_2) (slow)</td>
<td>4.89</td>
<td>310</td>
<td>620</td>
<td>10000</td>
</tr>
<tr>
<td>BaF(_2) (fast)</td>
<td>220</td>
<td></td>
<td>0.6</td>
<td>1800</td>
</tr>
</tbody>
</table>

The light emitted in the scintillator is converted to detectable electronic signals by a photo-multiplier, a PM tube, or by a photo-diode. The conversion is linear and the amplitude of the output signal will therefore be proportional to the absorbed energy of the radiation, the kinetic energy of the electrons. In the photo-cathode of the PM tube, the incoming light will stimulate electron emission. The electrons are accelerated by a potential to an electrode, called a dynode, with a properly selected surface to give a multiplying effect. Tubes with 10 – 14 dynodes are available and the number of electrons is multiplied in the dynode chain of the pulses.
Schematic of photomultiplier operation. Electrons released from the cathode are attracted to the first dynode and multiplied. Each successive dynode is at higher potential than the previous one, a typical tube might have 10 or 14 dynodes. At each stage, the number of electrons is increased by a factor of approx. 3.

An alternative to a PM tube is a semiconductor photo-diode. These devices offer the advantages of higher quantum efficiency, (i.e. potential for better energy resolution), lower power consumption, compact size and ruggedness. However, since the diode has no internal gain, an external amplifier has to be used.

The wavelength of the emitted light from the scintillator and the sensitive region of the photo-detecting devise, PM tube or photodiode, is selected to be as close as possible.

CsI(Tl) has the advantage that it is non-hygroscopic, does not cleave and in addition the color of the emitted light suits the spectral response of silicon photodiodes. These detectors are compact and do not require any high voltage, are rugged and can be operated in high magnetic fields. These detectors are frequently used in arrays or matrices.
Elements of a two dimensional position-sensitive scintillation detector, commonly called a gamma camera.

Example of a human bone scan produced by using a gamma camera to image the distribution of methylene diphosphorurate labeled with $^{99m}$Tc. The upper and lower halves of the image each were produced with approximately $10^6$ counts over 3 min period, (W.L. Rogers, University of Michigan Medical Center).
Nuclear electronics and counting systems

Most detectors can be represented as a capacitor into which a charge is deposited. By applying HV on the anode of the PM tube, or detector bias on a solid state detector, an electric field is created which causes the charges to migrate and be collected. During the charge collection a small current flows, and the voltage drop across a resistor is the pulse voltage. The amplitude of this pulse is proportional to the absorbed energy of the radiation and its measurement is the goal of the analysis.

Preamplifiers and amplifiers.

The preamplifier is isolated from the high voltage by a capacitor. The rise time of the preamplifier’s output pulse is determined by the collection time of the charge in the detector while the decay time is the characteristic RC time constant of the preamplifier itself. Charge sensitive preamplifiers are used for solid state detectors with output voltage signal that is proportional to the input charge.

To maximize the performance, the preamplifier should be located at the detector to reduce capacitance of the leads, which can degrade the rise time and lower the signal size. Additionally, the preamplifier also serves to provide a match between the high impedance of the detector and the low impedance of the coaxial cables to the amplifier, which may be located at great distances from the preamplifier.

In germanium detector systems the preamplifier is usually permanently attached to the detector and the first transistor of the preamplifier is mounted on and cooled by the cryostat.

The amplifier serves to shape the pulse as well as further amplify it. The long delay time of the preamplifier pulse, (~50 μs), may not be returned to zero voltage before another pulse occurs, so it is important to shorten it by differentiation and only preserve the detector information in the pulse rise time. In amplifiers an active circuitry is used to form a short unipolar near-Gaussian shaped output pulse with optimum signal-to-noise characteristics.
Typical preamplifier and amplifier pulses are shown in the figure below.

Pulse height analysis

Pulse height analysis may consist of a simple discriminator that can be set above noise level and which produces a standard logic pulse for counting in a pulse counter. However most data consists of a range of pulse heights of which only a small portion is of interest. A most simple but time consuming method is to use a Single Channel Analyzer and a Counter.

A Single Channel Analyzer, SCA, has a lower and an upper level discriminator, and produces an output logic pulse whenever an input pulse falls between the discriminator levels. With this device, all voltage pulses in a specific range can be selected and counted. If a full voltage (i.e. energy) spectrum is desired, the SCA discriminators can be set to narrow range, window, and then stepped through a range of voltages. If the counts are recorded and plotted for each step, an energy spectrum will result. The method requires some time meaning that the decay of short lived radioactive (life times close to the measuring time) sources can not be studied by this method.
The block diagram of a SCA is shown below.

Basic operation of a single channel analyzer, SCA: only those signals whose amplitudes fall within the window defined by the upper and lower level threshold trigger the signal.

Multichannel Analyzer, MCA

The Multichannel Analyzer, MCA, which can be considered as a series of SCAs with incrementing narrow windows, basically consists of an Analog to Digital Converter, ADC, control logic, memory and display. The MCA collects pulses in all voltage ranges at once and displays this information in real time, providing a major improvement over SCA spectrum analysis.

To-days MCA often use a PC for data storage and presentation. The ADC and other control circuits are located on an extension card. The total number of ADC- and memory channels is often large to mach the resolution of solid state detectors (up to 16 K). The heart of a MCA is its ADC and apart the above mentioned Wilkinson type other type of ADC based on the successive approximation method resulting in a constant conversion time is used. Here, the incoming pulse is compared to a series of reference voltages to determine the height of the pulse.
Measurement of the Bone Mineral Density, BMD.
(A medical application of the interaction process of electromagnetic radiation.)

As mentioned above, the main interactions of X-rays and gamma rays with matter are

1. Photoelectric effect
2. Compton Scattering and
3. Pair Production above 1022 keV, the energy threshold for creation of an electron-positron pair

The penetrating photon beam, contrary to the charged particles, is not degraded in energy as it passes an absorber, only attenuated in intensity. This is due to the fact that the three interacting processes remove the photon from the beam either by scattering or by absorption. The photon that passes straight through will keep its original energy. The attenuation depends on the absorbing material and its thickness and will follow an exponential dependence, Beer’s Law.

\[ I(x) = I_o \ e^{-\mu x} \]

With \( I_o \) the incident beam intensity, \( x \) the thickness of the absorber and \( \mu \) the linear absorption coefficient (in units 1/length, [cm\(^{-1}\)]). The linear attenuation coefficient is dependent on the density of the absorbing material. This dependency can be overcome by normalizing the linear attenuation coefficient. This normalized term is called the mass attenuation coefficient, \( \mu_m \) [cm\(^2\)/g] = \( \mu \) [cm\(^{-1}\)] / \( \rho \) [g/cm\(^3\)] and

\[ I(x) = I_o \ \exp (-\mu_m \rho \ x) \]

In the figure below, the total mass attenuation coefficient as function of the photon energy is displayed for aluminum and lead with its three components due to the above mentioned processes (photo effect, Compton scattering and pair production). Note the presence of the K and L edges. These edges are reflecting the different binding energies of the electron in the material. The edges are not visible in the diagram for Al due to their low energies.

The advantage of the mass attenuation coefficient, \( \mu_m \), is that it depends solely on the energy of the incident photons and the atomic composition of the attenuating medium. The dependence on physical density is removed. For example, if calcium atoms are present in a material, their contribution to \( \mu_m \) depends only on the fraction of all atoms that are calcium, and not on the physical state of the particular compound, crystalline state or mixture that are present.
Absorption curves for other material, elemental and compounds, can be obtained from the National Institute of Standards and Technology, NIST:

The mass absorption coefficient for aluminum and lead for the three individual processes involved (Photo, Compton and pair production) and the sum of the three components as function of the photon energy. Note the presence of the K and L edges reflecting the binding energy of the electron in the material. These edges are not visible for Al due to their low energies.

For the purpose of bone densitometry measurements it is helpful to consider the quantity \( \rho x \) as the local value of the areal density in units g/cm\(^2\) defined as the integral mass of the bone mineral divided by the projected area. In practice when making bone density measurements it is necessary to separate the bone mineral and soft tissue contribution to the net attenuation of the X-ray or gamma beam. If the attenuation measurement is performed at two energies with different attenuation coefficients for the bone and the soft tissue the Bone Mineral Density, BMD, can be determined. The method based on this technique is therefore called DXA or DPA (Dual X-ray Attenuation or Dual Photon
Attenuation). In the figure below the variation of $\mu$ with photon energy for bone, fat and lean soft tissue is displayed.

![Graph showing variation of mass attenuation coefficients for bone and soft tissue with photon energy.](image)

*Variation of the mass attenuation coefficients for bone and soft tissue with photon energy.*

For a composed object (bone and soft tissue) there will be two transition equations, one for each photon energy:

Low energy:

$$ I^l = I^l_o \exp \left( - (\mu^l_s \rho_s x_s + \mu^l_b \rho_b x_b) \right) $$

High energy:

$$ I^h = I^h_o \exp \left( - (\mu^h_s \rho_s x_s + \mu^h_b \rho_b x_b) \right) $$

Provided that the values of the four attenuation coefficients ($\mu^l_s$, $\mu^l_b$, $\mu^h_s$ and $\mu^h_b$) are known the values of $\rho_s x_s$ and $\rho_b x_b$ can be solved.

In practical (clinical) use the value is determined pixel by pixel by scanning the object to be examined. The BMD (the product $\rho_b x_b$) is then evaluated by subtracting the normalized values from areas without a bone content. The procedure is illustrated in the figure below.
Cross section of a bone imbedded in soft tissue.

Attenuation profile of high energy photons.

Attenuation profile of low energy photons.

The high-energy profile is multiplied by a factor $k$ so that the soft tissue attenuation matches that in the low-energy profile.

When the two profiles are subtracted the Bone Mineral Distribution, BMD, is left.

$A_{\text{high}} = \ln(I_h(x)/I_{h,0})$

$A_{\text{low}} = \ln(I_l(x)/I_{l,0})$

$k \cdot A_{\text{high}}$

$A_{\text{low}} - k \cdot A_{\text{high}}$
Questions:

1. The three major interaction modes of photons with matter are …

2. The probability of the three interaction processes are depending on the energy of the incoming photons. The three dominating energy regions are …

3. The amplitude of the attenuation coefficient exhibits drastic changes (edges) in the low energy region. These edges depend on the absorbing material. Explain!

4. Lead bricks, with dimensions 5 x 10 x 20 cm$^3$, are used for shielding against radiation in nuclear physics laboratories. What is the attenuation factor for a 5 cm thick Pb-wall for 100 keV, 500 keV and 1 MeV gamma rays.

5. The two types of medical imaging (transmission and emission) both are using photons as carrier of information. What are the necessary requirements for the energies used in transmission imaging, like traditional X-ray or CT? and

6. What is the preferable energy for photons used in emission technique (nuclear medicine) like gamma camera, SPECT or PET?

7. How is it possible that the GM tube gives few volt large pulses when the released energy from the ionization results only a sub-millivolt signal?

8. The two portraits on the French 500 francs banknote are of …
Laboratory work


Measure the activity in the room with a GM system. Compare the counting rate for the natural background with calibration sources. Check how the count-rate depends (1) on the type of radiation (beta or gamma), (2) on the intensity of radiation by varying the distance to the source and (3) on the absorber between the source and the detector.

2. Scintillator

Study the pulses from a NaI scintillator equipped with a PM tube. Measure the pulse-height with an oscilloscope for different calibration sources.

**Amplifier.** Compare the output and input pulse shapes of an amplifier connected to the scintillator. Adjust the pole-zero-cancellation.

**SCA.** Connect the SCA to the amplifier above. Adjust the threshold and window width.

**PC – based MCA.** Record the spectra from calibration sources with the MCA of the laboratory. Calibrate the spectrum; determine the resolution of the scintillator as function of energy.

**PC – based MSC (Multi SCaler function)** Internal energy gating. Each channel act as a scaler. The first channel counts for a preset time, than the next channel takes over and register the events for the same time period. The operation continues until the last channel has counted the events.


To illustrate the method for measuring BMD the absorption profile of a cross section of a test object, a phantom, is recorded. The object used is a piece of bone in a paraffin block. The phantom is fixed on the object table of the BMD
scanner allowing recording a narrow beam of transmitted photons. The table is movable (left or right) with constant speed and a collimated beam from a photon source $^{241}$Am is used to scan the object. The source emits photons with variable energies and intensities:

$^{241}$Am ($t_{1/2} = 433$ year)

<table>
<thead>
<tr>
<th>Energy (keV)</th>
<th>Intensity</th>
</tr>
</thead>
<tbody>
<tr>
<td>13.9</td>
<td>13.3 %</td>
</tr>
<tr>
<td>17.8</td>
<td>19.3 %</td>
</tr>
<tr>
<td>20.8</td>
<td>4.9 %</td>
</tr>
<tr>
<td>26.3</td>
<td>2.4 %</td>
</tr>
<tr>
<td>59.5</td>
<td>35.7 %</td>
</tr>
</tbody>
</table>

- Measure the gamma ray energy spectrum of the source with a NaI scintillation crystal that is behind a 3 mm diameter collimator. Identify the spectrum.

- Select two energy regions ($E_{\text{low}} = 13.9 + 17.8$ keV and $E_{\text{high}} = 59.5$ keV) using the MCA cursor. Note the channel numbers of the limits for the two energy windows.

- Change the operation mode to Multi Scaling, MCS. In the MCS mode the system will count the signals that are within a selected energy region during a pre-selected time (dwell time), then the counting continues in the next channel during the same time period and so on. The system will act as a Multi Scaler and allow to record time dependent changes of count rates. The phantom in the BMD apparatus moves with constant speed, (approx. 6 min for the full range) therefore, the count rate as function of position, the absorption profile, can thus be recorded.

- The time for each MCS channel should be selected to fit the speed of the BMD apparatus. A dwell time of 1 sec will give 360 channels MCS spectrum.

- Start the motor and the data collection simultaneously. To be able to compare different absorption profiles the start point and the direction of the movement should be identical for each measurement.

- Select the low energy region and record the transmission profile of the cross section of the phantom. Repeat it for the high-energy region.
To read the MCS spectra from in the “MATLAB” PC use the command TUKAN.

Try to normalize to the region containing only “soft” tissue and subtract the two curves to obtain the net BMD distribution. Example for a MATLAB program:

```matlab
>> tukan
The low energy profile stored and displayed in MATLAB

>>low=data(1:360);
Replace 360 with the number of channels used if necessary

>>tukan
The high energy profile stored and displayed in MATLAB

>>high=data(1:360);

>>x=1:360;
Create x-vector with the proper number of channels.

>>plot(x,low,'r',x,high,'b');
The low energy profile is shown in red and the high energy in blue.

>>slow=smooth(low,10);
Smoothing of data to filter out statistical fluctuation. The average value of 10 neighbouring channels (on each side) are calculated for each point.

>>shigh=smooth(high,10);
Now find the intensity levels in air for the low and high energies, respectively. Use the zoom facility

>> corr_low=slow.*k;
Adjust the value of k to be approx. equal in the intensity ratio in air.

>> plot(x,corr_low,'r',x,shigh,'b');
Plot and check that two intensities in air are equal. We call this intensity value q.

>>alow=q-corr_low;
>>ahigh=q-shigh;
We now transform the data from transmission intensity to absorption.

>>plot(x,alow,'r',x,ahigh,'b')
Now, find the ratio of the two absorption levels in the soft tissue region. We find the ratio w.

>>corr_alow=alow.*w;

>>plot(x,corr_alow,'r',x,ahigh,'b')
Check that the two curves overlap everywhere except in the bone region.

>>bmd= ahigh-corr_alow;
subtract to find the BMD.

>>plot(bmd,'m')
```