Chapter 5  Photon Spectrometry with Scintillators

The most important application of scintillation detectors is photon (X- and γ-rays) spectroscopy. Although first introduced in the early 1950s, the NaI(Tl) detector still remains the most popular scintillator for photon spectrometry applications. In this chapter, we will discuss detailed principles and features of the photon spectroscopy with scintillation detectors as well as imaging applications.

Two laboratory experiments in this course are designed for γ-ray spectrometry. The first laboratory experiment uses a NaI(Tl) scintillation detector to measure the γ-ray spectrum of an air filter, the second experiment uses a Hyper-Pure Germanium (HPGe) detector, a semiconductor detector type, to measure and identify an unknown sample. HPGe detectors will be covered in the second term.

5.1. Photon interactions with matter

There are four interaction mechanisms of photons with matter and three of them are the means by which photons are detected. The four mechanisms are:

◆ Photoelectric absorption  ◆ Compton scattering
◆ Rayleigh scattering       ◆ Pair production

The predominant mode of interaction depends on the energy of the incident photons and the atomic number of the material with which they are interacting. At low energies, in high atomic number materials, the photoelectric effect is the main interaction of photons with the material. At intermediate energies in low atomic number materials (e.g. during a chest x-ray) the dominant interaction is Compton scattering. At very high energies, the main mechanism by which photons are detected is pair production. Rayleigh scattering is not involved in the detection process.

It should be noted that photons deposit energy in a material by transferring their energy to a secondary charged particle (electron), and it is the energy imparted to the electron that is deposited in the detector.

A. Photoelectric effect

In this case an incident photon ejects an electron, designated a photoelectron, from an absorber atom shell and the photon is completely absorbed as shown in Fig. 5.1. The kinetic energy of the photoelectron $E_e$ is given by

$$E_e = E_\gamma - E_b$$

where $E_\gamma$ is the incident photon energy and $E_b$ the electron binding energy. $E_b$ is usually small compared to $E_\gamma$, so that the photoelectron carries off most of the photon energy.

The vacancy in the electron shell is quickly filled by electron rearrangement, electrons from higher energy levels fill the vacancy. The binding energy is liberated as characteristic
(fluorescence) X-rays or Auger electrons. Assuming the binding energy is absorbed, the feature that appears in the measured spectrum as a result of photoelectric events is a full energy peak. The interaction cross section of the photoelectric absorption has a dependence of

$$\tau \propto \frac{Z^5}{E_\gamma^{3.5}} \quad (Z: \text{atomic number, } E_\gamma: \text{incident photon energy})$$

An example of the photon energy dependence is shown in Fig. 5.1 for NaI.

![Photoelectric effect and its mass attenuation coefficients of NaI](image)

**B. Compton and Rayleigh scatterings**

There are two kinds of the photon scatterings, Compton scattering and Rayleigh scattering. As shown in Fig. 5.2, a whole atom works as the target in Rayleigh scattering. When the incident photon is scattered by the atom and changes its direction, the target atom recoils to conserve momentums before and after scattering. Since the atomic mass is relatively heavy, the recoil energy of the atom is negligible and the photon changes its direction only and retains the same energy after scattering. Hence, no energy is transferred.

In Compton scattering, single electron works as a target and the recoil energy of the electron is larger than the electron binding energy. Therefore, the incident photon transfers part of the energy and the recoil (or Compton) electron is emitted after scattering.

![Rayleigh scattering](image)
![Compton scattering](image)
Fig. 5.4 shows Compton and Rayleigh interaction coefficients of NaI as a function of photon energy. Compton scattering is dominant above 0.3 MeV while photoelectric absorption is dominant in the low energy region.

![Fig. 5.4. Mass attenuation coefficients of Compton and Rayleigh scattering for NaI.](image1)

![Fig. 5.5. Polar plot of the Klein-Nishina differential cross section of the Compton scattering.](image2)

The result of a Compton scattering interaction is the creation of a recoil electron and a scattered photon as described. We can analyze the kinematics of Compton scattering assuming a free electron when the energy of the Compton electron is high enough. The division of energy between the scattered photon and the Compton electron is dependent on the photon scattering angle $\theta$.

From momentum and energy conservations, the wavelength of the scattered photon $\lambda'$ is given in terms of the scattering angle $\theta$ and the original photon wavelength $\lambda$ as

$$\lambda' = \lambda + \frac{\hbar}{m_e c} (1 - \cos \theta)$$

where, $m_e$ is the rest mass of electron. The corresponding energy of the scattered photon is

$$E'_\gamma = \frac{\hbar c}{\lambda'} = \frac{E_\gamma}{1 + \frac{E_\gamma}{m_e c^2} (1 - \cos \theta)}$$

where, $m_e c^2$ is the electron rest mass energy ($= 511$ keV).

Then, the energy of the recoil electron $E_k$ is

$$E_k = E_\gamma - E'_\gamma = E_\gamma - \frac{E_\gamma}{1 + \frac{E_\gamma}{m_e c^2} (1 - \cos \theta)}$$

Two extreme cases can be identified; the minimum and maximum energy transferred to the recoil electron.
Minimum energy transfer to recoil electron
Grazing angle scattering where \( \theta = 0; \quad E'_{\gamma} = E_{\gamma} \) and \( E_k \approx 0 \)

Maximum energy transfer to recoil electron

Head on collision in which \( \theta = \pi; \quad E'_{\gamma} = \frac{E_{\gamma}}{1 + \frac{2E_{\gamma}}{m_e c^2}}, \quad E_k = E_{\gamma} \frac{2E_{\gamma}}{m_e c^2} \Rightarrow \text{Compton edge} \)

\[
\frac{d\sigma}{d\Omega}_{KN} = \frac{r_e^2}{2} \left( \frac{E'_{\gamma}}{E_{\gamma}} \right)^2 \left( \frac{E'_{\gamma}}{E_{\gamma}} + \frac{E_{\gamma}}{E_{\gamma}} - \sin^2 \theta \right)
\]

Fig. 5.5 shows the polar plot of the Klein-Nishina differential cross section for several photon energies. The scattering occurs into all directions for low energy photons while the forward scattering is dominant for high energy photons. Since the electron binding energy has an effect when the energy of the Compton electron is not high enough, a correction term is required for low energy photon Compton scattering.

In normal circumstances, all scattering angles will occur. A continuum of energies can be transferred to the recoil electron, from 0 up to the maximum, i.e. Compton edge, as shown in Fig. 5.6. The shape of the Compton profile can be deduced from the angular distribution of Compton scattering.

Fig. 5.6. Distribution of the Compton continuum for 1 MeV photon.

C. Pair production

Fig. 5.7. Pair production interaction.
In the Coulomb field of nucleus (or electron), the incident photon energy can be converted to the creation of an electron-positron pair. A minimum of $2m_e c^2$ is required (twice the electron rest mass of 511 keV) and therefore pair production only occurs above the threshold of 1.02 MeV.

If the incident photon energy is above 1.02 MeV, the excess energy appears as kinetic energies of the positron-electron pair

$$E_{k-} + E_{k+} = E_\gamma - 2m_e c^2$$

Once the kinetic energy of the positron has reduced to approximately thermal energy; the positron annihilates with an electron in the absorbing medium with the resultant creation of 2 photons, each of 511 keV ($= m_e c^2$)\(^1\). The time required for slowing down and annihilation is small; the annihilation radiation appears in virtual coincidence with the original pair production interaction.

Each of the annihilation photons can then interact with the surrounding medium via the photoelectric effect or, more likely, Compton scattering.

![Fig. 5.8. Mass attenuation coefficients of NaI for all interactions with photon.](image)

\(^1\) 2 photons are created in > 95% of occurrences. 3 photons can be created, dependent on the state the positronium (electron-positron bound pair) is in when annihilation takes place.
5.2. Predicted detector response (spectrum)

A. Very large detectors

All photons, no matter how complex their mode of interaction, ultimately deposit all of their energy in the detector. For a monoenergetic incident photon beam, the pulse height analyzer will produce a single full-energy peak in the spectrum.

![Response of a very large detector for a mono-energy photon.](image)

B. Small detectors

Each interaction mechanism creates features within the observed pulse height spectrum:

- Photoelectric interactions give rise to a full-energy peak.
- Compton scattering events can occur where the scattered photon escapes the detector without further interaction. This gives rise to the Compton continuum.
- Pair production events can occur where both annihilation photons escape the detector. This gives rise to a peak of energy $E_\gamma - 2m_e c^2$. This feature is known as the double escape peak.

![Diagram showing interaction mechanisms](image)
Each interaction mechanism creates features within the observed pulse height spectrum:

- Photoelectric interactions give rise to a full-energy peak.
- Compton scattering events can occur where the scattered photon escapes the detector without further interaction. This gives rise to the Compton distribution.
Compton scattering events can occur where the scattered photon itself undergoes Compton scattering and then escapes the detector. This gives rise to energy deposition in the detector, higher in energy than the Compton edge, but lower than the full-energy peak. Finally, the scattered photon can also be completely absorbed, which gives a full-energy peak event.

Pair production events can occur where both annihilation photons escape the detector. This gives rise to a peak of energy \(E \gamma - 2m_e c^2\). This feature is known as the **double escape peak**.

Pair production events can occur where only one annihilation photon escapes the detector. This gives rise to a peak of energy \(E \gamma - m_e c^2\). This feature is known as the **single escape peak**.

**D. Complications in the response function**

**Secondary electron escape**: For high energy photons, the range of the secondary electrons can be larger than the detector size and a significant fraction of the electrons may leak from the detector surface. The Compton electron energy is not fully collected, which leads to a distortion of the Compton profile.

**X-ray escape peak**: A characteristic X-ray is emitted following the photoelectric absorption. In most cases, the X-ray is reabsorbed, however, if the photoelectric absorption occurred near surface region, the X-ray can escape, which results in decrease of the deposited energy. For low energy photons, the escape events can happen frequently and “X-ray escape peak” is formed.

**Backscatter peak**:

In many cases, a measured photon spectrum has a peak near 0.2 ~ 0.25 MeV, which is defined as **backscatter peak**. This peak is caused by the photons scattered from the detector surrounding materials as shown in Fig. 5.11. From the formula of the scattered photon energy in Compton scattering, the scattered photon energy of \(\theta = \pi\) becomes \(\sim m_e c^2/2\) if the incident photon energy is large. Thus, the backscatter peak always occurs at the energy of 0.25 MeV or less. In addition to Compton scattering, other interactions of the incident photons in the surrounding materials can give an X-ray peak and the annihilation peak.

![Fig. 5.11. Influence of surrounding materials on detector response.](image-url)
E. Summation effects

If two or more photons are detected coincidently, a detection system cannot separate individual events and the total sum of the deposited energies is treated like interactions induced by a single photon. The coincidence leads to loss of events for individual photon responses and therefore, should be analyzed carefully in both efficiency calibration and unknown radiation source measurement. If the summing effect is not negligible, a correction should be done. On the other hand, the coincidence can produce additional peaks (sum peaks), which are located in the higher energy region of the spectrum. There are two kinds of the summing effects, (true) coincidence summing and random (or chance) coincidence.

(True) coincidence summing is defined as the coincident detection of photons from the same decay event, and happens for radiation from a cascade photon emitter. Assuming the simplest decay scheme for the cascade gamma-ray emission as shown in the figure, the counting rates of the full energy and the sum peaks can be represented as [3-7]

\[
\begin{align*}
\dot{C}_1 &= A \varepsilon_1 (1 - \varepsilon_2 w) \\
\dot{C}_2 &= A \varepsilon_2 (1 - \varepsilon_1 w) \\
\dot{C}_s &= A \varepsilon_1 \varepsilon_2 w
\end{align*}
\]

where, \(A\) is the radioactivity of the source, \(\varepsilon_1, \varepsilon_2\) are peak efficiencies, and \(\varepsilon_{t1}, \varepsilon_{t2}\) are total efficiencies. The efficiencies in the equations are absolute rather than intrinsic, of course. \(w\) is the angular correlation between two gamma-rays which becomes 1 if the correlation does not exist, i.e. both gamma-rays are independent in emission direction. For convenience, the angular correlation effect is neglected. Due to the summing out effects, the full energy peak counting rate \(\dot{n}_i\) is decreased by a factor of \((1 - \varepsilon_{t0})\). In other words, the measured counting rate of each full-energy peak must be corrected by a factor of \((1 - \varepsilon_{t0})^{-1}\), which is defined as the coincidence summing correction factor. The (true) coincidence summing effect is dependent on the detection efficiency and is independent of the event rate.

If more than one excited states are populated after \(\beta^-\) decay, the counting rates of peaks are

\[
\begin{align*}
\dot{C}_1 &= A p_i \varepsilon_1 (1 - \varepsilon_{t2}) \\
\dot{C}_2 &= A p_i \varepsilon_2 (1 - \frac{p_1}{p_2} \varepsilon_{t1}) \\
\dot{C}_s &= A p_i \varepsilon_3 + A p_i \varepsilon_1 \varepsilon_2 = A p_i \varepsilon_3 (1 + \frac{p_1 \varepsilon_1 \varepsilon_2}{p_3 \varepsilon_3})
\end{align*}
\]

where, \(p_i\) is the emission probability of \(\gamma_i\) per decay. In this case, \(\gamma_1\) and \(\gamma_2\) undergo the summing out effect while \(\gamma_3\) has the summing in effect. For radionuclides with more complicated decay schemes, the counting rate of each peak can be modeled by extending the basic principles used in the simple schemes above.
The dependence of coincidence summing on the detection efficiency can be demonstrated with the $4\pi$ NaI(Tl) array of our department [8,9], which has been used for in vivo neutron activation analysis applications. The layout of the array is shown in Fig. 5.12. The array consists of eight large detectors (each one $10\times10$ cm$^2$, 40 cm long) as shown in the figure. Gamma-ray spectra from $^{137}$Cs and $^{60}$Co were measured by varying the number of detectors connected to the pulse processing system. As the number of detectors increases, the counting rate of the $^{137}$Cs 0.66 MeV peak increases while those of two $^{60}$Co full-energy peaks changes little due to the coincidence summing out effect.

![Diagram of NaI(Tl) array](image1)

**Fig. 5.12.** $4\pi$ NaI(Tl) array for in vivo neutron activation analysis and test spectra [8,9].

Another example is given for an annihilation photon emitter, $^{22}$Na, which decays either by $\beta^+$ or by electron capture. From the decay scheme of $^{22}$Na, the 1.27 MeV gamma-ray is almost always emitted (emission probability $\approx 1$) regardless of the decay mode. The probability of $\beta^+$ decay is 90.5 % while that for electron capture is 9.05 %. The annihilation radiations (511 keV) are produced when the $\beta^+$ radiation combines with an electron and the emission probability of the 511 keV photon becomes $2 \times 0.905 = 1.81$. In the spectrum from a standard $3''\times3''$ NaI(Tl) detector (Fig. 5.13), the most prominent peak is the annihilation radiation peak due to its high emission probability and high detection efficiency. There is a small sum peak induced by the coincidence summing between an annihilation photon and a 1.27 MeV gamma-ray. The spectrum in Fig. 5.14 from the $4\pi$ NaI(Tl) array shows a totally different pattern. First of all, the 511 keV peak is smallest due to an extreme coincidence summing out effect. The biggest peak is the triple sum peak (two annihilation photons + 1.27 MeV gamma) and the next one is the sum...
peak of two annihilation photons. The correction of the coincidence summing is absolutely required for a high efficiency detection system as illustrated here.

When a volume source is used for measurement, the analysis of the summing effect becomes more complicated. The most famous sample geometry is the Marinelli Beaker, which is used for radionuclide analyses for a liquid sample. In case of a volume source, the summing effect can be estimated as a volume-averaged value of the point source summing effect:

$$\frac{1}{V} \int c(p) \, c(r) \, dV \neq \left( \frac{1}{V} \int c(p) \, dV \right) \left( \frac{1}{V} \int c(r) \, dV \right)$$

The correction is complicated because, as shown in the equation, both position dependent peak and total efficiencies are required to estimate the average value. In general, the summing effect is not consistent with the simple multiplication of the volume averaged peak and total efficiencies. Thus, the experimental efficiency calibration is a tedious process for volume sources if the summing effect is fully included. In general, it is possible to calibrate the system for the specific geometry used by preparing standards for the radionuclides to be analyzed.

The other summing effect is the random summing (or chance coincidence) effect. Random summing occurs when the second pulse arrives within the resolving time following the preceding signal pulse. The probability of the random summing is dependent on counting rate, or event rate. The random summing results in a pulse pile-up, which can be actively rejected in the pulse processing system and will be discussed later in the chapter of the pulse processing.

F. Coincidence methods in gamma-ray spectrometry

Continuum reduction: The Compton continuum part of the detector response makes the spectrum complicated when there are many radionuclides. The coincidence or anticoincidence technique is an active way to simplify the detector response function so that the spectral analysis becomes easier. The most common approach involves the use of an annular detector surrounding
the primary detector for Compton suppression by anticoincidence gating. Since this technique is more important in HPGe detectors, it will be discussed in detail in the chapter of the HPGe detector.

**Compton spectrometer**: The combination of two separate detectors operated in coincidence, as shown in Fig. 5.15, is another way to simplify the response at the expense of efficiency. By selectively accumulating only those pulses from the first detector that are in coincidence with a pulse from the second detector, the measured spectrum reflects only single Compton scattering events. Then the response is reduced to a single peak, which is located at a position within the original Compton continuum.

![Fig. 5.15. The arrangement of a Compton spectrometer.](image)

### 5.3. Properties of scintillation spectrometer

**A. Response function**

The dependence of the NaI(Tl) response on photon energy was characterized experimentally many years ago. Extensive catalogs of measured gamma-ray spectra for ~ 300 radionuclides measured by 3”×3” NaI(Tl) detector have been published by several research groups. An online catalog is available at the Gamma-ray Spectrometry Center, Idaho National Laboratory [10] and is very useful for the scintillation spectrometers. Some examples of gamma-ray spectra are given in Figs. 5.16 and 5.17.

![Fig. 5.16. 2”×2” NaI(Tl) response for $^{57}$Co.](image)

![Fig. 5.17. Responses of 3”×3” NaI(Tl).](image)

A direct comparison of gamma-ray spectra measured with equal size crystals of NaI(Tl) and BGO is shown in Fig. 5.18. The better resolution of of NaI(Tl) is well identified from the
comparison of peak widths. On the other hand, the efficiency of BGO is much higher from the larger area of the full-energy peaks.

Fig. 5.18. Comparison of NaI(Tl) and BGO responses.

If a proper calibration source is not available, the response function should be calculated theoretically. Since the main interaction is Compton scattering over wide energy region and a photon may undergo many scatterings before it is finally absorbed by the photoelectric effect, the analytical calculation is hard to implement for full interaction processes. Monte Carlo simulation is best for this purpose because photons and secondary electrons can be fully followed until those particles escape or lose all kinetic energy, and the deposited energy in the detector volume can be estimated reliably.

Fig. 5.19. Monte Carlo simulation of a response.
B. Linearity

An ideal radiation detector would be perfectly linear, i.e. its output pulse height should be exactly proportional to the deposited energy in its sensitive volume. For real detectors, deviation from linearity is always present. In the case of NaI(Tl) detectors, it was recognized early that the scintillation light is not proportional to the deposited energy. Fig. 5.20 shows the relative light yield for fully stopped electrons as a function of the electron energy. It is still uncertain why electrons show this non-proportional behavior. Since the energy deposition occurs through the energy loss of the secondary electrons (photoelectron, Compton electron, …), the nonlinearity may be related with the electron energy loss pattern. Since a mono-energetic photon produces different energy secondary electrons depending on the interaction, the linearity for the photon can be deduced by combining the secondary electron spectrum with the relative light yield for electrons so that an average value can be calculated.

Fig. 5.20. NaI(Tl) relative light yield for fully stopped electrons.

C. Efficiency

Scintillation detectors are usually produced either with the right cylinder shape or with a well shape for general purposes. The right cylinder is simple to manufacture and can be mounted directly to the circular face of standard PMTs. A well type crystal is a cylinder into which a cylindrical well has been machined. A great advantage of the well geometry is, of course, the large solid angle (~ 4 \(\pi\) sr), which leads to a high efficiency.

Fig. 5.21. Absolute total efficiency for a 2"\(\times\)2" cylindrical NaI(Tl).

Fig. 5.22. Absolute total efficiency for a well-type NaI(Tl).
The (absolute) total efficiencies for a right cylinder and a well-type are presented as functions of the source position and the photon energy in Figs. 5.21 and 5.22. When the sources are located on the surface, the total efficiencies for low energy photons are 0.5 and ~1, respectively. This means every photon incident on the detector produces an output pulse considering the solid angles of both geometries (2\(\pi\) for right cylinder, ~4\(\pi\) for well type), regardless of the energy deposited.

Although the total efficiency is good for getting more number of counts, the analysis of the total areas from a measured spectrum is a complicated and difficult work in most applications. Practically, peak areas are preferred to simplify analysis. Fig. 5.23 shows the intrinsic peak efficiencies of BGO and NaI(Tl) of equal 38×38 mm\(^2\) size.

Another quantity conveniently used in efficiency calibration is the **peak to total ratio**, which is defined as the ratio between the peak and the total efficiencies. Fig. 5.24 shows the reported ratio of various cylindrical NaI(Tl) detectors. As the incident photon energy decreases, the ratio increases due to higher fraction of the photo electric absorption. At a fixed photon energy, smaller detectors give lower ratio since the escape probability of the scattered photons are higher.
A simple method used for the peak area determination is shown in Fig. 5.25. If the continuum is not present, although hard to be encountered, in the spectral region where the peak is located, the peak area is simply the integral of the count in this region. If a linear continuum is formed below the peak, the continuum area can be subtracted linearly as shown in the figure. A model description of a peak is a Gaussian (Fig. 5.26). In this case, both Gaussian peak and linear background are fitted simultaneously with the least square fitting method.

Fig. 5.25. Peak area analysis by a simple integration subtracted by a linear background.

Fig. 5.26. Peak fitting with a Gaussian combined with a linear background.
5.4. Position-sensitive scintillation detectors in medicine

There are two techniques used for imaging patients using ionizing radiation as shown in Fig. 5.27 [11,12]. Transmission imaging is usually carried out with X-ray beam while radionuclides are used for emission imaging.

The important features of the two methods are:

<table>
<thead>
<tr>
<th>Feature</th>
<th>X-ray imaging</th>
<th>Gamma-ray imaging</th>
</tr>
</thead>
<tbody>
<tr>
<td>Photon flux at the detector surface ((\text{mm}^{-2}\text{s}^{-1}))</td>
<td>(10^7)</td>
<td>(10^{-1})</td>
</tr>
<tr>
<td>Dynamic range ((10^3-10^4 \text{ events}))</td>
<td>Upto (5\times10^4 \text{ event/s})</td>
<td>(70-511 \text{ keV})</td>
</tr>
<tr>
<td>Photon energy ((10-125 \text{ keV}))</td>
<td>(70-511 \text{ keV})</td>
<td>(70-511 \text{ keV})</td>
</tr>
<tr>
<td>Spatial resolution ((5-15 \text{ line pairs/mm}))</td>
<td>(0.05-0.25 \text{ line pairs/mm})</td>
<td>(10-20 \text{ min})</td>
</tr>
<tr>
<td>Exposure times ((0.01-1.0 \text{ s}))</td>
<td>(10-20 \text{ min})</td>
<td></td>
</tr>
</tbody>
</table>

A. X-ray imaging detectors

Film detector [13]:

The most common type of the analog detector for X-ray imaging is screen-film cassette, which consists of intensifying screen (a phosphor, or scintillation material), a light sensitive film, and a
cassette holder as shown in Fig. 5.28. The material of the intensifying screen should have high X-ray absorption cross section, and Gd based (Gd$_2$O$_2$S, gadolinium oxysulfide) or tungsten-based (CaWO$_4$, calcium tungstate) phosphors have been widely used. X-ray interactions in the screen cause it to scintillate and emit visible light, therefore, intensification of imaging is provided by the screen. However, the light is emitted in all directions, which degrades spatial resolution. A compromise must be reached between sensitivity and spatial resolution as illustrated in Fig. 5.29. The sensitivity of films to X-rays can be increased by as much as a factor of 10 using intensifier screens. A light-sensitive film emulsion in direct contact with the screen produces a latent image, which is subsequently rendered visible by chemical processing.

**Computed radiography [14,15]:**

Computed radiography (CR) is a simple way to digitize X-ray images and is most successful. CR is based on the photostimulable phosphors (or storage phosphors). The photostimulable phosphors used are most often in the barium fluorohalide family in powder form and deposited onto a substrate to form an imaging plate. Its concept is similar to that of the intensifying screen. The plate is exposed in the same manner as film. They differ in that the useful optical signal is not derived from the light emitted in prompt response to the incident ionizing radiation.

![Fig. 5.30. Energy level diagrams for conventional phosphors and photostimulable phosphors.](image)

![Fig. 5.31. Computed radiography process.](image)

The excited electrons in the photostimulable phosphors are trapped into the crystalline structure as shown in Fig. 5.30. The electron in such an excited state can only return to the normal lower energy state by migrating through the lattice to a recombination center. In order to become free of the trapping centre about which it is localized, the electron must acquire energy. It is this inability of the electron to escape which gives rise to the metastable nature of the orbit, so that the material becomes an information storage device.

When exposure is over, by stimulating the crystal with a laser beam of appropriate wavelength (red light) as shown in Fig. 5.31, electrons are released from the traps and raised to the conduction band of the crystal, subsequently triggering the emission of shorter-wavelength (blue) light. This process is called photostimulated luminescence. The emitted light is collected and detected with a PMT, whose output signal is digitized to form the image.

The same principle is applicable to Thermoluminescent dosimeters (TLD). A record of the radiation exposure is stored in the number of excited electrons that have been created by ionizing
radiation. In this case, the number of electrons is proportional to the dose. After exposure, heating the TLD chip provides the necessary source of energy which allows the electron to leave the trapping center. The total number of photons emitted is then proportional to the radiation dose.

**Computed Tomography (CT) [11,16]:**

![X-ray CT system](image1)

In X-ray CT, the body is consecutively irradiated from a large number of directions by an X-ray fan beam as shown in Fig. 5.32. Attenuation profiles are recorded and cross-sectional images of the body are constructed. CT has made its name due to its ability to distinguish small changes in tissue properties and its ability to image in 3-D volumes of tissue. A segment of a circle, ~ 1 m, is covered with up to ~ 1,000 small detectors, which form a 1-D position sensitive detector (PSD). This PSD is rotating with the X-ray source. As detector material, inorganic scintillators like CsI(Tl) and BGO coupled to photodiodes were used past but have been replaced with polycrystalline ceramic scintillators due to radiation damage effect or bad matching with the quantum efficiency of the photodiodes. Fig. 5.33 shows a typical sensor array: the bare photodiode array, the ceramic scintillator coated array and the complete sensor housing.

**Mammography [17-19]:**

Mammography is one of the most challenging of diagnostic imaging techniques. The normal and abnormal tissues have such a tiny difference that demonstration of pathology is extremely difficult. Therefore, high spatial resolution is required and CCD-based detectors have been developed for conversion of scintillation light. A typical detector for digital mammography is illustrated in Fig. 5.34.

![Detector for digital mammography](image2)
B. Gamma-ray imaging detectors

Scintillation (Anger) camera [20]:

The scintillation camera is based on the original design made by Anger. A typical arrangement is shown in Fig. 5.35. The scintillation detector is a flat crystal with a typical size 50 × 50 cm$^2$ and a thickness of 1 cm. An array of PMTs covers the surface of the crystal. Each scintillation event generates output pulses for the PMTs located near the interaction position and the two-dimensional position of each event is deduced from the relative size of the PMT signals. The spatial resolution is usually limited to ~ 3 mm FWHM.

![Fig. 5.35. Arrangement of an Anger camera.](image)

Positron Emission Tomography (PET) [21-25]:

PET is a nuclear medical imaging technique whereby a biologically active compound (or drug) labeled with a positron emitting radionuclide ($^{18}$F, $^{11}$C, $^{13}$N, $^{15}$O) is introduced into the body. The drug then accumulates in the patient and the pattern of the annihilation radiation emissions is used for estimating the distribution of the drug. Therefore, PET can target where a certain metabolic process occurs and measure the rate of the process. A typical PET detector module with BGO scintillation crystal has 80 % efficiency, 20 % FWHM resolution, 2 ns FWHM timing resolution, 4 μs dead time, and 5 mm FWHM position resolution for 511 keV photons. The performance is limited by the scintillation crystal. Since a short attenuation length is critical for a high spatial resolution, BGO has mostly been used. Among recently developed scintillators, LSO has four times higher light output and eight times faster decay time than BGO. The attenuation length of LSO is similar, therefore, LSO is the most promising candidate which will replace BGO.

![Fig. 5.36. Principle of PET scanning.](image)

![Fig. 5.37. Conventional PET detector module (8×8 elements coupled to 4 PMTs).](image)
One of the serious problems degrading the position resolution is the parallax error (or radial elongation) as shown in Fig. 5.38. Photons incident with a glancing angle can penetrate detector layers and make a false signal. This problem become worse for the radionuclides positioned further away from the center of the ring. If the information of the depth of the interaction within the scintillation crystal is available this artifact can be reduced in a great portion by constructing a line between the interaction positions rather than the interaction crystals.

An important advance achieved recently is the **multimodality imaging systems**, where PET scans and high spatial resolution anatomic images are acquired at the same time in a single system. The most popular combination is PET with CT as shown in Fig. 5.39.

Fig. 5.38. Parallax error (or radial elongation).

Fig. 5.39. Integrated PET/CT scammers: cancer patient’s chest and mouse.
References
