CHAPTER 2

Scintillation Detectors

Learning Outcomes

1. Identify the essential steps of radiation detection using a scintillation detector.
2. Describe the delocalized bonding structure of thallium-activated sodium iodide, the emission of scintillation photons in response to absorption of a gamma ray, and the relationship between gamma ray energy and scintillation light emission.
3. Identify the parts of a photomultiplier tube and state the function of each.
4. List additional electronic components needed for a scintillation detector and the function of each.
5. Discuss why and how a scintillation detector is calibrated, for both single-channel and multichannel analyzer types.
6. Outline the causes of peak broadening, the calculation of a percent energy window and the full-width at half-maximum (FWHM), and the use of the FWHM as a quality control test.
7. List recommended quality control tests for scintillation detectors and their frequencies.

Introduction

Scintillation detectors constitute the other major class of radiation detectors used in nuclear medicine. They have significant advantages over gas-filled detectors. Because they are solid rather than gaseous, they have much greater efficiency for interactions with gamma rays compared to gas-filled detectors. This in turn provides a means to measure the energy of a radiation interaction, which allows us to identify radionuclides based on their gamma ray energies, or to distinguish unscattered from scattered gamma rays. Scintillation detectors have wide application in many processes that involve detection of gamma rays. In nuclear medicine, they are found in thyroid probes, well counters, gamma cameras, and positron emission tomography (PET) systems. Liquid scintillation counters operate on somewhat similar principles, but are rarely used in clinical nuclear medicine.

Basic Principles of Scintillation

Scintillation is a general term referring to the process of giving off light; it is used both literally and figuratively. More specifically in the sciences, a *scintillator* is any material that can release a photon in the UV or visible-light range, when an excited electron in the scintillator returns to its ground state. These *scintillation photons* are detected by a photomultiplier tube (PMT) and converted into an electronic signal. Some of the terms used to describe nuclear medicine studies, such as *scintigraphy* and *scintiscans*, derive from this aspect of the detection process.

Scintillator

Many different kinds of materials have the ability to scintillate. Organic materials, particularly conjugated ring compounds, produce scintillation photons in the process of de-excitation of orbital electrons. Scintillators can also be made of glass or of noble gases such as xenon or helium, both of which may be used for detection of particulate radiation. However, all of these scintillators have a low average atomic number, and therefore are not very efficient for interactions with gamma rays.

The scintillators used in nuclear medicine applications are inorganic crystalline scintillators, often with small amounts of impurities that help them scintillate more efficiently. The most common inorganic scintillator employed in nuclear medicine is thallium-activated sodium iodide or NaI(Tl) developed for use in radiation detection by Robert Hofstadter in 1948. It is used primarily in detectors designed to detect lower-energy gamma rays such as those emitted by Tc-99m. Detectors designed for high-energy photons, such as annihilation photons, are made of scintillators with a higher atomic number and density, such as bismuth germanate, lutetium oxyorthosilicate, and gadolinium oxyorthosilicate. These are discussed further in Chapter 14.

Scintillation crystals are made to exacting tolerances and require exceptional care in the manufacturing process. The crystal must be optically transparent, without cracks or boundaries that could cause scintillation photons to be reflected. Scintillation crystals are quite fragile, and can

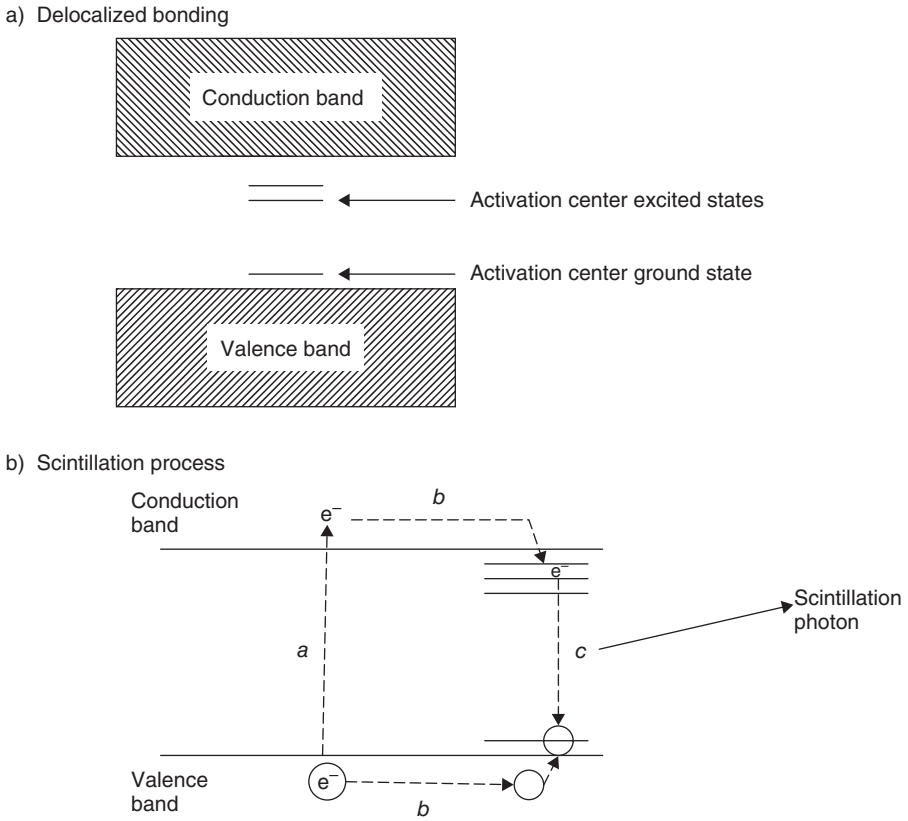

Figure 2-1 Band structure and scintillation process in thallium-activated sodium iodide. The three steps in the scintillation process are: a) some of the free electrons created by a gamma ray interaction have enough energy to jump across the forbidden gap to the conduction band; b) once in the conduction band, these electrons migrate to the activation center excited-state orbitals, and the holes in the valence band to the activation center ground-state orbitals; c) the transition of an electron from an activation center excited-state orbital to an activation center ground-state orbital produces a scintillation photon.

fracture under conditions of mechanical stress or rapid temperature change (>5°C or 9°F per hr). Extreme care must be used when working near an exposed crystal. In addition, sodium iodide is *hygroscopic*, meaning that it absorbs moisture from the air. When this happens, the crystal turns yellow and absorbs scintillation photons rather than transmitting them. As a result, scintillation crystals must be hermetically sealed, so that no air or water comes into contact with the crystal.

The most desirable characteristic of sodium iodide is an excellent *scintillation light yield* (number of scintillation photons emitted per eV of radiation energy absorbed); higher than most other scintillators. This is important because the greater number of scintillation photons leads to greater precision in measuring the energy of the absorbed gamma ray. The scintillation photons produced in sodium iodide range from 325 to 550 nm, primarily in the ultraviolet spectrum but just touching the high end of the visible-light spectrum. This range perfectly matches the response function of the bialkali photomultiplier tube (1), the most common type of PMT used in radiation detection applications.

Another important characteristic of a scintillator is its *decay time*, which indicates how long scintillation photons are released after a radiation interaction. A long decay time means that radiation events will need to be more widely spaced if we desire to count them in pulse mode. The decay time for sodium iodide is relatively long at 230 nsec, which in turn contributes to the detector's overall dead time. As with any material, sodium iodide shows decreased efficiency for interaction as gamma ray energy increases (linear attenuation coefficient = 2.16/cm at 150 keV vs 0.294/cm at 600 keV).

The Scintillation Process in NaI(Tl)

Thallium-activated sodium iodide is crystalline in physical form, with a small amount of thallium replacing sodium (about 1 in 1000 atoms) to provide the impurity. It is the delocalized molecular bonding of the crystal that gives it the ability to scintillate (**Figure 2-1**). Delocalized bonding is discussed in more detail in Appendix B; briefly, the electron orbitals of the sodium and iodine atoms combine to form a valence band and a conduction band of molecular orbitals. The valence band (analogous to the bonding orbital of a covalent bond) is at a lower energy than the conduction band (analogous to the antibonding orbital). The thallium impurities create electron orbitals between the conduction band and the valence band called *activation centers*. In the ground state, the valence band and the ground-state level of each activation center are filled with electrons. The conduction band and the excited energy levels of the activation centers are empty. Thus there is no electron movement in the absence of radiation.

When a gamma ray enters the crystal and undergoes an interaction, it produces one or more secondary electrons with high kinetic energy (see Appendix A). These electrons cause ionizations and excitations as they move, creating many low-energy electrons. Most of these low-energy electrons lose their energy as heat, but some have the right amount of energy to jump up to the conduction band (Fig. 2-1b, Step *a*). A hole is left in the valence band for each electron excited to the conduction band. The electrons that get into the conduction band are free to move around, but are not allowed to drop directly back into the valence band (this is "forbidden," meaning highly unlikely according to quantum mechanics). However, they will still seek to find the lowest energy levels available to them, which are the excited state levels of the activation centers (Fig. 2-1b, Step *b*). Likewise, the holes will allow electrons in the valence band to move, and eventually they will be filled by electrons from activation centers, leaving vacancies in the activation center ground-state orbitals. Quantum mechanics does allow the transition of an electron from the excited state to the ground state in an activation center. The electron drops down and fills the hole, with the release of the excess energy in the form of a scintillation photon (Fig. 2-1b, Step *c*).

Let us consider two important consequences of this process. First, of the energy deposited in the scintillation crystal by the gamma ray, some but by no means all of it is transferred to scintillation photons. (As shall be discussed below, the detection of radiation by scintillation is an inherently inefficient process. Sodium iodide, with a scintillation efficiency of 13%, is considered a highly efficient scintillator.) From the standpoint of radiation detection, the critical fact is that a proportionality exists between the *energy* that the gamma ray deposits in the crystal and the *number* of scintillation photons that are produced. A 20-keV gamma ray will produce one-fifth as many scintillation photons as a 100-keV gamma ray. Thus, if we count the scintillation photons, and if we carry this proportionality through the rest of the detection process, we can determine the energy of each gamma ray.

The other important fact is that this all happens very quickly. Because the gamma ray and the secondary electrons it creates are traveling very fast, the process of absorbing the gamma ray's energy takes only a few nanoseconds. The emission of scintillation photons reaches a maximum at about 30 nsec, and declines thereafter with a half-life of about 150 nsec, such that two-thirds of the scintillation light is emitted within 230 nsec. As long as the count rate stays relatively low, the interaction of one gamma ray in the crystal will be completed before the next one occurs. Thus, scintillation detectors used in nuclear medicine generally operate in pulse mode.

Photomultiplier Tube

Having scintillation photons is not enough. We need to convert them into an electronic signal. This is accomplished by a PMT. The thyroid probe and well counter each have one PMT. Gamma cameras have many PMTs, allowing position as well as energy to be determined. A diagram of a PMT is shown in **Figure 2-2**. In order to understand its function, we need to consider each of its parts.

The *photocathode* is closest to the scintillation crystal; it converts scintillation photons into electrons. It is made of a very thin layer of material that is *photoemissive* (i.e., emits electrons when exposed to UV and visible-light photons). Photoemissive materials used in PMTs are usually metal alloys that have extra electrons, usually bialkali antimonide compounds such as K_2CsSb and Na_2KSb. As scintillation photons are released in the scintillation process, some will strike the photocathode and cause it to emit electrons that are released into the vacuum space of the PMT itself. Once the electrons are emitted from the photocathode, the *focusing grid* provides the proper electric field to direct the electrons toward the first dynode. Note that a vacuum must be maintained inside the PMT, so that no electrons are lost due to interaction with air molecules. The PMT must also be shielded against stray magnetic fields, as these will alter the trajectory of the electrons (1).

A series of *dynodes* multiply the small number of electrons generated at the photocathode into a measurable electronic signal. They are made of metal alloys (again with an excess of electrons) and are maintained at positive electrical potentials, each one higher than the last. They are curved or positioned so as to direct the emitted electrons toward the next dynode. The absorption of one electron at a dynode causes emission of three to six electrons, depending on the potential difference from dynode to dynode. The electrons released at each dynode are accelerated toward the next dynode, again because of the potential difference between dynodes. The process continues for 9 to 12 dynodes, for a total multiplication factor of about a million, producing a small but measurable electronic signal. Finally, the *anode* collects all the electrons produced from the final dynode and emits an output voltage signal.

The high voltage supply of the PMT determines the potential difference between the dynodes, and therefore the final size of the PMT output pulse. A typical scintillation detector has a high voltage of 1000 volts (V) or more, with 100-V increments between dynodes. The exact setting of the high voltage is determined each day using the calibration procedure discussed later in this chapter. Scintillation detectors are very sensitive to changes in the high voltage; a 1% change in the high voltage results in a 10% change in the pulse size (measured at the anode). The high voltage supply must therefore be steady and well regulated.

To recapitulate the process of radiation detection in a scintillation detector: A gamma ray interacts in the scintillation crystal to bump electrons into the conduction band, which produce scintillation photons as they transition back to ground state. The scintillation photons in turn cause the emission of a few electrons at the photocathode of the PMT. These few electrons are converted to a measurable electronic signal at the PMT's anode. The height of the pulse generated in the PMT is therefore proportional to the energy absorbed in the scintillator from the gamma ray interaction. Note, however, that three

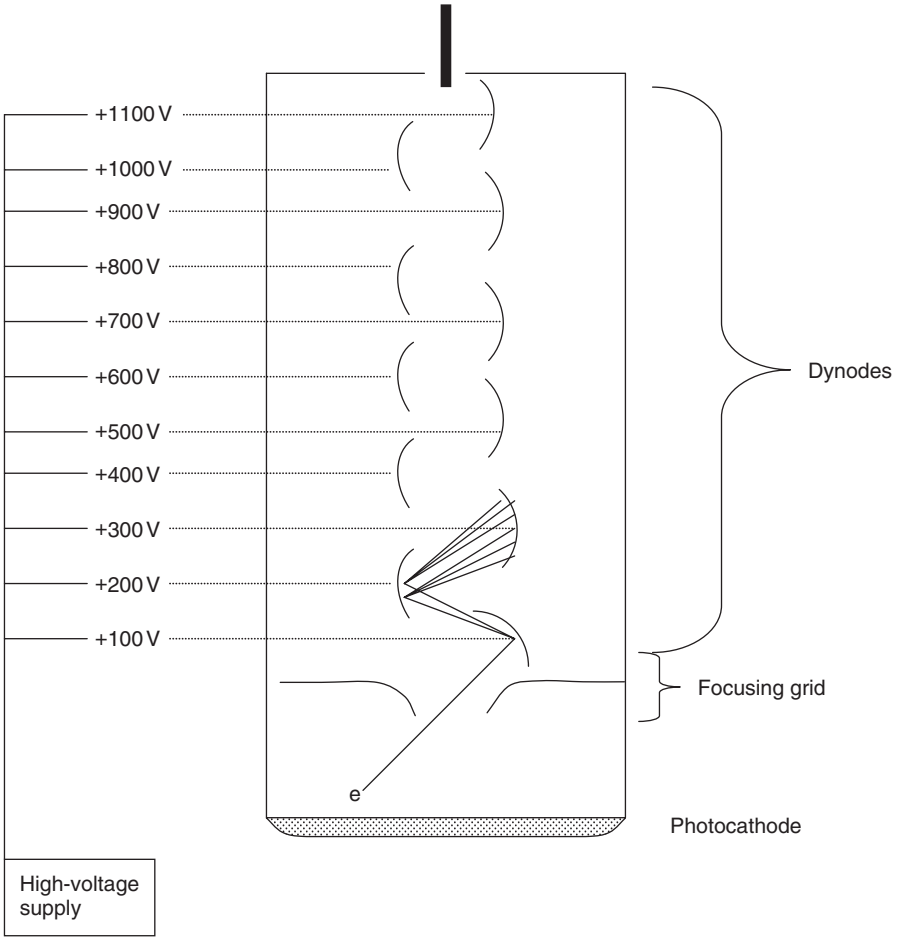

Figure 2-2 Photomultiplier tube. Its basic structure is an evacuated cylinder enclosed in glass, with a photocathode on one end, an anode at the opposite end, and small curved dynodes in between. The electrical potential to the dynodes is what causes multiplication of the electrical signal created at the photocathode. The passage of an electron through the focusing grid, its interaction with the first dynode, and its multiplication at the second dynode is shown, using a multiplication factor of 3.

conversions between energy (photons) and mass (electrons) are required. With each conversion step, there is some inefficiency, and therefore a loss of energy. Consider the process for Tc-99m and Cs-137 (**Table 2-1**). The number of information carriers (the scintillation photons or electrons in each step) decreases with each conversion between mass and energy, a fact that has important implications.

Looking at Table 2-1, we can see that the number of information carriers decreases as the process moves toward the photocathode, and then increases as the dynodes multiply the signal beyond the photocathode. However, the number of information carriers is approximate, not exact, and is subject to statistical variation. The point of greatest statistical variation is found at the point in the process where the number of information carriers is the lowest, namely the release of electrons from the photocathode. As a result, the measured output (the pulse height) is not always exactly proportional to the energy of the photon creating the signal, but has random fluctuations from one gamma ray to the next. For the total absorption of a Tc-99m gamma ray, we do not measure a single pulse height corresponding to exactly 140 keV. Instead, we get a distribution of pulse heights with a Gaussian shape, centered at 140 keV (**Figure 2-3**). This *peak broadening* prevents the separate detection of gamma rays with closely spaced energies. It has important implications for the use of scintillation detectors in nuclear medicine, as is discussed below.

Associated Electronics

Figure 2-4 shows the scintillation crystal and PMT in a block diagram of a scintillation detector. The scintillation crystal sits in an aluminum "can," which has an inner surface coated with a reflective layer to reflect scintillation photons back into the crystal. The back end of the crystal may be optically coupled to a light pipe made of quartz or Lucite, which is in turn coupled to the PMT. The light pipe acts as a guide for the scintillation photons, internally reflecting wide-angle scintillation photons toward the photocathode. Optical coupling grease (usually some kind of silicon grease) reduces the loss of scintillation photons by preventing reflection at the scintillation crystal/light pipe and light pipe/photocathode interfaces (or the crystal/

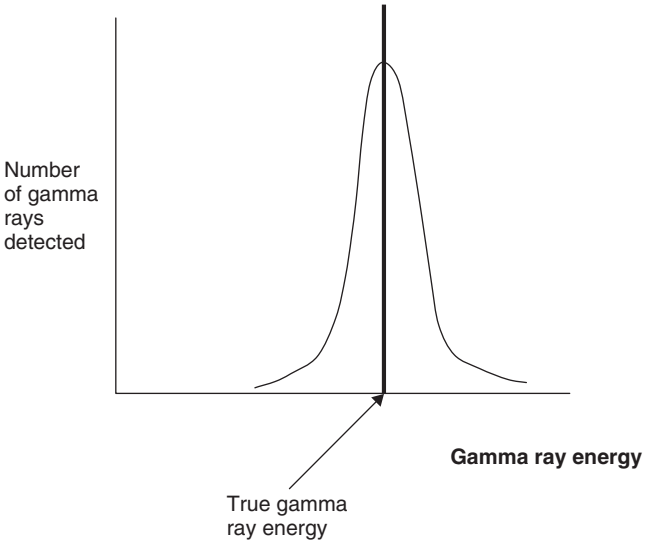

Figure 2-3 Peak broadening as seen with scintillation detectors. The straight line is what would be seen with a detector that translates gamma ray energy into an electronic signal with no statistical variability; the Gaussian curve is seen with scintillation detectors because of the statistical variation that occurs in the detection process.

photocathode interface, in systems without a light pipe). If the coupling grease degrades, the number of scintillation photons reaching the PMT is greatly diminished.

Signal Amplification

While measurable, the signal output of the PMT anode is still small. A *preamplifier* (attached directly to the PMT, to minimize signal distortion) boosts the strength of the PMT output signal to allow its passage through the remaining detector electronics. In order to facilitate complete charge collection, the preamplifier is adjusted to have a long decay time, resulting in a pulse with a long tail, requiring about 50–100 μsec to return to baseline. An *amplifier* (housed in an accompanying electronic module) then amplifies the preamplifier output signal by a variable amount. Multiplication factors in amplifiers are typically between 1 and 100, and are adjusted via the coarse and fine gain controls or the energy multiplication knob on the electronics module. The amplifier also shapes the pulse and shortens it to provide a series of discrete pulses that are easier to analyze (see Appendix B for further details).

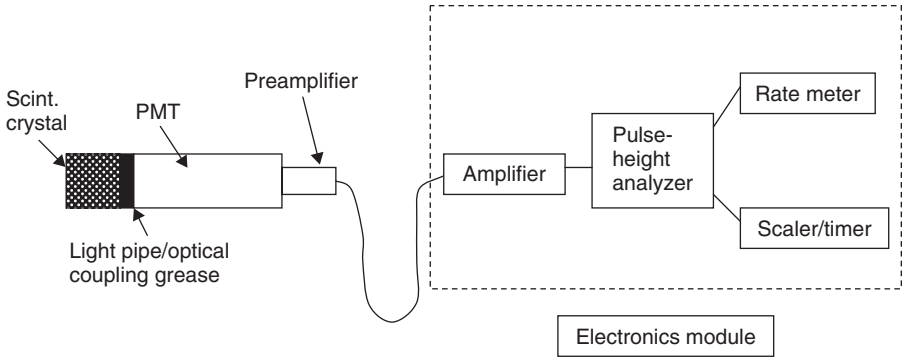

Figure 2-4 Block diagram of a scintillation detector. The detector/PMT assembly is usually separated from the electronics module by a cable.

Table 2-1
Scintillation Detector Efficiency

Site in Detector	Conversion Factor	Information Carriers for Tc-99m	Information Carriers for Cs-137
		140 keV photon	662 keV photon
Scintillation crystal	Scintillation efficiency = 12%	16.8 keV converted to scintillation photons	79 keV converted to scintillation photons
Scintillation photons	Each scintillation photon has energy of 3 eV	5600 scintillation photons	26,480 scintillation photons
PMT at front of photocathode	75% absorption at photocathode	4200 scintillation photons absorbed	19,860 scintillation photons absorbed
PMT after photocathode	Photocathode efficiency = 20%	840 electrons emitted into PMT space	3972 electrons emitted into PMT space
Dynodes	Multiplication factor = 10^8	84 trillion	400 trillion

Adapted from: Knoll GF. *Radiation Detection and Measurement*. 3rd ed. New York: Wiley and Sons; 2000:330.

Determination of Energy Acceptability

Next, the *pulse height analyzer* (PHA) determines the amplitude of the pulse, which correlates with the gamma ray's energy. There are two types of PHAs, the single-channel analyzer (SCA) and the multichannel analyzer (MCA). An SCA consists of a lower-level discriminator (LLD), an upper-level discriminator (ULD), and an anti-coincidence logic circuit. The values of the LLD and the ULD are set by the operator and are related to the energy of the photons being measured, as discussed below. Each pulse in turn is compared to the LLD and the ULD, and pulses with amplitude greater than the LLD but less than the ULD, as detected by the anticoincidence circuit, produce PHA output pulses. All output pulses from the PHA are the same size, and are sent on to the scaler and the rate meter. The SCA thus essentially extracts the gamma ray energy information from the voltage signal and, based on that energy, either accepts or rejects each pulse. Its output therefore is a string of logic (yes/no or 1/0) pulses that can be counted, but which contain no additional information about the gamma rays they represent beyond the fact that those gamma rays fell within the preset energy range.

Energy Spectrum

The chief drawback of an SCA is that it rejects radiation events that do not fall between the LLD and the ULD, thus preventing us from visualizing a large amount of potentially useful information. An MCA shows all radiation events on an energy graph or *pulse-height spectrum* (**Figure 2-5**). Most MCAs work by digitizing the PMT output, and assigning the event to one of a series of predefined bins or channels based on the signal's size. We can think of an MCA as consisting of a series of PHAs, each representing a small portion of the energy spectrum. Each gamma ray is recorded as a "1" in one PHA and a "0" in all of the other PHAs.

In the pulse-height spectrum of Figure 2-5, each column or bin on the *X*-axis represents a specific narrow range of pulse heights (note that height is represented on the *X*-axis), which in turn represents event energy. The *Y*-axis represents the number of events with energy in that range. The spectrum has several identifiable peaks. The right-most peak in the figure corresponds to events in which the entire gamma ray energy is registered in the scintillation crystal. (The identification of the peaks of an energy spectrum are discussed in Chapter 4.) This peak is called the *photopeak*, because it corresponds to gamma rays that have undergone photoelectric absorption in the scintillation crystal. Counts to the left of the photopeak represent gamma rays that did not deposit all of their energy in the scintillation crystal. Most have undergone a Compton interaction and are considered "scattered" photons.

The photopeak is used for the calibration procedure described below and for most counting applications. An energy range encompassing the photopeak, shown as the shaded area, contains "acceptable" events that fall within the energy window and hence generate a PHA output pulse. The limits of the shaded area thus correspond to the LLD and the ULD of the SCA. Similar energy spectrum information can be obtained from an SCA, by manually changing the LLD and the ULD to determine the counts at each energy interval, and then plotting the spectrum by hand.

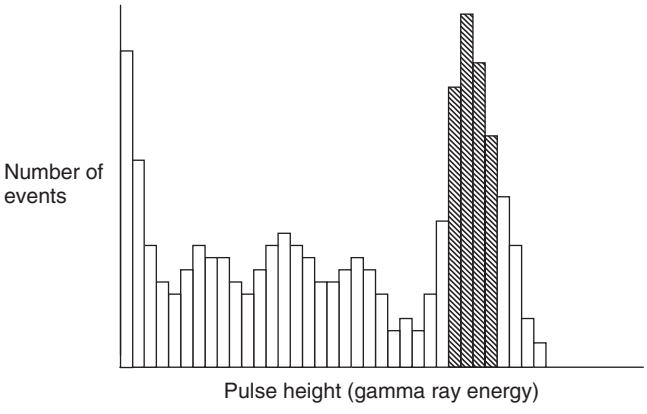

Figure 2-5 Output of a multichannel analyzer. The size or pulse height of each event is plotted on the *X*-axis, and the number of events at each pulse height provides the *Y*-axis value. The shaded bars of the histogram are those within the chosen energy range.

Tracking Counts

The last two devices in the block diagram are the scaler/timer and the rate meter. The *scaler/timer* records accepted PHA output pulses during a specified time interval. (The term *scaler* stems from the days when digital registers could not go very high, requiring a scaling circuit to divide the incoming pulse rate by a fixed factor (e.g., 100 or 1000) so that the register could record all the pulses.) A *rate meter* determines the average current produced by the SCA pulses, which drives a meter calibrated in average counting rate.

Pulse Shapes

Let us consider the block diagram from the standpoint of the pulse shapes at each point in the process (**Figure 2-6**). The PMT output is a very small pulse. The preamplifier, in matching the PMT output pulse to the impedance of the detector electronics, widens it considerably, to about

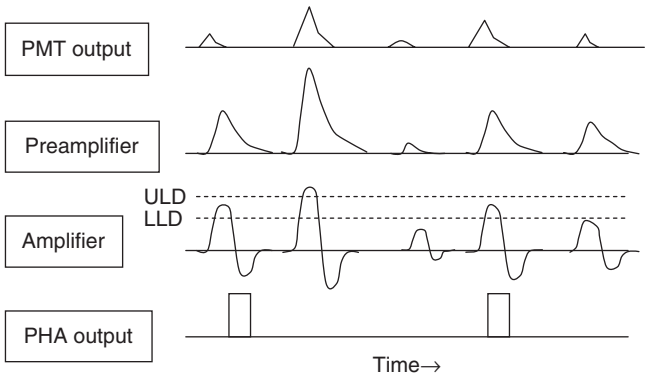

Figure 2-6 Pulse shapes in a scintillation detector. The amplifier line shows the LLD and the ULD levels superimposed on the amplifier output pulses.

2–5 μsec. The amplifier amplifies and shapes the pulse to make it narrower, usually with a negative overshoot. The PHA output line shows that the PHA creates an output pulse for every input pulse of acceptable height. All of the PHA output pulses are the same height. This sequence works well when the count rate is relatively low, but creates problems called baseline shift and pulse pile-up when counting rates are high (see Chapter 4).

Calibration

This section may seem to some superfluous, as most modern scintillation detector systems operate on a "push-button" basis, and many are self-calibrating. However, the author firmly believes that technologists should understand the operation of the machines they work with. The text therefore discusses the energy window determination and calibration of both SCAs and MCAs; the reader can take what he or she needs about the details.

Single-Channel Analyzer

Older scintillation detector systems have two knobs located on the electronics module labeled "threshold" and "window." In relation to our prior discussion, the threshold corresponds to the LLD. The window sits on top of the threshold, so that the value of the ULD is equal to the threshold plus the window. *Calibration* refers to the correlation of these knobs or settings with gamma ray energy in keV. This is typically done with Cs-137, which has a long (30-year) half-life and emits 662-keV gamma rays. The essence of calibration, for single-PMT systems, is to adjust the high voltage so that the photopeak sits in the middle of the window. Once calibrated, the high voltage is locked in, and the threshold and window are changed as needed for clinical measurements.

The threshold and window knobs on an SCA are typically potentiometers with 1000 distinct settings. It would be highly advantageous to calibrate the detector so that each possible setting on the potentiometer is equal to 1 keV. We first must set the threshold and window to appropriate calibration values. For Cs-137, if a 3% (20-keV) window centered on the 662-keV photopeak is desired, we would set the threshold knob to 652 and the window knob to 20. Once properly calibrated, this would provide a convenient one-to-one correlation between the potentiometer settings and the gamma ray energy.

Once the threshold and window are set, we must next adjust the high voltage so that the 662-keV photons actually register in the SCA window we have set (**Figure 2-7**). To start from scratch, the voltage is turned down to a low level and the Cs-137 source placed in (or in front of) the detector. Then the high voltage is slowly increased, causing, the height (or voltage) of each pulse to get larger (recall that pulse height is plotted along the *X*-axis of the pulse-height spectrum). When the count rate in the window reaches a maximum, the bell-shaped curve is centered on the window, and 662 on the detector dial is equivalent to 662 keV of photon energy. (Note that if we continue to increase the high voltage beyond the maximum, the number of counts will drop but will not reach zero, because scattered photons will be registered in the window. Thus, always start with the high voltage below its expected value.)

Multichannel Analyzer

An MCA is easier to calibrate than an SCA because we can see the whole range of gamma ray energies. We need only tweak the high-voltage knob until the highest point of the photopeak corresponds to a specified bin. However, MCAs often don't have a scale that is easily correlated to photon energy in a one-to-one fashion. They may require a conversion factor (e.g., channel 150 = 662 keV) in order to determine the energy of any other channel.

Energy Linearity

An important quality of a scintillation detector is its ability to measure high vs low energy gamma rays in a proportional manner, a concept called *energy linearity* (compare this definition to activity linearity in Chapter 1). If a detector properly calibrated with Cs-137 demonstrates good energy linearity, it is able to register any other gamma ray energy at the proper pulse-height setting. For example, if SCA knobs are calibrated such that a window centered on 662 corresponds to 662 keV, then a window centered on 140 should register maximum counts with a Tc-99m source. In essence, what we have done by calibration is to set our scale, and the detector's energy linearity characteristics determine how well that scale applies throughout the energy range of interest (**Figure 2-8**). In fact, sodium iodide and other alkali halide scintillators do demonstrate some nonlinearity, mainly at the low end of the energy scale, below about 200 keV (1). For most nuclear medicine applications, this nonlinearity will be small, but its presence should be recognized and measured at least occasionally (2).

Scintillation Detector Applications

Detector Operation

Once the scintillation detector has been calibrated, it is typically used to make a timed measurement of a radioactive source. The output is a number of counts, each count representing a gamma ray whose total energy deposition in the scintillation crystal fits within the PHA window. A typical measurement has hundreds or thousands of counts, for a typical counting time of 0.5–10 minutes.

A single measurement thus produces a single value for the amount of radioactivity present in the source. This value is generally expressed in a rate format of counts per min (cpm). A repeat measurement of the same source in the same geometry will usually produce a different value, due to the statistical nature of radioactive decay. We can use this inherent variability to our advantage, as seen in Chapter 4. In clinical situations, a measurement may be repeated to verify that the procedure was performed correctly. If the two readings are similar, they might then be averaged to generate the final result. Two readings that are wildly different should prompt the operator to consider whether the procedure was done correctly.

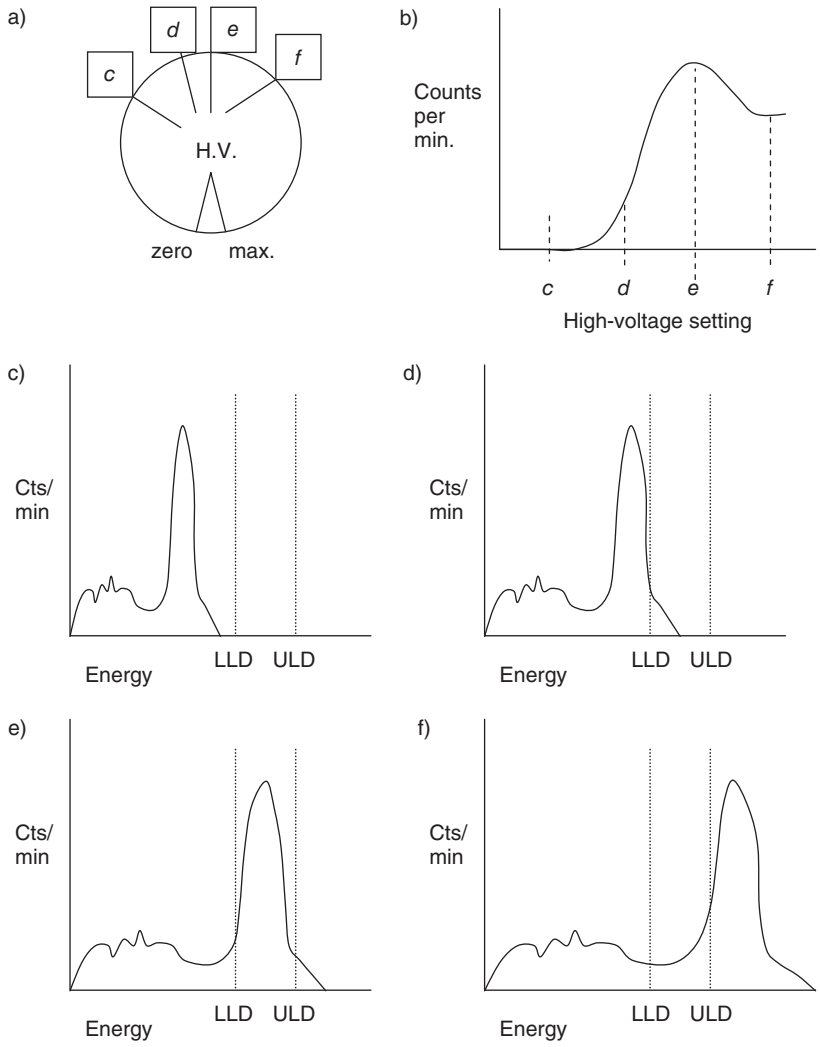

Figure 2-7 Calibration of a scintillation detector. Figure 2-7a shows the high-voltage (or calibration) knob with four settings corresponding to the four energy spectra in Figure 2-7c–2-7f. Figure 2-7b shows the detector count rate at each knob setting, with the letters corresponding to the pulse-height spectra shown in Figures 2-7c–2-7f. The operator-determined LLD and ULD are shown superimposed on the energy spectra. As the high voltage is increased, the sizes of all pulses increase, and the energy spectrum is "stretched out" along the *X*-axis. The detector is correctly calibrated in Figure 2-7e, when the number of counts registered in the LLD-ULD window reaches a maximum. In Figure 2-7f, the high voltage has increased the pulse size beyond the center of the window; note that the measured count value does not decrease all the way to zero. Adapted from: Christian P. "Radiation Detection." In Fahey FH, Harkness BA, eds. *Basic Science of Nuclear Medicine* CD-ROM. 2001. Reston, VA: Society of Nuclear Medicine. Reprinted with permission.

Configurations

Scintillation detectors with a single PMT are made in two specific configurations: the probe configuration and the well configuration. These are diagrammed in **Figure 2-9**, with common dimensions given. The probe configuration is designed for external counting of radioactivity in organs such as the thyroid gland. A cylindrical crystal is optically coupled to a single PMT, which is shielded from magnetic fields with a mu-metal housing. A wide-bore lead collimator called a *flat-field collimator* is used to shield the scintillation crystal from stray gamma rays and protect it from mechanical damage. This geometric arrangement allows the probe to accurately measure a small amount of radioactivity, while still being directional enough to give meaningful measurements of individual organs. Appendix E on collimator mathematics discusses the calculation of organ-detector distances when using a flat-field collimator. The operator should bear in mind the possibility that the probe could be detecting radiation coming from sources other than the patient; for example, a radioactive patient in the next room, or a source left unshielded and in the probe's "line of sight." Alternative causes for an unexpectedly high value should be explored.

Most commercially available thyroid probes include a positioning rod, so that patients can be positioned at a reproducible distance from the scintillation crystal. These rods can be made from a variety of materials. One manufacturer uses graphite and states, "up to 10% attenuation can occur if the rod obstructs the field of view of the capsule from the crystal surface" (2). Good practice requires that the rod be moved out of the field of view after positioning is verified.

The well arrangement is designed to measure samples of radioactive materials in test tubes. The well is formed

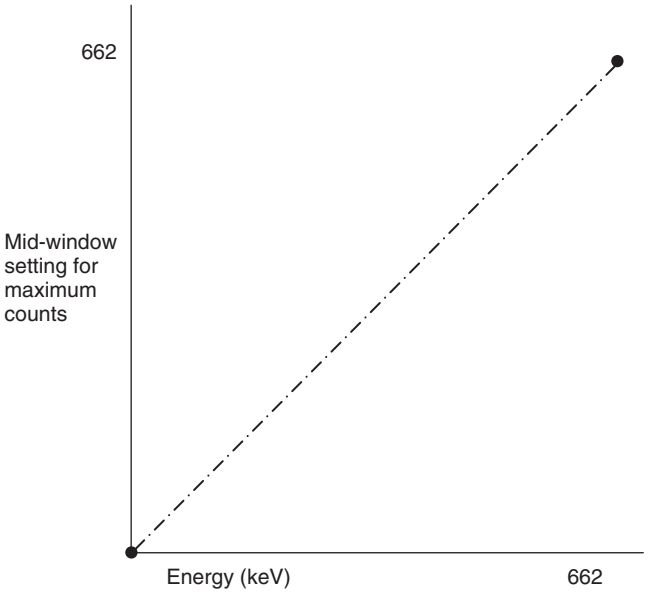

Figure 2-8 Graph illustrating energy linearity. The calibration point at (662, 662) is determined by the calibration procedure, and energy linearity assumes a straight line between this point and (0, 0). It is up to the operator to determine whether a specific scintillation detector is linear with respect to energy; that is, if the detector actually behaves as indicated by the line between 0 and 662 keV.

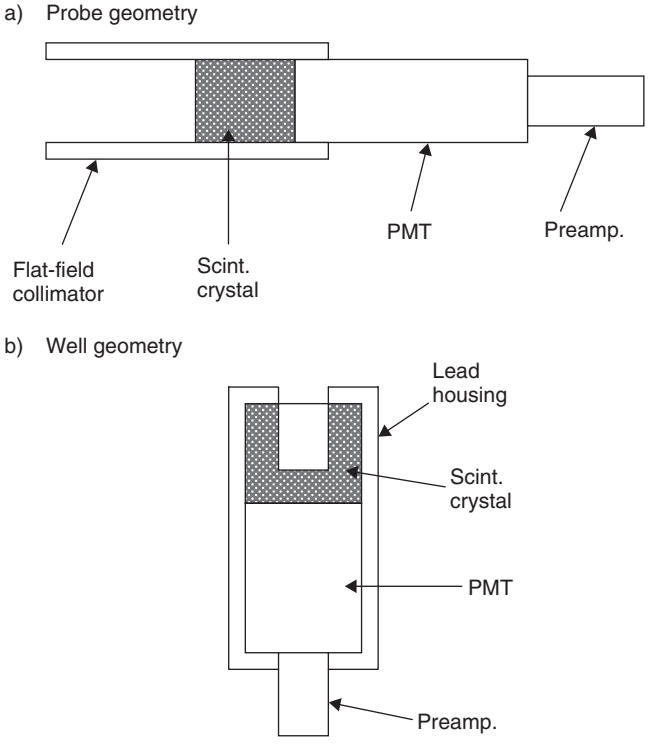

Figure 2-9 Two common scintillation detector configurations. In both detectors, the scintillation crystal is a right cylinder about 2 in. in diameter and 2 in. in height; the well detector has a ¾-in. diameter well cut into the cylinder.

out of the middle of a cylindrical scintillation crystal, and the PMT is attached to the opposite end of the crystal. This allows for excellent detection efficiency, because a radioactive source in the well is almost completely surrounded by the scintillation crystal, so that more than 90% of emitted photons are absorbed. Well counters work excellently for small sample volumes with low activity (less than about 1 μCi [37 kBq]).

Commercially available scintillation detectors used in nuclear medicine often come as a set, with a probe and a well counter and a single electronics module. While this is a cost-effective arrangement, it requires the user to be aware of several facts. First and most obviously, one must be sure that the correct detector is selected. Secondly, because they are two separate scintillation detectors, each one must be calibrated before use. Finally, due to their differences in geometric efficiency (see Chapter 4), one should not expect that measurement of the same source will produce similar results with each detector. While all of these facts may be obvious to the experienced user, they have been known to trip up even the best technologists.

Peak Broadening and Energy Resolution

Sources of Peak Broadening

Recall that the gamma rays from a given nuclear transition are always monoenergetic. For example, all Tc-99m gamma-2 photons are emitted from the nucleus with energy of exactly 140.5 keV. However, the detector registers the photons as having slightly different signals, and hence slightly different energies. The actual spectrum seen with a sodium iodide detector, instead of consisting of a sharp line at 140.5 keV, shows a broad peak. *Peak broadening* occurs because the gamma ray energy is not reproduced exactly in the detector electronics. The reasons for peak broadening include statistical variations in electron production at the crystal, nonuniformity of photocathode and/or PMT sensitivity, statistical variations in electron production at the photocathode, statistical variations in dynode multiplication, energy nonlinearity, electrical noise, and high voltage fluctuations. As can be seen in Table 2-1, it is at the photocathode that the number of information carriers reaches its lowest value, and thus the statistical variation in the number of electrons produced at the photocathode causes the majority of the peak broadening seen with sodium iodide.

Percent Energy Window

We deal with peak broadening in nuclear medicine by setting the detector to look at a range of energies around the photopeak. Because we are generally using a single radionuclide with a photopeak that is adequately separated from the other peaks in the energy spectrum, we can set up an *energy window* to include most of the photopeak and exclude everything else. Window widths are specified as a percent of the photopeak energy. A narrow window (2–5%) should be used for calibration, and a wider window (15–20%) for imaging and other measurements. Some scintillation detectors have a "Window In/Out" toggle. When the toggle is "Out," all events above the LLD are counted. This is used when counting wipe test swabs for radioactive contamination.

Sample Calculation 2-1 Calculation of a Percent Energy Window

Calculate a 20% window centered on the 140-keV photopeak of Tc-99m.

Twenty percent of 140 keV is 28 keV. Since the photopeak is to be centered on 140 keV, we need to put 28 ÷ 2 = 14 keV on each side of 140 keV (we could also call this a ±10% window). Thus:

Threshold (lower level discriminator) = 140 − 14 = 126 keV

Window = 28 keV

Upper level discriminator = 140 + 14 = 154 keV

Energy Resolution

Peak broadening is also used as a performance measure for scintillation detectors. *Energy resolution* evaluates how well a given detector distinguishes between gamma rays of closely spaced energies. (*Note*: a very important concept in imaging is spatial resolution, the ability of an imaging device to separate objects close together in space. Throughout this text, energy resolution will be specifically identified and "resolution" [without further qualification] will refer to spatial resolution.) We can quantify the energy resolution by calculating the FWHM, which measures the width of the photopeak relative to the photopeak energy. This is done as follows (**Figure 2-10**):

- Obtain an energy spectrum using Cs-137.
- Measure the height of the photopeak in counts.
- Find the line on the *Y*-axis corresponding to half of the peak's height.
- Measure the full width of the peak in keV at this height.
- Divide the width by the photopeak energy to get the FWHM.

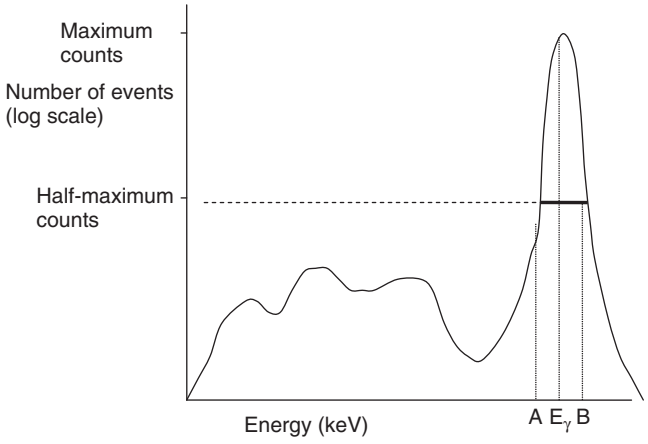

Figure 2-10 Determination of the FWHM of a scintillation detector. Note that A, B, and Eγ must all be in like units (either keV or channel number on a multichannel analyzer).

$$\text{FWHM} = \frac{B - A}{E_\gamma} \times 100\% \qquad (2\text{-}1)$$

The FWHM can be followed over time to detect changes in detector performance. A significant change in energy resolution should prompt a service call.

Sample Calculation 2-2 Full-Width at Half-Maximum

On an MCA, the photopeak for Cs-137 is found in channel number 380, and the half-height channels are numbers 365 and 396. Calculate the FWHM.

We can work either in energy units (keV) or channel numbers, but not both in the same calculation. To put everything into keV, convert the half-height channels to keV using the conversion factor 662 keV = 380 channels.

$$365 \text{ channels} \times \frac{662 \text{ keV}}{380 \text{ channels}} = 636 \text{ keV}$$

$$396 \text{ channels} \times \frac{662 \text{ keV}}{380 \text{ channels}} = 690 \text{ keV}$$

$$\text{FWHM} = \frac{690 \text{ keV} - 636 \text{ keV}}{662 \text{ keV}} \times 100\% = 8.16\%$$

Or we can do the entire calculation in channel units:

$$\text{FWHM} = \frac{\text{channel } 396 - \text{channel } 365}{\text{channel } 380} \times 100\% = 8.16\%$$

Variation of FWHM with Energy

Peak broadening is primarily due to the statistical variations in the number of electrons created at the photocathode, and higher-energy gamma rays create more electrons at the photocathode than do lower-energy gamma rays (Table 2-1). Therefore, the FWHM of a given detector will be larger for lower-energy photopeaks than for higher-energy photopeaks. Sodium iodide scintillation detectors have energy resolution in the range of 6–8% FWHM for Cs-137, and 10–12% for Tc-99m (3). Because of this variability, Cs-137 is always used to compare energy resolution between detectors.

Quality Control

Daily Tests

Scintillation detectors should be calibrated each day the detector is used (2). This ensures that the high voltage is multiplying the gamma ray signal by an amount that is appropriately related to the settings of the detector. The high-voltage setting should be nearly constant from day to day; a fluctuating value may indicate an unstable high-voltage supply, while a gradual change may signal a loss of optical coupling or hermetic seal.

As a part of the calibration procedure, a sensitivity or *constancy* measurement is made. When a long-lived source

Table 2-2
Quality Control Requirements for Well Counters and Thyroid Probes

Test	Recommended Frequency
Calibration (peaking)	Daily[1,2]
Constancy/sensitivity	Daily[1,2]
Energy resolution	Quarterly/annually[1,2]
Chi-square test	Monthly/quarterly[1,2]
Detector efficiency	Annually[1]

1. Hall A. A review of the Captus® 3000 thyroid uptake module. Capintec. Fall 2006 newsletter.
2. Graham LS, ed. *Nuclear Medicine: Self-Study Program II: Instrumentation*. Reston, VA: Society of Nuclear Medicine; 1996:8–9.

like Cs-137 is used, the cpm should be about the same from day to day (assuming that the energy window and the geometric arrangement of the source and the detector are the same). A change of 10% or more indicates a significant high-voltage fluctuation or other malfunction (4). Calibration and constancy checks will detect most of the common errors associated with scintillation detectors. A background measurement, also done on a daily basis, will ensure that there are no extraneous sources of radioactivity or contamination of the detector.

Infrequent Tests

The energy resolution should be checked at least quarterly; an increasing value indicates a problem with electronic noise, decoupling of the crystal and the PMT, yellowing of the crystal, or a cracked crystal (5). With an MCA, the FWHM is easily checked in conjunction with daily calibration. The measured FWHM should be less than 10%, using Cs-137 as the source (4). Another quality control measure that should be checked quarterly is the *chi-square test*, a measurement of the statistical variability of the detector. See Chapter 4 for a complete explanation of this test. Finally, a detector that is being used to determine absolute radioactivity in μCi or kBq must have an efficiency determination on at least an annual basis. Efficiency correlates measured count rate to absolute activity, using a radioactive source of known activity strength, and is also explained in Chapter 4. **Table 2-2** summarizes these tests, and **Table 2-3** provides some suggestions for troubleshooting.

Summary

A scintillation detector provides us with more information than a gas-filled detector, because it has the ability to determine the energy of the interacting gamma ray. The resulting ability to perform energy discrimination improves the quality of the information available, be it in identifying different radionuclides or discriminating between unscattered and scattered gamma rays. This energy discrimination depends on the proportionality of the information carriers through the scintillation crystal, the PMT, and subsequent electronics, and we can correlate it to an energy scale in keV using the calibration procedures described.

The development of the sodium iodide scintillation detector was an important milestone for nuclear medicine as well as for high-energy physics. Thyroid probes and well counters are important clinical tools, and the technologist must understand the underlying principles as well as their operation. Otherwise, incorrect numeric values and therefore incorrect clinical diagnoses may result.

Table 2-3
Troubleshooting for Scintillation Detectors

Problem	Possible Causes
Lower-than-expected constancy reading	• Incorrect high-voltage setting/incorrectly calibrated instrument • Incorrect threshold and/or window setting • Incorrect source-detector geometry (farther away than usual) • Toggle switch pointing to the other detector
Higher-than-expected constancy reading	• Extraneous radiation source in detector's active volume • Incorrect source-detector geometry (closer than usual) • Contamination of detector or housing
Widening FWHM	• Decoupling of the PMT from scintillation crystal • Crystal damage

Adapted from: Knoll GF. *Radiation Detection and Measurement*. 3rd ed. New York: John Wiley and Sons; 2000:331.

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1. Knoll GF. *Radiation Detection and Measurement*. 3rd ed. New York: Wiley and Sons; 2000:233, 234, 286–287, 330, 331.
2. Hall A. A review of the Captus® 3000 thyroid uptake module. Capintec. Fall 2006 newsletter.
3. Cherry SR, Sorenson JA, Phelps ME. *Physics in Nuclear Medicine*. 3rd ed. Philadelphia: Saunders; 2003:159.
4. Graham LS, ed. *Nuclear Medicine: Self-Study Program II: Instrumentation*. Reston, VA: Society of Nuclear Medicine; 1996:8–9.
5. Henkin RE, Boles MA, Dillehay GL et al., eds. *Nuclear Medicine*. St. Louis, MO: Mosby-Year Book; 1996:82.

Additional Resources

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Graham LS, ed. *Nuclear Medicine: Self-Study Program II: Instrumentation*. Reston, VA: Society of Nuclear Medicine; 1996:5–9.

Henkin RE, Boles MA, Dillehay GL et al., eds., *Nuclear Medicine*. St. Louis, MO: Mosby-Year Book; 1996:Chap. 6.

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