chapter

7

Radiation Detectors

When radiations from a radioactive material pass through matter, they interact with atoms and molecules and transfer energy to them. The transfer of energy has two effects: *ionization* and *excitation.* Ionization occurs when the energy transferred is sufficient to cause an orbital electron to be stripped away from its parent atom or molecule, thus creating an *ion pair* (a negatively charged electron and a positively charged atom or molecule). Excitation occurs when electrons are perturbed from their normal arrangement in an atom or molecule, thus creating an atom or molecule in an *excited state.* Both of these processes are involved in the detection of radiation events; however, ionization is the primary event, and hence the term *ionizing radiation* is used frequently when referring to the emissions from radioactive material. Radiation interactions were discussed in detail in Chapter 6. In this chapter, we describe the basic principles of radiation detectors used in nuclear medicine.

A. GAS-FILLED DETECTORS

1. Basic Principles

Most gas-filled detectors belong to a class of detectors called *ionization detectors.* These detectors respond to radiation by means of ionization-induced electrical currents. The basic principles are illustrated in Figure 7-1. A volume of gas is contained between two electrodes having a voltage difference (and thus an electric field) between them. The negative electrode is called the *cathode,* the positive electrode the *anode.* The electrodes are shown as parallel plates, but they may be a pair of wires, concentric cylinders, and so forth. Under normal circumstances, the gas is an insulator and no electrical current flows between the electrodes. However, radiation passing through the gas causes ionization, both direct ionization from the incident

radiation and secondary ionization from δ rays (see Chapter 6, Section A.1). The electrons produced by ionization are attracted to the positive electrode and the ionized atoms to the negative electrode, causing a momentary flow of a small amount of electrical current.

Gas-filled detectors include *ionization chambers, proportional counters,* and *Geiger-Müller (GM) counters.* The use of these detectors in nuclear medicine is somewhat limited because their stopping power and detection efficiency for x rays and γ rays are quite low; however, they find some use for applications in which detection efficiency is not a major factor and for detection and measurement of nonpenetrating, particle-type radiations. Some of their applications are discussed in Chapters 12 and 23.

2. Ionization Chambers

In most ionization chambers, the gas between the electrodes is air. The chamber may or may not be sealed from the atmosphere. Many different designs have been used for the electrodes in an ionization chamber, but usually they consist of a wire inside of a cylinder or a pair of concentric cylinders.

For maximum efficiency of operation, the voltage between the electrodes must be sufficient to ensure complete collection of ions and electrons produced by radiation within the chamber. If the voltage is too low, some of the ions and electrons simply recombine with one another without contributing to electrical current flow. Figure 7-2 shows the effect of voltage difference between the electrodes on the electrical current recorded by an ionization chamber per ionizing radiation event detected. Recombination occurs at low voltages (*recombination region* of the curve). As the voltage increases there is less recombination and the response (electrical current) increases. When the voltage becomes sufficient to cause complete collection of all of the charges produced, the curve enters a plateau
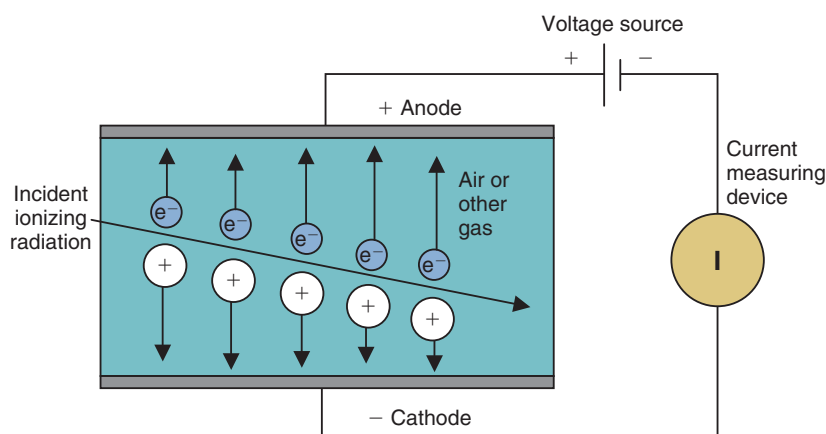
FIGURE 7-1   Basic principles of a gas-filled detector. Electrical charge liberated by ionizing radiation is collected by positive (anode) and negative (cathode) electrodes.
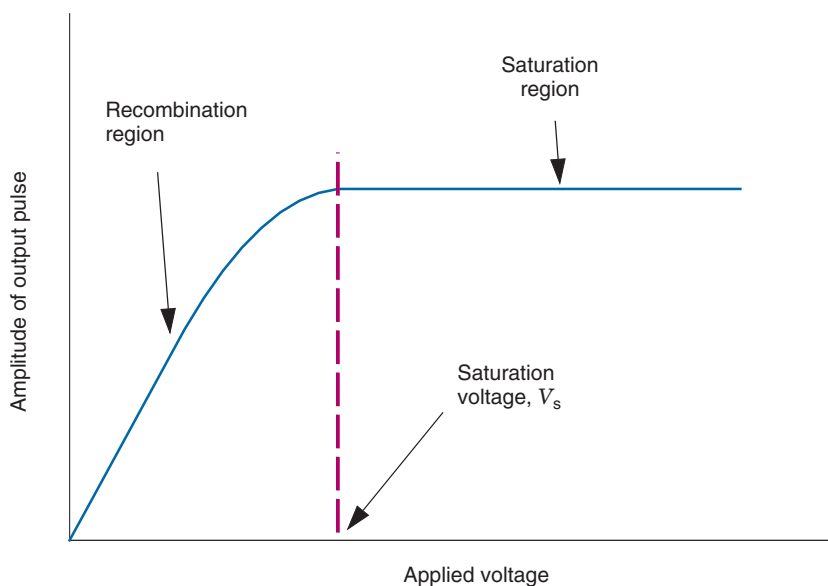

FIGURE 7-2   Voltage response curve (charge collected vs. voltage applied to the electrodes) for a typical ionization chamber. In usual operation, applied voltage exceeds saturation voltage V_s to ensure complete collection of liberated charge.

called the *saturation region*. The voltage at which the saturation region begins is called the *saturation voltage* (V_s). Typically, $V_s \approx$ 50-300 V, depending on the design of the chamber. Ionization chambers are operated at voltages in the saturation region. This ensures a maximum response to radiation and also that the response will be relatively insensitive to instabilities in the voltage applied to the electrodes.

The amount of electrical charge released in an ionization chamber by a single ionizing radiation event is very small. For example, the energy expended in producing a single

ionization event in air is approximately 34 eV.[*] Thus a 1-MeV β particle, for example, causes approximately $(10^6/34) \approx 3 \times 10^4$ ionizations in air and releases a total amount of electrical charge of only approximately 3×10^{-15} coulombs.

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[*]The average energy expended in producing a single ionization event is symbolized by W. This is not the same as the average energy required to ionize an air molecule, but is the average energy expended per ionization by the ionizing particle, including both ionization and excitation effects. This is discussed in detail in Chapter 6, Section A.4. Values of W for some detector materials are listed in Table 7-1.

TABLE 7-1

SOME PROPERTIES OF DETECTOR MATERIALS USED AS IONIZATION DETECTORS

	Si(Li)	Ge(Li) or Ge	CdTe*	Air
ρ(g/cm³)	2.33	5.32	6.06	0.001297
Z	14	32	48 & 52	~7.6
W(eV)†	3.6	2.9	4.43	33.7

CdTe, cadmium telluride; Ge, germanium; Li, lithium; Si, silicon.

*Cadmium zinc telluride (CZT) is CdTe in which some of the Te atoms (typically 20%) are replaced by zinc atoms. CZT has properties similar to CdTe.

†Average energy expended per electron-hole pair created or per ionization.

Because of the small amount of electrical charge or current involved, ionization chambers generally are not used to record or count individual radiation events. Instead, the total amount of current passing through the chamber caused by a beam of radiation is measured. Alternatively, the electrical charge released in the chamber by the radiation beam may be collected and measured.

Small amounts of electrical current are measured using sensitive current-measuring devices called *electrometers*. Two devices consisting of ionization chambers and electrometers in nuclear medicine are *survey meters* and *dose calibrators*. A typical ionization chamber survey meter is shown in Figure 7-3. The survey meter is battery operated and portable. The ionization chamber consists of an outer cylindrical electrode (metal or graphite-coated plastic) with a wire electrode running down its center. There is often a protective cap on the end of the chamber for most measurements; however, it is removed for measurement of nonpenetrating radiations such as α particles, β particles, and low-energy (≲10 keV) photons.

Survey meters are used to monitor radiation levels for radiation protection purposes (see Chapter 23, Section E). Ionization current is displayed on a front-panel meter. Many older units are calibrated to read traditional units of *exposure rate* in roentgens per hour (R/hr) or mR/hr. Newer units are calibrated to read Systeme International units of *air kerma* in grays per hour (Gy/hr), mGy/hr, and so forth, or have a switch-selectable option for choosing between the two systems of units. The definitions and relationships between these units

FIGURE 7-3 A battery-powered radiation survey meter. An ionization chamber is contained in the base of the unit, with the entrance window on the bottom face of the device (not shown). The meter indicates radiation level. The rotary switch is used to select different scale factors. (*Courtesy Ludlum Measurements, Inc., Sweetwater, TX.*)

are discussed in Chapter 23. A typical survey meter can measure exposure rates down to approximately 1 mR/hr or air kerma rates down to approximately 10 µGy/hr.

Dose calibrators are used to assay activity levels in syringes, vials, and so forth containing materials that are to be administered to patients. Unlike other types of ionization chambers discussed in this section, dose calibrators employ sealed and pressurized chambers filled with argon gas. This eliminates the effect of changing barometric pressure on output readings. Dose calibrators typically are calibrated to read directly in units of activity (becquerels or curies), with switches to set the display for different radionuclides. Dose calibrators are discussed in detail in Chapter 12, Section D.1.

A device that records total charge collected over time is the *pocket dosimeter*. The basic principles are illustrated in Figure 7-4. The ionization chamber electrodes are a central charging electrode and the outside case of the dosimeter. They are insulated electrically from one another and form an electrical capacitor. The capacitor is first charged to a reference voltage V by connecting the charging rod to a separate charging unit. If the capacitance between the charging electrode and the case is C, the charge stored on the capacitor is $Q = V \times C$. When the chamber is exposed to
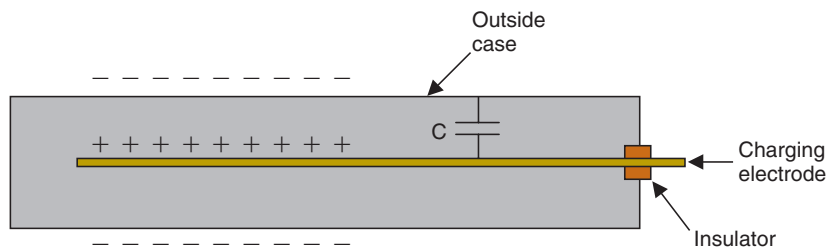
FIGURE 7-4 Schematic representation of a pocket dosimeter.

radiation, electrical charge ΔQ is collected by the electrodes, discharging the capacitor. The voltage change across the capacitor is measured and is related to the amount of electrical charge collected by the ionization chamber electrodes ($\Delta Q = \Delta V \times C$).

Pocket dosimeters are used in nuclear medicine to monitor radiation levels for radiation protection purposes. A typical system is shown in Figure 7-5. The ionization chamber is contained in a small metal or plastic cylinder (~1.5 cm diameter × 10 cm long) that can be clipped to a shirt pocket or collar. Electrodes recessed into one end of the chamber are used to connect the dosimeter to a separate charger unit to charge up the capacitor to the reference voltage. Voltage on the capacitor causes a fine wire within the chamber to be deflected. The position of the wire changes as the voltage on the capacitor changes. The wire is observed through a viewing window at one end of the chamber. Its position is read against a scale that has been calibrated in terms of the total radiation recorded by the chamber, usually in units of air kerma (gray) or exposure (roentgens) (see Chapter 23, Section E). Pocket

dosimeters are suitable for measuring radiation exposures down to approximately 10 mR (air kerma of 0.1 mGy) to an accuracy of approximately 20%.

A basic problem with ionization chambers is that they are quite inefficient as detectors for x rays and γ rays. Only a very small percentage (<1%) of x rays or γ rays passing through the chamber actually interact with and cause ionization of air molecules. Indeed, most of the electrical charge released in an ionization chamber by photon radiations comes from secondary electrons knocked loose from the walls of the chamber by the incident radiations rather than by direct ionization of air molecules. The relatively low detection efficiency of ionization chambers is not a serious limitation in the applications described earlier; however, it precludes their use for most other applications in nuclear medicine, such as imaging.

Two additional problems with ionization chambers should be noted. The first is that for x rays and γ rays, their response changes with photon energy because photon absorption in the gas volume and in the chamber walls (i.e., detection efficiency) and relative penetration

FIGURE 7-5 Pocket dosimeter with charging system. (*Courtesy Ludlum Measurements Inc., Sweetwater, Tx.*)
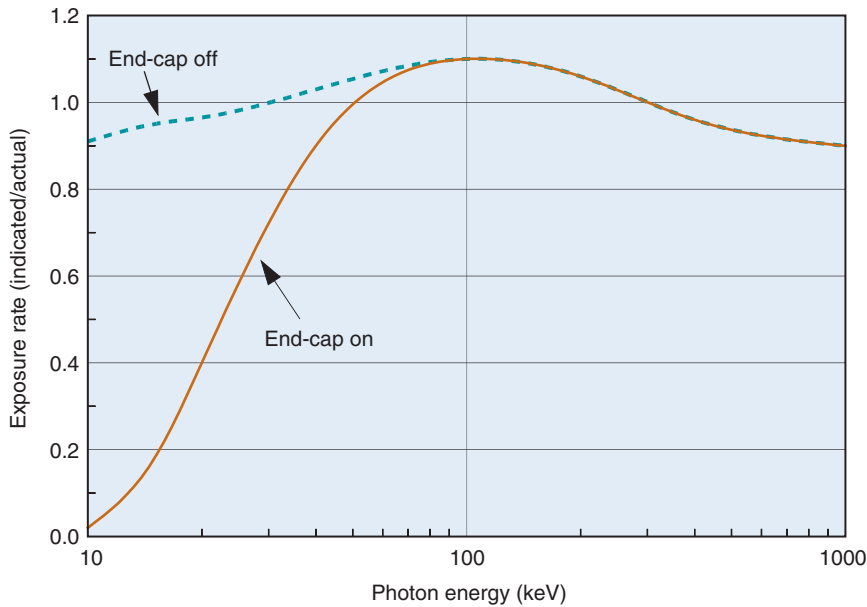
FIGURE 7-6 Energy response curve for a typical ionization chamber survey meter with and without a removable protective end cap.

of photons through the chamber walls are both energy-dependent processes. Figure 7-6 shows a typical energy-response curve for a survey meter. A second problem is that in *unsealed* chambers the density of the air in the chamber, and hence its absorption efficiency, changes with atmospheric pressure ($\rho \propto P$) and temperature ($\rho \propto 1/T$). Most chambers are calibrated to read accurately at sea-level pressure ($P_{ref} = 1.013$ N/m$^2 = 760$ mm Hg) and average room temperature ($T_{ref} = 22°C = 295$K). For other temperatures T and pressures P the chamber reading must be corrected (multiplied) by a temperature-pressure correction factor

$$C_{TP} = (P_{ref} \times T)/(P \times T_{ref}) \qquad (7\text{-}1)$$

Temperature must be expressed on the Kelvin scale in this equation (K = °C + 273). The correction is significant in some cases, for example, at higher elevations ($P \approx 0.85$ N/m$^2 \approx 640$ mm Hg at 1600-meter elevation). Note that temperature-pressure corrections are *not* required with sealed chambers, such as in most dose calibrators. A defective seal on such an instrument obviously could lead to erroneous readings.

3. Proportional Counters

In an ionization chamber, the voltage between the electrodes is sufficient only to collect those charges liberated by direct action of the ionizing radiations. However, if the voltage is increased to a sufficiently high value, the electrons liberated by radiation gain such high velocities and energies when accelerated toward the positive electrode that they cause additional ionization in collisions with other atoms in the gas. These electrons in turn can cause further ionization and so on. This cascade process is called the *Townsend avalanche* or the *gas amplification* of charge. The factor by which ionization is increased is called the *gas amplification factor*. This factor increases rapidly with applied voltage, as shown in Figure 7-7. The gas amplification factor may be as high as 10^6, depending on the chamber design and the applied voltage.

Detectors that operate in the ascending portion of the curve shown in Figure 7-7 are called *proportional counters*. In this region, the ionization caused by an incident radiation event is multiplied (amplified) by the gas amplification factor. The total amount of charge produced is equal to the number of ionizations caused by the primary radiation event (at 34 eV/ionization in air) multiplied by the amplification factor. Thus the total charge produced is proportional to the total amount of energy deposited in the detector by the detected radiation event.

Actually, proportional counters are not simply ionization chambers operated at high voltages but are specially constructed chambers designed to optimize the gas amplification effect, both in terms of the amount of
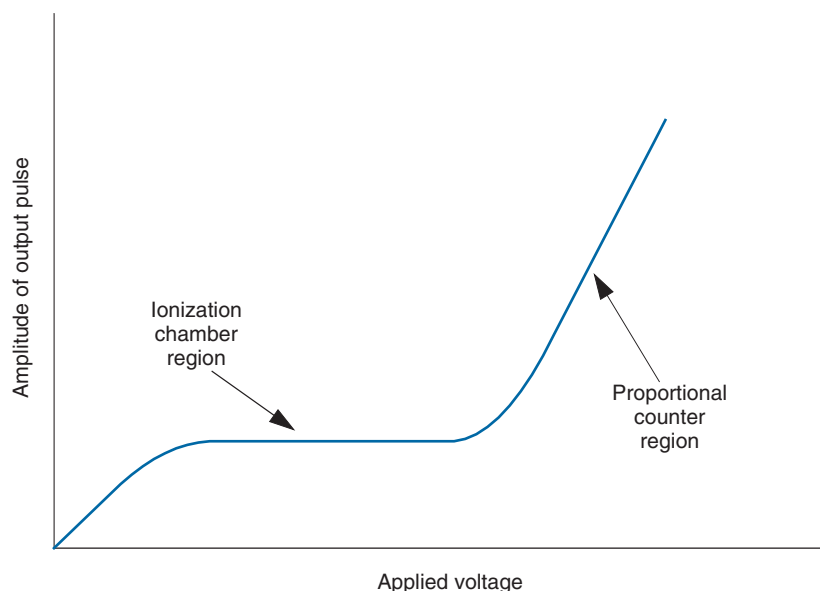
FIGURE 7-7   Voltage response curve for a proportional counter. With increasing applied voltage, the charge collected increases because of the gas amplification effect.

amplification and the uniformity of this amplification within the chamber. In particular, proportional counters are filled with gases that allow easy migration of free electrons, because this is critical for the amplification effect. Common fill gases are the noble gases, with argon and xenon being the most popular.

The major advantage of proportional counters versus ionization chambers is that the size of the electrical signal produced by an individual ionizing radiation event is much larger. They are, in fact, useful for detecting and *counting* individual radiation events. Furthermore, because the size of an individual current pulse is proportional to the amount of energy deposited by the radiation event in the detector, proportional counters can be used for energy-sensitive counting, such as to discriminate between radiation events of different energies on the basis of electrical pulse size (see Chapter 10). They are still inefficient detectors for higher energy x rays and γ rays. Consequently, they find very limited use in nuclear medicine. Proportional counters are used mostly in research applications for measuring nonpenetrating radiations such as α particles and β particles. A practical application is discussed in Chapter 12, Section D.2.

4. Geiger-Müller Counters

A *Geiger-Müller* (GM) *counter* is a gas-filled detector designed for maximum gas amplification effect. The principles of a GM counter

are shown in Figure 7-8. The center wire (anode) is maintained at a high positive voltage relative to the outer cylindrical electrode (cathode). The outer electrode may be a metal cylinder or a metallic film sprayed on the inside of a glass or plastic tube. Some GM counters have a thin radiation *entrance window* at one end of the tube. The cylinder of the tube is sealed and filled with a special gas mixture, typically argon plus a quenching gas (discussed later).

When ionization occurs in a GM counter, electrons are accelerated toward the center wire. Gas amplification occurs in the GM counter as in a proportional counter. In addition to ionizing gas molecules, the accelerating electrons also can cause excitation of gas molecules through collisions. These excited gas molecules quickly (~10^{-9} sec) return to the ground state through the emission of photons at visible or ultraviolet (UV) wavelengths. If a UV photon interacts in the gas, or at the cathode surface by photoelectric absorption (see Chapter 6, Section C.2), this releases another electron, which can trigger a further electron avalanche as it moves toward the anode (see Fig. 7-8). In this way, an *avalanche ionization* is propagated throughout the gas volume and along the entire length of the center wire.

As the avalanche progresses, the electrons, being relatively light, are quickly collected, but the heavy, slow-moving positive ions are not. Eventually, a "hose" of slow-moving
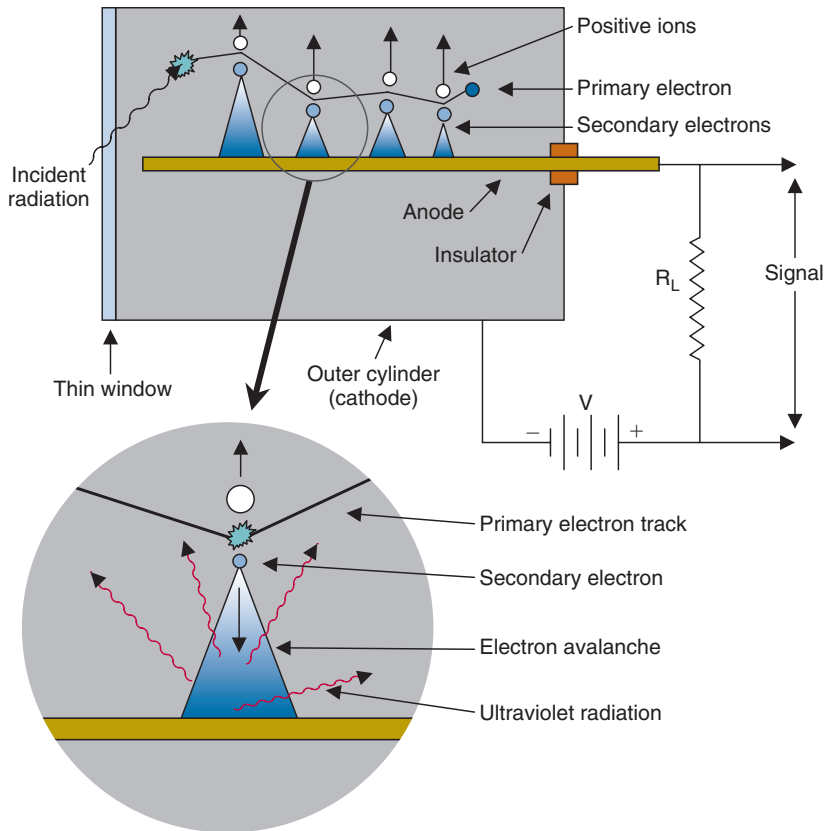
FIGURE 7-8   Operating principles of a Geiger-Müller counter. The incoming radiation produces ion pairs by direct ionization and through secondary fast electrons (δ rays) created in the ionization process. These ion pairs are then multiplied by an avalanche process that in turn triggers further avalanches through the emission of ultraviolet radiation. This process is terminated when a sufficient number of positive ions collect around the anode, effectively reducing the electric field experienced by the electrons owing to charge buildup at the anode.

positive charges is formed around the center wire. The avalanche then terminates because the positive ions reduce the effective electric field around the anode wire, eventually dropping it below the level required for gas multiplication.

The avalanche ionization in a GM tube releases a large and essentially constant quantity of electrical charge, regardless of voltage applied to the tube (Fig. 7-9) or the energy of the ionizing radiation event. The gas amplification factor may be as high as 10^{10}. The large electrical signal is easily detected with electronic circuits. Thus a GM counter, like a proportional counter, can be used to detect and count individual ionizing radiation events. However, because the size of the electrical signal output is constant, regardless of the energy of the radiation detected, a GM counter cannot be used to distinguish between radiation events of different energies.

Once the avalanche has terminated in a GM counter, an additional problem arises. The positive ion cloud moves toward the outer electrode. When the ion cloud is very close to the outer electrode, electrons are pulled out from it to neutralize the positive ions. Some of these electrons enter higher-energy orbits of the positive ions; when they eventually drop into the lower-energy orbits, UV radiation is emitted. This can cause the release of more electrons from the outer wall and set off another avalanche. Thus if no precautions are taken, a single ionizing radiation event can cause the GM counter to go into a pulsating series of discharges.

This problem is prevented by the introduction of a *quenching gas* into the GM counter gas mixture. Such GM counters are called *self-quenched*. Effective quenching gases have three properties: First, they tend to give up electrons easily. When the positive ion cloud is formed, molecules of the quenching gas
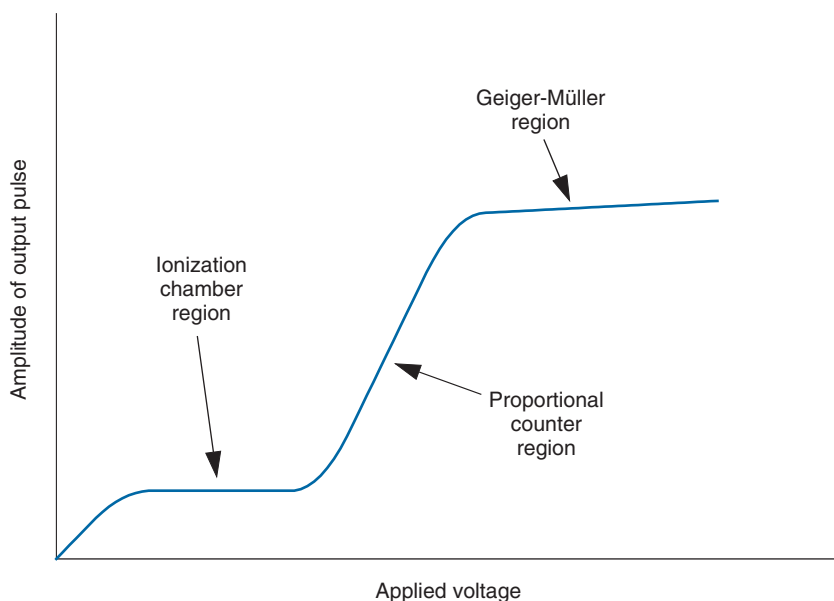
FIGURE 7-9   Voltage response curve (pulse amplitude vs. applied voltage) for a Geiger-Müller counter.

neutralize other ions by donating electrons to them. The ion cloud is thus converted into ionized molecules of quenching gas. Second, when the quenching gas molecules are neutralized by electrons entering higher energy orbits, they deenergize themselves by dissociating into molecular fragments rather than by emitting UV photons. Third, the quenching gas molecules are strong absorbers of UV radiation. Thus the few UV photons that are released during neutralization of the positive ion cloud are quickly absorbed before they can set off another avalanche.

Commonly used quenching gases include heavy organic vapors (e.g., alcohol) and halogen gases (e.g., Cl_2). The organic vapors are more effective quenching agents but have the disadvantage that their molecular fragments do not recombine after dissociation. Thus an organic quenching gas eventually is used up, typically after approximately 10^{10} radiations have been detected. Halogen gas molecules recombine after dissociation and thus have an essentially unlimited lifetime in a GM counter.

A certain minimum voltage is required between the electrodes of a GM counter to sustain an avalanche ionization and to raise the amplitude of the pulses to the threshold of the counting system. This voltage can be determined by exposing the GM counter to a constant source of radiation and observing the counting rate as a function of voltage applied to the counter electrodes. Figure 7-10 shows the results of such an experiment. This curve is called the *counting curve* or *plateau curve* of the GM counter. As the high voltage is increased, the counting rate increases rapidly as more and more of the output pulses exceed the counter threshold. When the voltage is sufficient that essentially all pulses are above the threshold and are counted, a *plateau region* is reached. The point at which the plateau begins is called the knee of the curve. Further increases in voltage may still increase the amplitude of the output pulses; however, the counting rate remains constant as the radiation source is constant.*

When the voltage is increased to a very high value, the counting rate again begins to increase. This happens when the voltage is so high that spontaneous ionization begins to occur in the chamber. The curve then enters the *spontaneous discharge region.* GM counters should not be operated in the spontaneous discharge region because no useful information can be obtained there. Furthermore, if the counter contains an organic quenching gas, it is rapidly used up by the spontaneous discharges, thus shortening the

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*Actually, for most GM counters the counting rate increases by 1% to 2% per 100 volts in the plateau region. This is of no practical consequence in nuclear medicine.
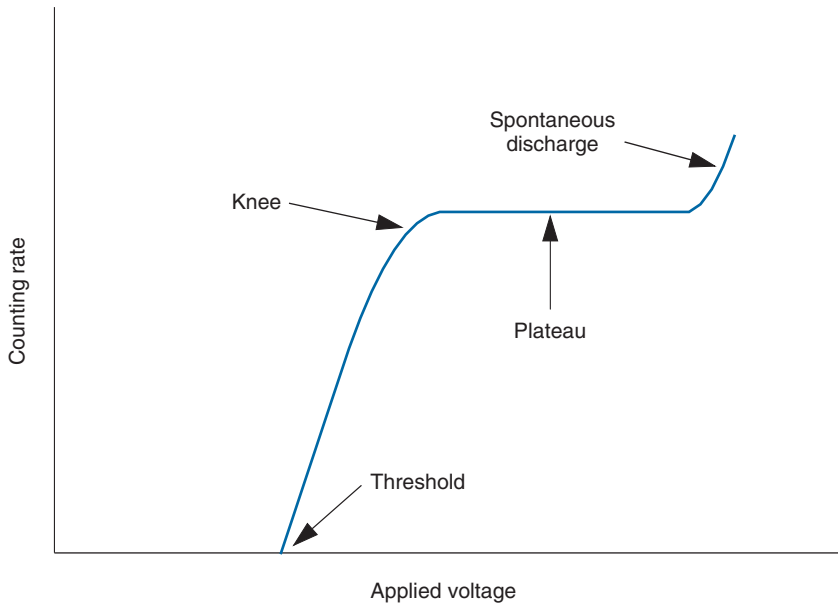
FIGURE 7-10 Counting curve (counting rate from a fixed radiation source vs. applied voltage) for a Geiger-Müller counter. As voltage increases, pulse amplitude increases above threshold of counting system electronics. When all events produce a signal above the threshold, a plateau is reached. At very high voltages, spontaneous discharge events occur within the chamber. These are not caused by radiation events but by electrical breakdown in the gas.

life of the counter. The proper operating voltage is the plateau region, about one third the distance from the knee to the spontaneous discharge region.

GM counters are simple, rugged, and relatively inexpensive radiation detectors. Much of the early (pre-1950s) work in nuclear medicine was done with GM counters; however, they have since been replaced for most applications by other types of detectors. The major disadvantages of GM counters are low detection efficiency (<1%) for γ rays and x rays and an inability to distinguish between radiation events of different energies on the basis of pulse size for energy-selective counting (because all pulses from a GM counter are the same size).

GM counters are used mostly in survey meters for radiation protection purposes. An example is shown in Figure 7-11. The detector in this survey meter is of the *pancake* type. The entrance window at the end of the counter tube is a thin layer of mica (0.01-mm thick) that is sufficiently thin to permit passage of particles and low-energy photons into the counter. The rather fragile window is protected by a wire screen. GM counters designed for counting only relatively penetrating radiations, such as γ rays and high-energy β particles, have thicker, sturdier windows,

FIGURE 7-11 Radiation survey meter with an external Geiger-Müller (GM) "pancake" counter radiation detector attachment. In addition to the external detector shown, some units have built-in GM counters. (*Courtesy Ludlum Measurements, Inc. Sweetwater, TX.*)

for example, 0.1-mm-thick aluminum or stainless steel. Many GM counters are provided with removable covers on the entrance window that can be used to distinguish between penetrating and nonpenetrating

radiations by observing the difference between counting rates with and without the cover in place. GM survey meters are more sensitive than ionization chamber survey meters, typically by a factor of approximately 10.

B. SEMICONDUCTOR DETECTORS

Semiconductor detectors are essentially solid-state analogs of gas-filled ionization chambers. Because the solid detector materials used in semiconductor detectors are 2000 to 5000 times more dense than gases (see Table 7-1), they have much better stopping power and are much more efficient detectors for x rays and γ rays.

Semiconductor detectors normally are poor electrical conductors; however, when they are ionized by an ionizing radiation event, the electrical charge produced can be collected by an external applied voltage, as it is with gas-filled detectors. This principle could not be applied using a conducting material for the detector (e.g., a block of metal) because such a material would conduct a large amount of current even without ionizing events. Insulators (e.g., glass) are not suitable detector materials either, because they do not conduct even in the presence of ionizing radiation. Hence only semiconductor materials can function as "solid ionization chambers."

The most commonly used semiconductor detector materials are silicon (Si) and germanium (Ge). More recently, cadmium telluride (CdTe) or cadmium zinc telluride (CZT) have been used as the detector material in small nuclear medicine counting and imaging devices. Characteristics of these semiconductor materials are listed in Table 7-1. One ionization is produced per 3 to 5 eV of radiation energy absorbed. By comparison, this value for gases (air) is approximately 34 eV per ionization. Thus a semiconductor detector not only is a more efficient absorber of radiation but produces an electrical signal that is approximately 10 times larger (per unit of radiation energy absorbed) than a gas-filled detector. The signal is large enough to permit detection and counting of individual radiation events. Furthermore, the size of the electrical signal is proportional to the amount of radiation energy absorbed. Therefore semiconductor detectors can be used for energy-selective radiation counting. For reasons discussed in Chapter 10, Section C.1, they are in fact the preferred type of detector for this application.

In spite of their apparent advantages, semiconductor detectors have a number of problems that have limited their use in nuclear medicine. The first is that both Si and Ge (especially Ge) conduct a significant amount of thermally induced electrical current at room temperature. This creates a background "noise current" that interferes with detection of radiation-induced currents. Therefore Si detectors (usually) and Ge detectors (always) must be operated at temperatures well below room temperature.

A second problem is the presence of impurities even in relatively pure crystals of Si and Ge. Impurities (atoms of other elements) enter into and disturb the regular arrangement of Si and Ge atoms in the crystal matrix. These disturbances create "electron traps" and capture electrons released in ionization events. This results in a substantial reduction in the amount of electrical signal available and limits the thickness of a practical detector to approximately 1 cm. Because of the relatively low atomic numbers of Si and Ge, this restricts their efficiency for detection of γ rays.

Two approaches have been used to solve the impurity problem. One is to prepare very pure samples of the detector material. This has been accomplished only with Ge [*high-purity germanium* (HPGe)] and is, unfortunately, quite expensive. Also, the size of pure crystals is limited to approximately 5 cm in diameter by 1 cm thick. Detectors made of HPGe are sometimes called *intrinsic Ge* detectors. A second approach is to deliberately introduce into the crystal matrix "compensating" impurities that donate electrons to fill the electron traps created by other impurities. Lithium (Li) is commonly used in Si and Ge detectors for this purpose. Detectors made of "lithium-doped" materials are called *lithium-drifted* detectors, or Si(Li) or Ge(Li) detectors. Unfortunately, the process of preparing Si(Li) or Ge(Li) crystals is time consuming and expensive. Crystal sizes are limited to a few centimeters in diameter by approximately 1 cm thick for Si(Li) and approximately 5 cm diameter by 5 cm thick for Ge(Li).

An additional problem is that Li ions tend to "condense" within the crystal matrix at room temperature, especially in Ge. Therefore Si(Li) and Ge(Li) not only must be operated at low temperatures (to minimize thermally induced background currents) but Ge(Li) detectors must and Si(Li) detectors should also be *stored* at low temperatures.

Liquid nitrogen (T = 77K or –196°C) is used for detector cooling. Ge(Li) detectors can be ruined by only an hour or so at room temperature. Si(Li) detectors can tolerate elevated temperatures, but they provide optimum performance if they also are stored at liquid nitrogen temperatures.

Because of the difficulties inherent in Li-drifted detectors, HPGe has become the detector material of choice for γ-ray spectroscopy applications (see Chapter 10), and most manufacturers have now stopped producing Ge(Li) detectors. Si(Li) finds applications in low-energy x-ray and β-particle spectroscopy, in which its low atomic number is not a disadvantage.

Figure 7-12 shows schematically a typical semiconductor detector assembly. The detector consists of a thin, circular disc of the detector material [Si(Li), Ge(Li), or HPGe] with electrodes attached to its opposite faces for charge collection. One electrode is a thin metal foil fastened to the front surface ("entrance window"), whereas the other is a wire or set of wires embedded in the opposite surface of the crystal. Other detector shapes and electrode configurations also are used.

Figure 7-12 also shows in cross-section an apparatus used to cool the crystal with liquid nitrogen. A "coldfinger" extends from the liquid nitrogen container (a Dewar flask) to cool the detector. Some of the preamplifier electronic circuitry also is cooled to reduce electronic noise levels. Liquid nitrogen

evaporates and the container needs periodic refilling—typically every 2 to 3 days, depending on container size and insulation characteristics.

CdTe and CZT (which has properties very similar to CdTe) are more recently developed semiconductor materials that overcome two of the major disadvantages of Si and Ge: (1) they can be operated at room temperature without excessive electronic noise, and (2) their high atomic number means that even relatively thin detectors can have good stopping efficiency for detecting γ rays. Although CdTe and CZT are now being used in some nuclear medicine counting and imaging devices, their use has generally been restricted to small detectors, or detectors comprising multiple small elements, because of the difficulty and expense of growing large pieces of CdTe or CZT with the required purity. Additional discussion of their properties for pulse-height spectrometry is presented in Chapter 10, Section C.1.

C. SCINTILLATION DETECTORS

1. Basic Principles

As indicated earlier in this chapter, radiation from radioactive materials interacts with matter by causing ionization or excitation of atoms and molecules. When the ionized or excited products undergo recombination or deexcitation, energy is released. Most of the energy is dissipated as thermal energy, such as molecular vibrations in gases or liquids or lattice vibrations in a crystal; however, in some materials a portion of the energy is released as visible light.* These materials are called *scintillators,* and radiation detectors made from them are called *scintillation detectors.*

The scintillator materials used for detectors in nuclear medicine are of two general types: inorganic substances in the form of solid crystals and organic substances dissolved in liquid solution. The scintillation mechanisms are different for these two types and are described separately in later sections.

A characteristic common to all scintillators is that the amount of light produced following
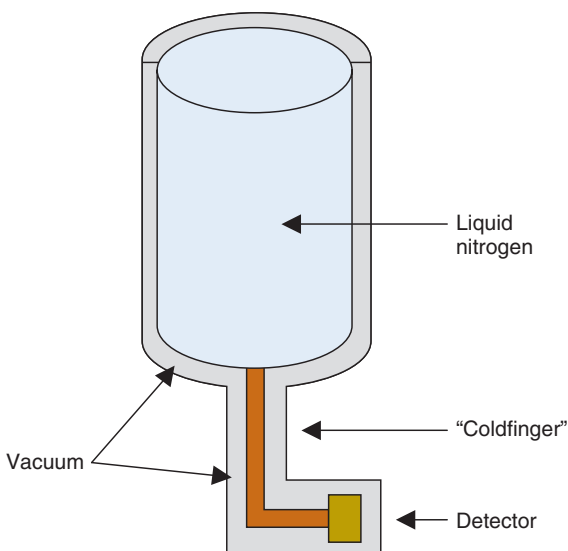

FIGURE 7-12 Schematic representation of a typical semiconductor detector assembly. "Coldfinger" is a thermal conductor for cooling the detector element.

*For simplicity, the term *visible light* is used to describe scintillation emission. In fact, the emissions from many scintillators extend into the UV portion of the spectrum as well.

the interaction of a single γ ray, β particle, or other ionizing radiation, is proportional to the energy deposited by the incident radiation in the scintillator. The amount of light produced also is very small, typically a few hundred to a few thousand photons for a single γ-ray interaction within the energy range of interest for nuclear medicine imaging (70-511 keV). In the early days of nuclear physics, it was common to study the characteristics of particles by observing and counting, in a darkened room, the scintillations produced by these particles on a zinc sulfide scintillation screen. The obvious limitations on counting speed and accuracy with this system have been eliminated in modern application with the introduction of ultrasensitive electronic light detectors called *photomultiplier* (*PM*) *tubes*.

2. Photomultiplier Tubes

PM tubes (also called *phototubes* and sometimes abbreviated *PMT*) are electronic tubes that produce a pulse of electrical current when stimulated by very weak light signals, such as the scintillation produced by a γ ray or β particle in a scintillation detector. Their basic principles are illustrated in Figure 7-13.

The inside front surface of the glass *entrance window* of the PM tube is coated with a photoemissive substance. A *photoemissive* substance is one that ejects electrons when struck by photons of visible light. Cesium antimony (CsSb) and other bialkali

compounds are commonly used for this material. The photoemissive surface is called the *photocathode,* and electrons ejected from it are called *photoelectrons*. The conversion efficiency for visible light to electrons, also known as the *quantum efficiency,* is typically 1 to 3 photoelectrons per 10 visible light photons striking the photocathode. The dependence of quantum efficiency on the wavelength of the light is shown for a conventional bialkali photocathode in Figure 7-14.

A short distance from the photocathode is a metal plate called a *dynode*. The dynode is maintained at a positive voltage (typically 200-400 V) relative to the photocathode and attracts the photoelectrons ejected from it. A *focusing grid* directs the photoelectrons toward the dynode. The dynode is coated with a material having relatively high secondary emission characteristics. CsSb also can be used for this material. A high-speed photoelectron striking the dynode surface ejects several *secondary electrons* from it. The electron multiplication factor depends on the energy of the photoelectron, which in turn is determined by the voltage difference between the dynode and the photocathode.

Secondary electrons ejected from the first dynode are attracted to a second dynode, which is maintained at a 50-150 V higher potential than the first dynode, and the electron multiplication process is repeated. This occurs through many additional dynode stages (typically 9 to 12 in all), until finally a
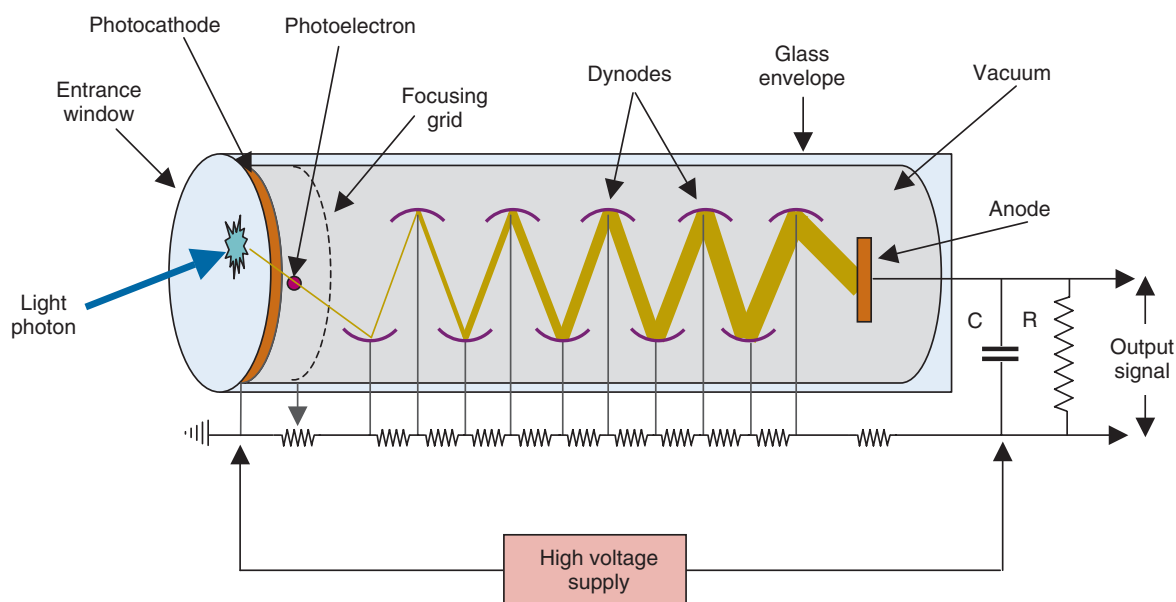

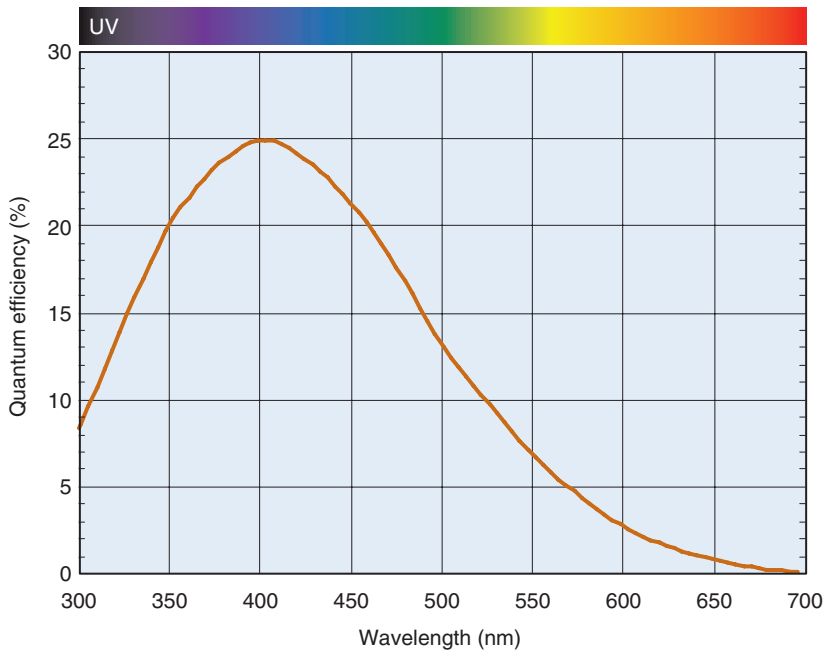**FIGURE 7-13**   Basic principles of a photomultiplier tube.

FIGURE 7-14   Quantum efficiency as a function of wavelength for a typical bialkali photocathode. The peak occurs at approximately 400 nm, which is well-matched to the emission wavelength of many scintillators. UV, ultraviolet light.

shower of electrons is collected at the *anode.* Typical electron multiplication factors are ×3 to ×6 per dynode. The total electron multiplication factor is very large—for example, 6^{10} ($\sim6 \times 10^7$) for a 10-stage tube with an average multiplication factor of 6 at each dynode. Thus a relatively large pulse of current is produced when the tube is stimulated by even a relatively weak light signal. Note that the amount of current produced is proportional to the intensity of the light signal incident on the photocathode and thus also to the amount of energy deposited by the radiation event in the crystal.

PM tubes require a high-voltage supply. For example, as shown in Figure 7-13, if the tube has 10 dynodes, with the first at +300 V relative to the photocathode and the remaining 9 dynodes and the anode at additional +100 V increments, a voltage of +1300 V is needed. Furthermore, the voltage supply must be very stable because the electron multiplication factor is very sensitive to dynode voltage changes. Typically a 1% increase in high voltage applied to the tube increases the amount of current collected at the anode by approximately 10%. This is of considerable importance in applications where pulse size is being measured, such as in pulse-height spectrometry to determine γ-ray energies (see Chapter 10).

PM tubes are sealed in glass and evacuated. Electrical connections to the dynodes, the photocathode, and the anode are made through pins in the tube. The focusing of the electron beam from one dynode to the next can be affected by external magnetic fields. Therefore PM tubes often are wrapped in metal foil for magnetic shielding. "Mu-metal," an alloy composed of iron, nickel, and small amounts of copper and chromium, is commonly used for this purpose. PM tubes come in various shapes (round, square, and hexagonal) and sizes (Fig. 7-15). Most of those used in nuclear medicine have photocathodes in the range of 1- to 7.5-cm diameter. There are also position-sensitive and multichannel PM tubes available that have the ability to determine the location of incident light on the photocathode.

3. Photodiodes

In some applications, the PM tube may be replaced by a light-sensitive semiconductor detector, such as a Si photodiode. Note that in this case, the semiconductor is not being used to detect the γ rays directly but to detect the visible light emitted from a scintillator material in which a γ ray has interacted. The photons from the scintillator have sufficient energy to cause ionization within Si, and the total charge produced is proportional to the

FIGURE 7-15 Assortment of photomultiplier tubes illustrating their wide variety of shapes and sizes. (*Courtesy Hamamatsu Corp., Bridgewater, NJ.*)

number of scintillation light photons incident on the photodiode. These photodiode detectors have the advantage that they can be made very small in area and that they are typically only a few millimeters thick, including the packaging. They also have significantly higher quantum efficiency than PM tubes, with values ranging typically between 60% and 80%. However, conventional Si photodiodes have unity gain (compared with 10^6 to 10^7 for a PM tube), requiring very low-noise electronics for readout.

A related device, the Si avalanche photodiode (APD) uses a very high internal electric field such that each electron produced within the device gains enough energy between collisions to create further ionization. This is analogous to the proportional region for gas-filled detectors that was discussed previously in Section A.3. APD detectors can reach gains of 10^2 to 10^3 but still require low-noise electronics for successful operation. The gain of these devices also is a very strong function of bias voltage and temperature, and these parameters therefore must be very carefully controlled for stable operation. These types of solid-state light detectors are used presently only in specialized nuclear medicine systems, such as small animal scanners (see Chapter 17, Section A.4, and Chapter 18, Section B.5) and dual-modality positron emission tomography–magnetic resonance imaging (PET-MRI) systems (see Chapter 19, Section F).

APDs also can be operated at higher bias voltages in *geiger mode*. This is analogous to the gas-filled GM counter (Section A.4), and the output signal of the APD becomes very large and independent of the energy of the incident radiation. The gain of such devices can be as high as 10^7. Light-sensitive detectors consisting of a large number of tiny (20-50 μm) *geiger-mode APDs* that are incorporated into a single device are being developed for use in scintillation detectors.

4. Inorganic Scintillators

Inorganic scintillators are crystalline solids that scintillate because of characteristics of their crystal structure. Individual atoms and molecules of these substances do not scintillate. They are scintillators only in crystalline form. Table 7-2 summarizes the properties of a number of inorganic scintillators of interest for nuclear medicine applications.

Some inorganic crystals are scintillators in their pure state; for example, pure NaI crystals are scintillators at liquid nitrogen temperatures. Most are "impurity activated," however. These are crystals containing small amounts of "impurity" atoms of other elements. Impurity atoms in the crystal matrix cause disturbances in its normal structure. Because they are responsible for the scintillation effect, the impurity atoms in the crystal matrix are sometimes called *activator centers.* Some impurity-activated scintillators that have been used in radiation detectors include sodium iodide [NaI(Tl)] and cesium iodide [CsI(Tl)]. In each case, the element in parentheses is the impurity that is added to create activator centers in the crystal.

The most commonly used scintillator for detectors in nuclear medicine is NaI(Tl). Pure

TABLE 7-2
PROPERTIES OF SOME SCINTILLATOR MATERIALS USED IN NUCLEAR MEDICINE

Property	NaI(Tl)	BGO	LSO(Ce)	GSO(Ce)	CsI(Tl)	LuAP(Ce)	LaBr$_3$(Ce)	Plastic*
Density (g/cm^3)	3.67	7.13	7.40	6.71	4.51	8.34	5.3	1.03
Effective atomic number	50	73	66	59	54	65	46	12
Decay time (nsec)	230	300	40	60	1000	18	35	2
Photon yield (per keV)	38	8	20-30	12-15	52	12	61	10
Index of refraction	1.85	2.15	1.82	1.85	1.80	1.97	1.9	1.58
Hygroscopic	Yes	No	No	No	Slightly	No	Yes	No
Peak emission (nm)	415	480	420	430	540	365	358	Various

*Typical values—there are many different plastic scintillators available.
BGO, Bi$_3$Ge$_4$O$_{12}$; GSO(Ce), Gd$_2$SiO$_5$(Ce); LSO(Ce), Lu$_2$SiO$_5$(Ce); LuAP(Ce), LuAlO$_5$(Ce)

NaI crystals are scintillators only at liquid nitrogen temperatures. They become efficient scintillators at room temperatures with the addition of small amounts of thallium. Single crystals of NaI(Tl) for radiation detectors are "grown" from molten sodium iodide to which a small amount of thallium (0.1-0.4 mole percent) has been added. Crystals of relatively large size are grown in ovens under carefully controlled temperature conditions. For example, crystals for gamma cameras (see Chapter 13) use NaI(Tl) crystals that are typically 30-50 cm in diameter by 1-cm thick.

Figure 7-16 shows the construction of a typical scintillation detector consisting of a NaI(Tl) crystal and PM tube assembly. The crystal is sealed in an aluminum or stainless-steel jacket with a transparent glass or plastic optical window at one end to permit the exit of scintillation light from the crystal to the PM tube. A transparent optical "coupling grease" is used between the crystal and the PM tube to minimize internal reflections at this interface. The crystal and PM tube are hermetically sealed in a light-tight jacket to keep out moisture and extraneous light and for mechanical protection. The inside surface of the radiation entrance window and sides of the crystal are coated with a highly reflective diffuse material to maximize the light collected by the PM tube photocathode. With efficient optical coupling, good reflective surfaces, and a crystal free of cracks or other opacifying defects, approximately 30% of the light emitted by the crystal actually reaches the cathode of the PM tube. Some NaI(Tl) detectors have very
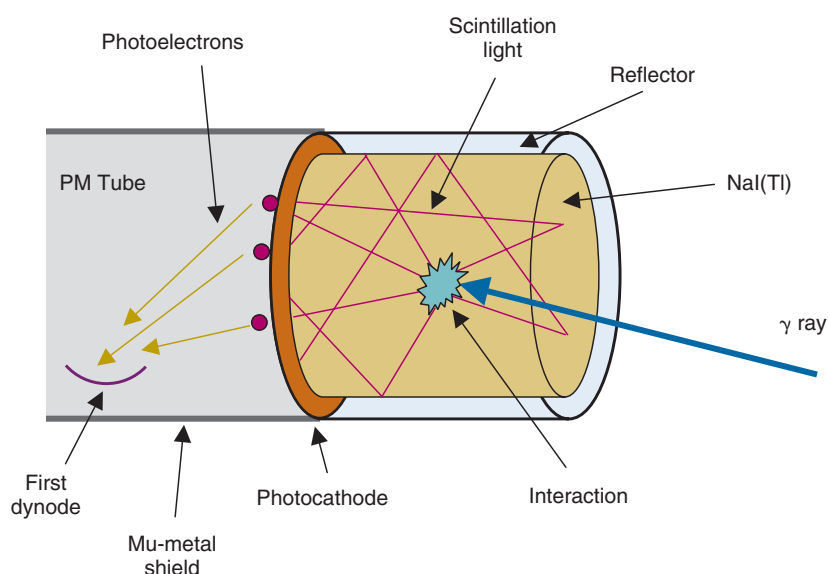

FIGURE 7-16 Arrangement of NaI(Tl) crystal and photomultiplier (PM) tube in a typical scintillation detector assembly.

thin aluminum or beryllium foil "entrance windows" to permit detection of radiations having relatively low penetrating power, such as low-energy x rays and γ rays ($E \lesssim 10$ keV) and β particles; however, most NaI(Tl) detectors have thicker entrance windows of aluminum or stainless steel and are best suited for detecting higher-energy γ rays ($E \gtrsim 50$ keV). Figure 7-17 shows some typical integral NaI(Tl) crystal and PM tube assemblies.

Some reasons for the usefulness of NaI(Tl) scintillation detectors include the following:

1. It is relatively dense ($\rho = 3.67$ g/cm^3) and contains an element of relatively high atomic number (iodine, $Z = 53$). Therefore it is a good absorber and a very efficient detector of penetrating radiations, such as x rays and γ rays in the 50- to 250-keV energy range. The predominant mode of interaction in this energy range is by photoelectric absorption.
2. It is a relatively efficient scintillator, yielding one visible light photon per approximately 30 eV of radiation energy absorbed.
3. It is transparent to its own scintillation emissions. Therefore there is little loss of scintillation light caused by self-absorption, even in NaI(Tl) crystals of relatively large size.
4. It can be grown relatively inexpensively in large plates, which is advantageous for imaging detectors.

FIGURE 7-17 NaI(Tl) crystal and photomultiplier tube assemblies. (*Courtesy Crystals and Detectors Division, Saint-Gobain Ceramics and Plastics, Inc., Newbury, OH.*)

5. The scintillation light is well-matched in wavelength to the peak response of the PM tube photocathode (see Fig. 7-14). The emission spectrum of light from NaI(Tl) is shown in Figure 7-18.

Some disadvantages of NaI(Tl) detectors are the following:

1. The NaI(Tl) crystal is quite fragile and easily fractured by mechanical or thermal stresses (e.g., rapid temperature changes). Fractures in the crystal do not necessarily destroy its usefulness as a detector, but they create opacifications within the crystal that reduce the amount of scintillation light reaching the photocathode.
2. Sodium iodide is hygroscopic. Exposure to moisture or a humid atmosphere causes a yellowish surface discoloration that again impairs light transmission to the PM tube. Thus hermetic sealing is required.
3. At higher γ-ray energies ($\gtrsim 250$ keV), the predominant mechanism of interaction is by Compton interaction, and larger volumes of NaI(Tl) are required for adequate detection efficiency.

Other types of detectors have advantages over NaI(Tl) detectors in certain areas. For example, gas-filled detectors are cheaper (but have much lower detector efficiency), and semiconductor detectors have better energy resolution (but are expensive to use in large-area imaging cameras). However, the overall advantages of NaI(Tl) have made it the detector of choice for almost all routine applications in nuclear medicine involving γ-ray detection in the 50-250-keV energy range.

At higher energies, particularly for detection of the 511-keV emissions from positron emitters, denser scintillators generally are preferred. Bismuth germanate ($Bi_4Ge_3O_{12}$; BGO) is a commonly used scintillator in PET imaging, because of its excellent detection efficiency at 511 keV. Lutetium oxyorthosilicate [$Lu_2SiO_5(Ce)$; LSO] is slightly less efficient at 511 keV than BGO, but is brighter and faster and may sometimes offer advantages over BGO when the counting rate on the detector is high, when fast timing information is needed, or when small scintillator elements are to be decoded in an imaging system (see Chapter 18, Section B). However, LSO is rather expensive to grow, because of its high melting point and its raw material costs. A related material, lutetium yttrium orthosilicate (LYSO—LSO in which a small fraction of the lutetium atoms are replaced
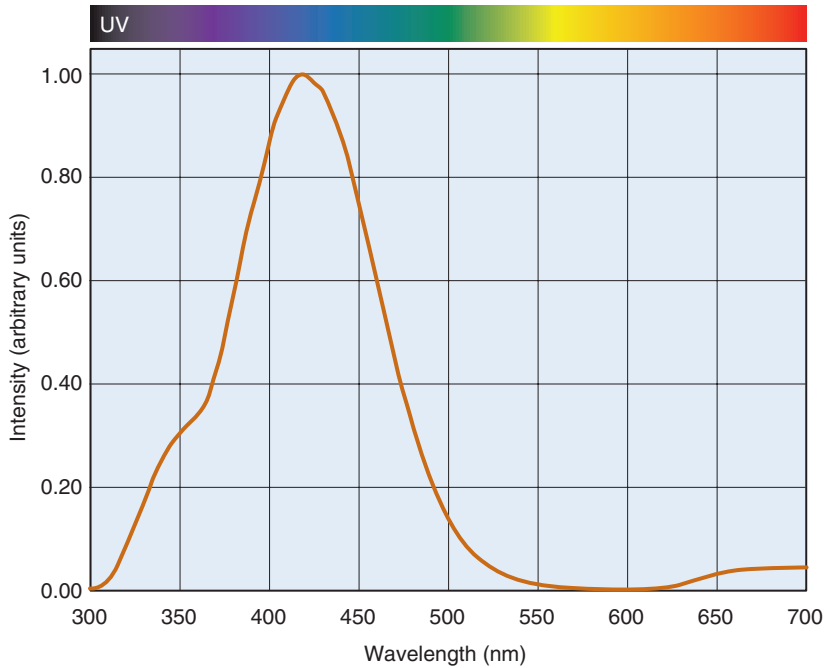
FIGURE 7-18   The emission spectrum of NaI(Tl) scintillator at room temperature. UV, ultraviolet light. (*Data courtesy Kanai Shah and Jarek Glodo, Radiation Monitoring Devices Inc., Watertown, MA.*)

with yttrium) has scintillation properties very similar to LSO. BaF_2 and CsF also have historically been used in positron cameras because of their very fast decay time (which is important for timing of γ-ray interactions in coincidence detection) (see Chapter 18, Section A.2). However, their low detection efficiency compared with BGO and LSO has prevented any widespread application.

New scintillator materials continue to be discovered and developed. Among the more promising recent candidates for nuclear medicine applications are LuAP [$LuAlO_3(Ce)$], lanthanum bromide [$LaBr_3(Ce)$], and lanthanum chloride [$LaCl_3(Ce)$].

5. Considerations in Choosing an Inorganic Scintillator

The ability of a scintillator to stop high-energy γ rays ($\gtrsim 100$ keV) is of particular importance in nuclear medicine, especially for imaging applications. As discussed in Chapter 6, Section D.1, the photoelectric component of the mass-attenuation coefficient increases strongly with the atomic number, Z, of a material, whereas the Compton component is essentially independent of atomic number. As illustrated in Fig. 6-18, although the Compton component is significant over much of the nuclear-medicine energy range, the

photoelectric component dominates in high-Z materials over much of that range. As well, high-Z materials tend to be denser than low-Z materials, thus increasing their linear attenuation coefficients and enhancing their stopping power per unit of detector thickness. Thus, high-Z detector materials generally are more efficient than low-Z materials for detecting high-energy γ rays.

Most detector materials consist of a mixture of elements. The effective atomic number, Z_{eff}, is a useful and convenient parameter for comparing the ability of such mixtures to stop high-energy γ rays. It is given by

$$Z_{eff} = \sqrt[x]{w_1 Z_1^x + w_2 Z_2^x + \cdots w_n Z_n^x} \qquad (7-2)$$

where w_i is a weighting factor proportional to the fractional number of electrons per gram for element i and can be calculated as

$$w_i = \frac{m_i Z_i}{\sum_{i=1}^{n} m_i Z_i} \qquad (7-3)$$

where m_i represents the number of atoms of element i present in the compound or mixture. The power x depends on the energy of the γ rays. For γ rays in the 100- to 600-keV range, x is typically taken to be between 3 and 3.5.[1]

EXAMPLE 7-1

Calculate the effective atomic number of BGO ($Bi_4Ge_3O_{12}$). Compare this with the value for NaI(Tl). (Ignore the contribution of the trace amounts of thallium.)

Answer

There are three elements contributing to BGO: Bi (Z = 83), Ge (Z = 32), and O (Z = 8). The denominator for the weighting factors is $(83 \times 4) + (32 \times 3) + (8 \times 12) = 524$. The weighting factor for Bi is $(83 \times 4) / 524 = 0.634$, for Ge is $(32 \times 3) / 524 = 0.183$, and for O is $(8 \times 12) / 524 = 0.183$. These three weighting factors add to 1, as they should. Then Z_{eff} can be calculated as

$$Z_{eff} = (0.634 \times 83^{3.5} + 0.183 \times 32^{3.5} + 0.183 \times 8^{3.5})^{1/3.5}$$
$$= 73.1$$

NaI(Tl) consists of Na (Z=11) and I (Z=53) atoms that are present in equal quantities. The weighting factor for Na is $11/(11 + 53) = 0.172$ and for I is $53/(11 + 53) = 0.828$. Therefore Z_{eff} for NaI(Tl) is

$$Z_{eff} = (0.172 \times 11^{3.5} + 0.828 \times 53^{3.5})^{1/3.5}$$
$$= 50.2$$

In addition to choosing a scintillator that has sufficient stopping power for the efficient detection of γ rays with a particular energy, the other properties listed in Table 7-2 also influence the choice of scintillator materials for a specific application. The *decay time* of the scintillator is important in two respects. Firstly, it determines the precision with which the time of γ-ray interaction in the scintillator can be determined. Faster light production within the scintillator (faster decay time) results in better timing precision. This is important in nuclear medicine applications in which timing is important, most notably in PET in which coincident 511 keV annihilation photons are detected (see Chapter 18, Section A). Secondly, the decay time of the scintillator is a limiting factor in how many γ-ray interactions a detector can process per unit time. To unambiguously detect two interactions, they should be separated by roughly 2-3 times the decay time; otherwise events "pile up" on top of each other, leading to dead time (see Chapter 11, Section C). As a rough rule of thumb, if the scintillator has a decay time of τ, the maximum event rate that a detector made using that scintillator can handle is approximately $1/2\tau$. Faster scintillators detectors therefore can handle higher event rates on the detector.

The *efficiency* of the scintillator in converting a γ ray into visible light photons (photon yield) is important in determining the precision with which the energy of the interacting γ ray can be determined. This becomes relevant in many counting and imaging systems in which it is important to distinguish between γ rays that have Compton scattered in the sample (and therefore lost energy; see Chapter 6, Section C.3) and those that remain unscattered. Higher photon yield also is important in determining the positioning accuracy in many imaging systems in which it is common to share the limited number of scintillation photons among multiple PM tubes to determine the location of an interaction (see Chapter 13 and Chapter 18, Section B).

The *index of refraction* of a scintillator plays a role in determining how efficiently scintillation light can be coupled from a scintillator crystal into a PM tube. The index of refraction of the glass entrance window on a PM tube is ~1.5. Therefore for best transmission of light from the scintillator into the PM tube with minimal internal reflection at the scintillator crystal and PM tube interface, the scintillator should have an index of refraction as close to 1.5 as possible. In practice, most scintillators of interest have indices of refraction significantly higher than 1.5 and this is one reason why only a fraction of the scintillation light produced actually reaches the PM tube.

Lastly, it is important that the *emission spectrum* of the light produced by the scintillator is a good match for the quantum efficiency of the photodetector that is used to convert the scintillation light into an electronic pulse. In the case of a PM tube with a standard bialkali photocathode (see Fig. 7-14), it is apparent that scintillators that have their peak light emission in the range of 350-475 nm are optimal. NaI(Tl) is an example of a scintillator that has an emission spectrum (see Fig. 7-18) that matches the quantum efficiency of PM tubes very well. The transmission of the scintillation light through the glass used in the PM tube entrance window also must be considered. Many glasses are efficient absorbers of UV light at wavelengths significantly shorter than ~350 nm.

6. Organic Scintillators

In contrast with inorganic scintillators, the scintillation process in organic scintillators is an inherent molecular property. The

scintillation mechanism is one of molecular excitation (e.g., by absorbing energy from a γ ray or β particle) followed by a deexcitation process in which visible light is emitted. These substances are scintillators whether they are in solid, liquid, or gaseous forms.

Certain plastics (e.g., see Table 7-2) are organic scintillators and have been used for direct detection of β particles emitted from radionuclides, particularly in compact probes designed for surgical use (see Chapter 12, Section F.2). A more common application for organic scintillators, however, is to employ them in liquid form for *liquid scintillation* (*LS*) *counting*. In these systems, the scintillator is dissolved in a solvent material in a glass or plastic vial and the radioactive sample is added to this mixture. The vial is then placed in a light-tight enclosure between a pair of PM tubes to detect the scintillation events (Fig. 7-19).

LS solutions consist of four components:

1. An organic *solvent* comprises most of the solution. The solvent must dissolve not only the scintillator material but also the radioactive sample added to it. The solvent actually is responsible for most of the direct absorption of radiation energy from the sample. High-speed electrons generated by ionizing radiation events in the solvent transfer energy to the scintillator molecules, causing the scintillation effect. Commonly used solvents include di-iso-propylnapthalene (DIN) and phenylxylylethane (PXE), which are replacing traditional, more environmentally harsh solvents such as toluene and xylene.
2. The *primary solute* (or primary *fluor*) absorbs energy from the solvent and emits light. Some common primary scintillators include *p*-bis-(omethylstyryl)benzene (abbreviated as bis-MSB) and 2,5-diphenyloxazole (also known as PPO).
3. The emissions of the primary solute are not always well matched to the response characteristics of PM tubes. Therefore a *secondary solute,* or *waveshifter,* is sometimes added to the solution. The function of this material is to absorb emissions of the primary solute and reemit photons of longer wavelength, which are better matched to the PM tube response. 1,4-di-(2-5-phenyloxazole) benzene (also known as POPOP) is a commonly used secondary scintillator.
4. LS solutions frequently contain *additives* to improve some aspect of their performance, such as the efficiency of energy transfer from the solvent to the primary solute. Solubilizers (e.g., hyamine hydroxide) are sometimes added to improve the dissolution of added samples such as blood.

The precise "cocktail" of solvent, primary and secondary solutes, and additives depends on the sample type that is being measured. A wide variety of different LS cocktails optimized for different applications are available commercially.

Because of the intimate relationship between sample and detector, LS counting is the method of choice for efficient detection of particles, low-energy x rays and γ rays, and other nonpenetrating radiations. It is widely used for measurement of ^{3}H and ^{14}C. In medical applications, it is used primarily for sensitive assay of radioactivity in biologic specimens, such as blood and urine.

Liquid scintillator solution containing dissolved radioactive sample

Photomultiplier tube
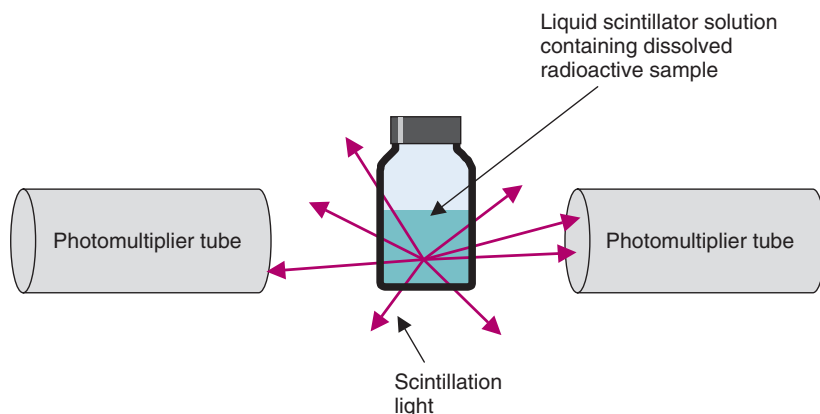
Photomultiplier tube

Scintillation light

FIGURE 7-19 Arrangement of sample and detector for liquid scintillation counting. The sample is dissolved in a liquid scintillator solution in a glass or plastic vial.

Although well suited for counting non-penetrating radiations in biologic samples, LS counters have numerous drawbacks as general-purpose radiation detectors. They are inefficient detectors of penetrating radiations such as γ rays and x rays of moderate energy because the detector solution is composed primarily of low-density, low-Z materials. In addition, liquid scintillators generally have low light output, only about one third that of NaI(Tl). This problem is worsened by the relatively inefficient light coupling from the scintillator vial to the PM tubes as compared with NaI(Tl) integral detectors.

For sample counting, special sample preparation may be required to dissolve the sample. Problems in sample preparation are discussed in Chapter 12, Section C.6. Also, the sample itself is "destroyed" when it is added to the scintillator solution.

Finally, all LS counting suffers from the problem of *quenching*. Quenching in this context refers to any mechanism that reduces the amount of light output from the sample (not to be confused with the electrical quenching that occurs in GM counters; see Section A.4). There are basically three types of LS quenching:

1. *Chemical quenching* is caused by substances that compete with the primary fluor for absorption of energy from the solvent but that are themselves not scintillators. Dissolved oxygen is one of the most troublesome chemical quenchers.
2. *Color quenching* is caused by substances that absorb the emissions of the primary or secondary solute. Blood and other colored materials are examples. Fogged or dirty containers can also produce a type of color quenching.
3. *Dilution quenching* occurs when a relatively large volume of sample is added to the scintillator solution. The effect is to reduce the concentration of primary and secondary solutes in the final solution, thus reducing the scintillator output efficiency.

Quenching can be minimized in various ways. For example, dissolved oxygen may be purged by ultrasound, and hydrogen peroxide may be added for color bleaching. However, there are no convenient ways to eliminate all causes of quenching; therefore a certain amount must be accepted in all practical LS counting. Quenching becomes a serious problem when there are wide variations in its extent from one sample to the next. This causes unpredictable variations in light output, for the same amount of radiation energy absorbed, from one sample to the next. *Quench correction* methods are employed in LS counters to account for this effect (Chapter 12, Section C.5).

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