

CHAPTER 10

NON-IMAGING DETECTORS AND COUNTERS

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10.1. INTRODUCTION

Historically, nuclear medicine has been largely an imaging based specialty, employing such diverse and increasingly sophisticated modalities as rectilinear scanning, (planar) gamma camera imaging, single photon emission computed tomography (SPECT) and positron emission tomography (PET). Non-imaging radiation detection, however, remains an essential component of nuclear medicine. This chapter reviews the operating principles, performance, applications and quality control (QC) of the various non-imaging radiation detection and measurement devices used in nuclear medicine, including survey meters, dose calibrators, well counters, intra-operative probes and organ uptake probes. Related topics, including the basics of radiation detection, statistics of nuclear counting, electronics, generic instrumentation performance parameters and nuclear medicine imaging devices, are reviewed in depth in other chapters of this book.

10.2. OPERATING PRINCIPLES OF RADIATION DETECTORS

Radiation detectors encountered in nuclear medicine may generally be characterized as either scintillation or ionization detectors (Fig. 10.1). In scintillation detectors, visible light is produced as radiation excites atoms of a crystal or other scintillator and is converted to an electronic signal, or pulse, and amplified by a photomultiplier tube (PMT) and its high voltage (500–1500 V). In ionization detectors, free electrons produced as radiation ionizes a stopping material within a sensitive volume are electrostatically collected by a bias voltage (10–500 V) to produce an electron signal. In both scintillation and ionization detectors, the ‘unprocessed’ signal is then shaped and amplified. For some types of detector, the resulting pulses are sorted by their amplitude (or pulse height), which is related to the X ray or γ ray energy absorbed in the detector.

10.2.1. Ionization detectors

Detector materials in the most common ionization detectors are gaseous and such detectors are, therefore, often known as gas filled detectors; however, as discussed in the following, solid state ionization detectors also exist. The two most widely encountered gas ionization detectors in nuclear medicine are dose calibrators and Geiger counters. The principal difference between these detectors is the magnitude of the bias voltage between the anode and cathode, as indicated graphically in Fig. 10.2. When the bias voltage is less than 300 V, ion pairs (i.e. free electrons and positive ions) produced as radiation passes through the sensitive volume may recombine, thereby preventing at least some electrons from reaching the anode and yielding an artefactually low signal. The 0–300 V range is, therefore, called the recombination region.

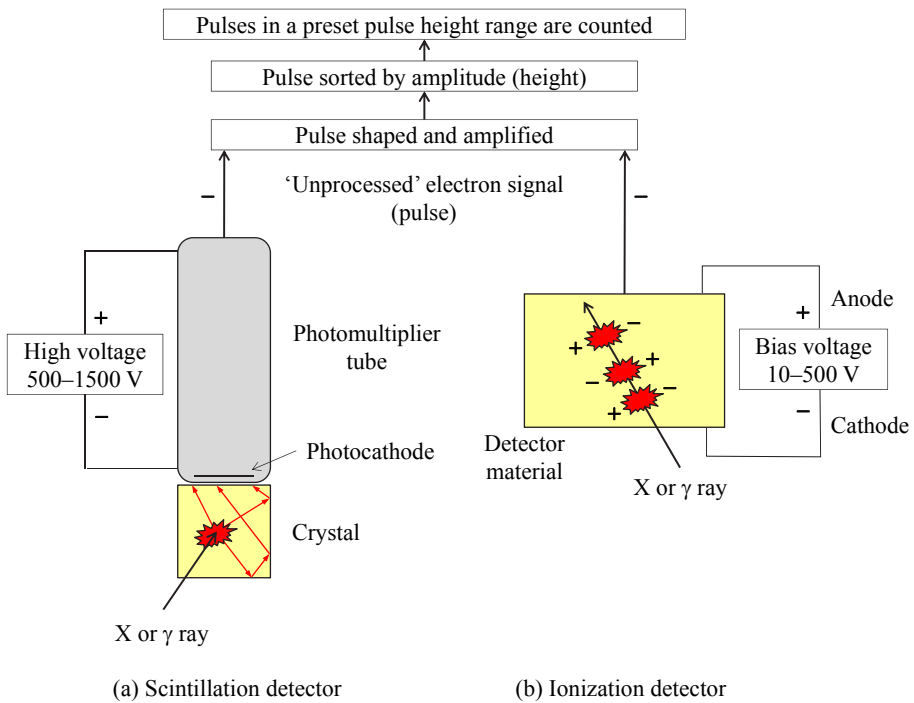


FIG. 10.1. Basic design and operating principles of (a) scintillation and (b) ionization detectors.

At a bias voltage of 300 V, all of the primary electrons (i.e. the electrons produced directly by ionization of the detector material by the incident radiation) are collected at the anode and the detector signal is, thereby, maximized. Since

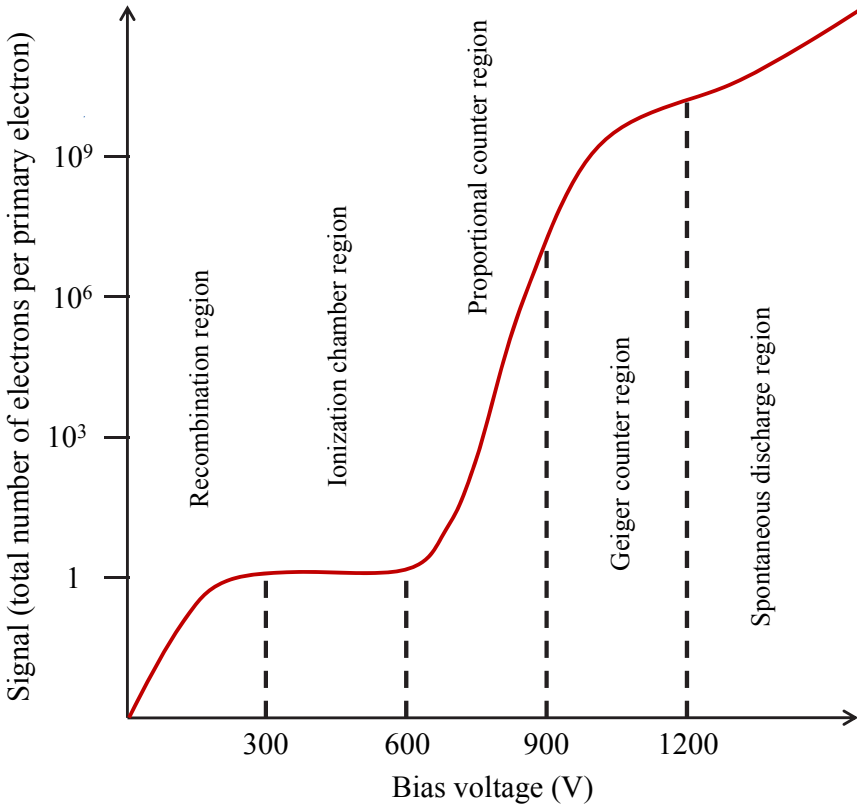


FIG. 10.2. The signal (expressed as the amplification factor, that is, the total number of electrons per primary electron produced in the detector material) as a function of the bias voltage for gas filled ionization detectors. The principal difference among such detectors is the magnitude of the bias voltage between the anode and cathode. The amplification factors and the voltages shown are approximate.

there are no additional primary electrons to collect, increasing the bias voltage further (up to 600 V) does not increase the signal. The 300–600 V range, where the overall signal is equivalent to the number of primary electrons and, therefore, proportional to the energy of the incident radiation, is called the ionization chamber region. At a bias voltage of 600–900 V, however, the large electrostatic force of attraction of the anode accelerates free electrons, as they travel towards the anode, to sufficiently high speeds to eject additional orbital electrons (i.e. secondary electrons) within the sensitive volume, contributing to an increasing overall signal — the higher the voltage, the more energetic the electrons and the more secondary electrons are added to the overall signal. The number of electrons comprising the overall signal is, thus, proportional to the

primary number of electrons and the energy of the incident radiation, and the 600–900 V range is, therefore, called the proportional counter region. As the bias voltage is increased further, beyond 900 V (up to 1200 V), free electrons (primary and secondary) are accelerated to very high speeds and strike the anode with sufficient energy to eject additional electrons from the anode surface itself. These tertiary electrons are, in turn, accelerated back to the anode surface and eject even more electrons, effectively forming an electron ‘cloud’ over the anode surface and yielding a constant overall signal even with further increase in the bias voltage. The 900–1200 V range is called the Geiger counter (or Geiger–Müller) region. Importantly, the magnitude of the charge represented by this electron cloud is independent of the number of electrons initiating its formation. Therefore, in contrast to ionization chamber and proportional counter signals, the Geiger counter signal is independent of the energy of the incident radiation. Finally, beyond a bias voltage of 1200 V, atoms within the detector material are ionized even in the absence of ionizing radiation (i.e. undergo spontaneous ionization), producing an artefactual signal; the voltage range beyond 1200 V is known as the spontaneous discharge region.

Although the bias voltage is the principal difference among different types of gas filled ionization detectors, there may be other differences. The sensitive volume, for example, may or may not be sealed. Unsealed sensitive volumes contain only air at atmospheric (ambient) pressure. For detectors with unsealed volumes, the signal must be corrected by calculation for the difference between the temperature and pressure at which the detector was calibrated (usually standard temperature and pressure: 27°C and 760 mm Hg, respectively) and the ambient conditions at the time of an actual measurement. For detectors with sealed volumes, gases other than air (e.g. argon) may be used and the gas may be pressurized, providing higher stopping power, and, therefore, higher sensitivity, than detectors having a non-pressurized gas in the sensitive volume. In addition, different geometric arrangements of the anode and cathode, such as parallel plates (used in some ionization chambers), a wire along the axis of a cylinder (used in Geiger counters), etc., may be used.

The functional properties and, therefore, the applications of the various types of ionization detector — ionization chambers, proportional counters and Geiger counters — are largely dictated by their respective bias voltage dependent signal (Table 10.1). Ionization chambers are widely used in radiation therapy to calibrate the output of therapy units and in nuclear medicine as dose calibrators (i.e. devices used to assay radiopharmaceutical activities). The relatively low sensitivity of ionization chambers is not a major disadvantage for such applications, as the radiation intensities encountered are typically rather large. The stability of the response is an important advantage, however, as it allows the use of unconditioned AC electrical power (i.e. as provided by ordinary wall

outlets). Proportional counters, because of their need for a stable bias voltage and, therefore, specialized power supplies, are restricted to research applications (e.g. in radiobiology) where both higher sensitivity and the capability of energy discrimination may be advantageous. Proportional counters often employ an unsealed, gas flow-through sensitive volume. Geiger counters, because of their high sensitivity and stability with respect to voltage (allowing the use of a portable power supply such as an ordinary battery), are widely used as survey meters to measure ambient radiation levels and to detect radioactive contamination. For such applications, sensitivity, and not energy discrimination, is critical. As with dose calibrators, Geiger counters have sealed sensitive volumes, avoiding the need for temperature–pressure corrections.

In addition to the more familiar gas filled ionization detectors, solid state ionization detectors are now available. Such detectors are based on a family of materials known as semiconductors. The pertinent difference among (crystalline) solids — conductors, insulators and semiconductors — is related to the widths of their respective electron ‘forbidden’ energy gaps. In a semiconductor, the highest energy levels occupied by electrons are completely filled but the forbidden gap is narrow enough (<2 eV) to allow radiative or even thermal excitation at room temperature, thereby allowing a small number of electrons to cross the gap and occupy energy levels among the otherwise empty upper energy levels. Such electrons are mobile and, thus, can be collected by a bias voltage, with the amplitude of the resulting signal being equivalent to the number of electrons produced by the radiation and, therefore, proportional to the radiation energy. Although many semiconductor materials have suitably large energy gaps (~ 2 eV), techniques must be available to produce crystals relatively free of structural defects. Defects (i.e. irregularities in the crystal lattice) can trap electrons produced by radiation and, thus, reduce the total charge collected, degrading the sensitivity and overall detector performance of semiconductors. Practical, reasonably economical crystal growing techniques have been developed for cadmium telluride (CdTe), cadmium zinc telluride (CZT) and mercuric iodide (HgI_2), and these detectors have been incorporated into commercial intra-operative gamma probes and, on a limited basis, small field of view gamma cameras.

TABLE 10.1. PROPERTIES OF GAS FILLED IONIZATION DETECTORS

	Ionization detector	Proportional counter	Geiger counter
Bias voltage operating range	300–600 V	600–900 V	900–1200 V
Response stable with respect to voltage? ^a	Yes	No	Yes
Sensitivity ^b	Low	Intermediate	High
Capable of energy discrimination? ^c	Yes	Yes	No
Applications	Dose calibrator	Research	Survey meter

^a The stability with respect to the bias voltage corresponds to a constant signal over the respective detector's operating voltage range. In contrast to ionization detectors and Geiger counters, proportional counters are unstable with respect to the bias voltage and, thus, require specialized, highly stable voltage sources for constancy of response.

^b The sensitivity of a detector is related to its amplification factor (see Fig. 10.2).

^c If the total number of electrons comprising the signal is proportional to the number of electrons directly produced by the incident radiation and, therefore, proportional to its energy, as in ionization detectors and proportional counters, radiations of different energies can be discriminated (i.e. separated) on the basis of the signal amplitude.

10.2.2. Scintillation detectors

In scintillation detectors, radiation interacts with and deposits energy in a scintillator, most commonly, a crystalline solid such as thallium-doped sodium iodide (NaI(Tl)). The radiation energy thus deposited is converted to visible light. As the light is emitted isotropically (i.e. in all directions), the inner surface of the light-tight crystal housing is coated with a reflective material so that light emitted towards the sides and front of the crystal are reflected back towards a PMT (Fig. 10.3); this maximizes the amount of light collected and, therefore, the overall sensitivity of the detector. Interposed between the back of the crystal and the entrance window of the PMT is the light pipe, nowadays simply a thin layer of transparent optical gel. The light pipe optically couples the crystal to the PMT and, thus, maximizes the transmission (>90%) of the light signal from the crystal into the PMT. When struck by light from the crystal, the photocathode coated on the inner surface of the PMT emits electrons. Immediately beyond the photocathode (which is at ground, that is, 0 V) is the focusing grid, maintained at a relatively low positive voltage on the order of 10 V. As electrons pass through the focusing grid, they are attracted by a relatively large positive voltage, ~300 V, on the first of a series of small metallic elements called dynodes. The resulting high speed impact of each electron results in the ejection from the dynode surface of an average of three electrons. The electrons thus ejected are then attracted by

the even larger positive voltage, ~ 400 V, on the second dynode. The impact of these electrons on the second dynode surface ejects an additional three electrons, on average, for each incident electron. Typically, a PMT has 10–12 such dynodes (or stages), each ~ 100 V more positive than the preceding dynode, resulting in an overall electron amplification factor of 3^{10} – 3^{12} for the entire PMT. At the last anode, an output signal is generated. The irregularly shaped PMT output signal is then shaped by a preamplifier and further amplified into a logic pulse that can be further processed electronically. The resulting electrical pulses, whose amplitudes (or ‘heights’) are proportional to the number of electrons produced at the PMT photocathode are, therefore, also proportional to the energy of the incident radiation. These pulses can then be sorted according to their respective heights by an energy discriminator (also known as a pulse height analyser) and those pulses with a pulse height (i.e. energy) within the preset photopeak energy window (as indicated by the pair of dashed horizontal lines overlying the pulses in Fig. 10.3) are counted by a timer/scaler.

Advantageous features of scintillation detectors include:

- High electron density (determined by mass density ρ and effective atomic number Z_{eff});
- High light output;
- For certain applications such as PET, speed of light emission.

High mass density and effective atomic number maximize the crystal stopping power (i.e. linear attenuation coefficient μ) and, therefore, sensitivity. In addition, a higher atomic number crystal will have a higher proportion of photoelectric than Compton interactions, thus facilitating energy discrimination of photons which underwent scatter before entering the crystal. High light output reduces statistical uncertainty (noise) in the scintillation and associated electronic signal and, thus, improves energy resolution and scatter rejection. Other detector considerations include:

- Transparency of the crystal to its own scintillations (i.e. minimal self-absorption);
- Matching of the index of refraction η of the crystal to that of the photodetector (specifically, the entrance window ($\eta \approx 1.5$) of the PMT);
- Matching of the scintillation wavelength to the light response of the photodetector (the PMT photocathode, with maximum sensitivity in the 390–410 nm, or blue, wavelength range);
- Minimal hygroscopic behaviour.

To date, the most widely used scintillators in nuclear medicine include: NaI(Tl), bismuth germanate (BGO), cerium-doped lutetium oxyorthosilicate (LSO(Ce) or LSO) and cerium-doped gadolinium oxyorthosilicate (GSO(Ce) or GSO). NaI(Tl) is used in γ cameras/SPECT systems, well counters and organ uptake probes, and remains the most widely used scintillator in clinical practice; BGO, LSO and GSO are the scintillators of choice in PET scanners because of their higher stopping power for the 511 keV positron–negatron annihilation photons. Thallium- and sodium-doped caesium iodide (CsI(Tl) and CsI(Na), respectively) and cadmium tungstate as well as NaI(Tl), BGO and LSO have also been used in intra-operative probes.

10.3. RADIATION DETECTOR PERFORMANCE

Radiation detectors may be quantitatively characterized by many different performance parameters, particularly for those detectors such as γ cameras which localize (image) as well as count radiation. For non-imaging radiation detectors and counters, however, the most important performance parameters are sensitivity (or efficiency), energy resolution and count rate performance (or ‘speed’).

10.3.1. Sensitivity

Sensitivity (or efficiency) is the detected count rate per unit activity (e.g. in counts per minute per megabecquerel). As the count rate detected from a given activity is highly dependent on the source–detector geometry and intervening media, characterization of sensitivity can be ambiguous. There are two distinct components of overall sensitivity, geometric sensitivity and intrinsic sensitivity. Geometric sensitivity is the fraction of emitted radiations which intersect, or strike, the detector, that is, the fraction of the total solid angle subtended at the detector by the source. It is, therefore, directly proportional to the radiation-sensitive detector area and, for a point source, inversely proportional to the square of the source–detector distance. Intrinsic sensitivity is the fraction of radiation striking the detector which is stopped within the detector. Intrinsic sensitivity is directly related to the detector thickness, effective atomic number and mass density, and decreases with increasing photon energy, since higher energy photons are more penetrating and are more likely to pass through a detector without interacting.

Characteristic X rays and γ rays are emitted from radioactively decaying atoms with well defined discrete energies. Even in the absence of scatter, however, output pulses from absorption of these radiations will appear to originate over a range of energies, reflecting the relatively coarse energy resolution of the detector. For this reason, many radiation detectors employ some sort of energy-selective

counting using an energy range, or window, such that radiations are only counted if their detected energies lie within that range (Figs. 10.3 and 10.4(a)). At least for scintillation detectors, a so-called ‘20% photopeak energy window’, $E_\gamma \pm 10\%$ of E_γ , (e.g. 126–154 keV for the 140 keV γ ray of ^{99m}Tc) is employed, where E_γ is the photopeak energy of the X ray or γ ray being counted. For such energy-selective counting, overall sensitivity appears to increase as the photopeak energy window is widened. However, this results in acceptance of more scattered as well as primary (i.e. unscattered) radiations.

For each radionuclide and energy window (if applicable) for which a particular detector is used, the detector should be calibrated, that is, its sensitivity (e.g. in cpm/MBq) S determined, at installation and periodically thereafter:

$$S = \frac{R_g - R_b}{A_0 e^{-\lambda \Delta t}} \quad (10.1)$$

where

- R_g is the gross (i.e. total) count rate (cpm) of the radionuclide source (RS);
- R_b is the background (BG), or blank, count rate (cpm);
- A_0 is the activity (MBq) of the radionuclide source at calibration;
- λ is the physical decay constant (in month^{-1} or a^{-1} , depending on the half-life) of the calibration radionuclide;

and Δt is the time interval (in months or years, respectively, again depending on the half-life) between the calibration of the radionuclide and the current measurement.

As noted, sensitivity is highly dependent on the source–detector counting geometry (including the size and shape of the source and the source–detector distance), and the measured value, thus, applies exactly only for the geometry used for the measurement.

10.3.2. Energy resolution

Energy resolution quantifies the ability of a detector to separate, or discriminate, radiations of different energies. As illustrated in Fig. 10.4(b), energy resolution is generally given by the width of the bell shaped photopeak, specified as the full width at half maximum (FWHM = ΔE) height expressed as a percentage of the photopeak energy E_γ , $\text{FWHM} (\%) = \frac{\Delta E}{E_\gamma} 100\%$. It is related

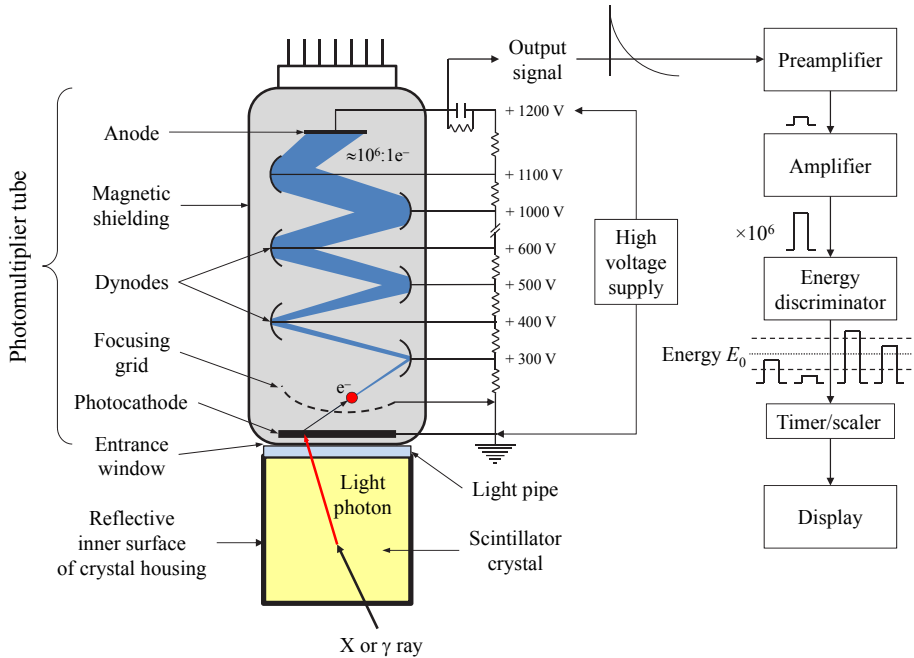


FIG. 10.3. The basic design and operating principle of photomultiplier tubes and scintillation detectors.

to the Poisson ‘noise’, or statistical uncertainty, inherent in the detection process. The importance of energy resolution lies in scatter rejection, particularly for imaging detectors. Radiation loses energy when undergoing Compton scatter within the source and the lower energy scattered radiations may, therefore, be discriminated from the primary radiations. However, the finite energy resolution of radiation detectors (i.e. the width of the photopeak in the energy spectrum) means that there will be overlap of scattered and primary radiations, as illustrated in Fig. 10.4(a). As energy resolution improves (i.e. the FWHM (%) decreases and the photopeak becomes narrower), the separation of unscattered and scattered radiations increases and more counts corresponding to scattered radiation may be eliminated, while discarding fewer counts corresponding to unscattered radiation.

10.3.3. Count rate performance (‘speed’)

Radiation detectors have a finite dead time or pulse resolving time τ — typically 5–10 μs for modern scintillation detectors — and associated count losses. The dead time is the length of time required for a counting system to

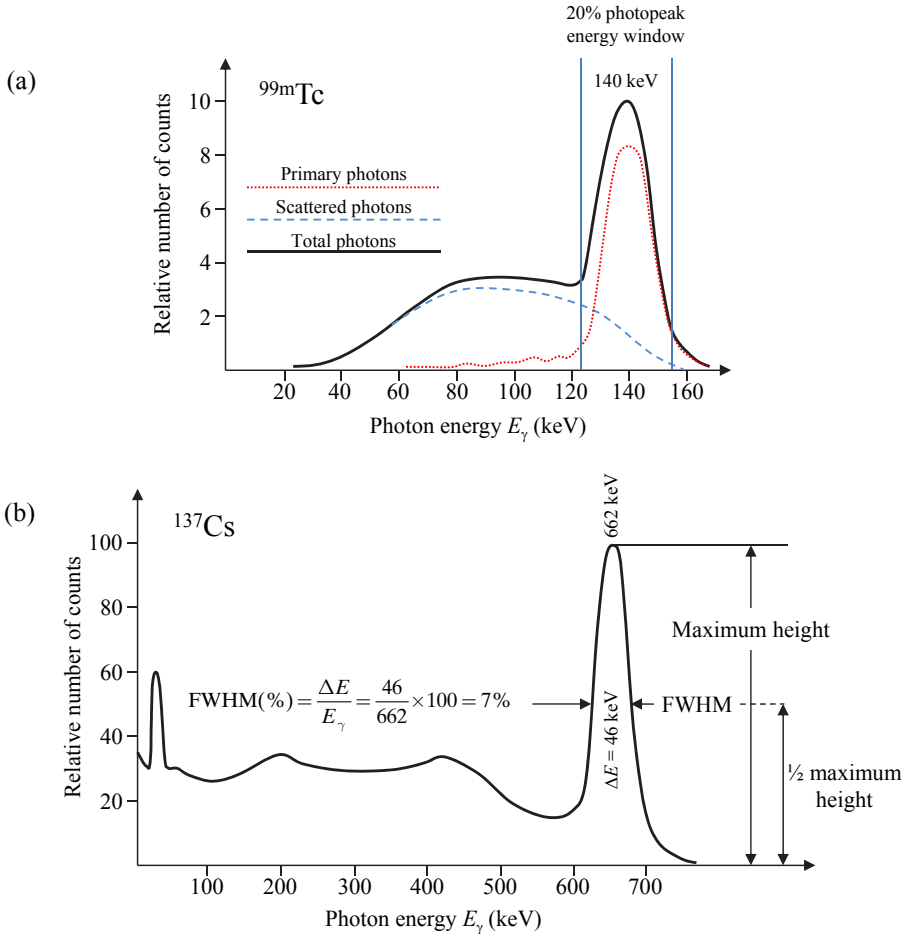


FIG. 10.4. (a) Energy spectrum for the 662 keV γ rays emitted by ^{137}Cs , illustrating the definition of energy resolution as the percentage full width at half maximum (FWHM) of the photopeak energy E_γ . (b) Energy spectrum for the 140 keV γ rays emitted by ^{99m}Tc , illustrating the contributions of primary (unscattered) and scattered radiation counts. In (a) and (b), the energy spectra were obtained with a thallium-doped sodium iodide (NaI(Tl)) scintillation detector.

record an event, during which additional events cannot be recorded. As a result, the measured count rate is lower than the actual count rate. Radiation detectors are characterized in terms of count rate performance as either non-paralysable or paralysable (Fig. 10.5). In non-paralysable systems, only radiation which is actually counted prevents the counting of subsequent radiation interacting with the detector during the dead time of that preceding radiation. In a paralysable

detector, however, even radiation which is not counted (i.e. which interacts with the detector during the dead time of a previous event) prevents counting of subsequent incoming radiations during the time interval corresponding to its dead time. Geiger counters (with quenching gas) behave as non-paralysable systems but most detectors, including scintillation detector based systems, such as well counters, γ cameras and PET scanners, are paralysable. Modern scintillation detectors generally incorporate automated algorithms to yield count rates corrected for dead time count losses.

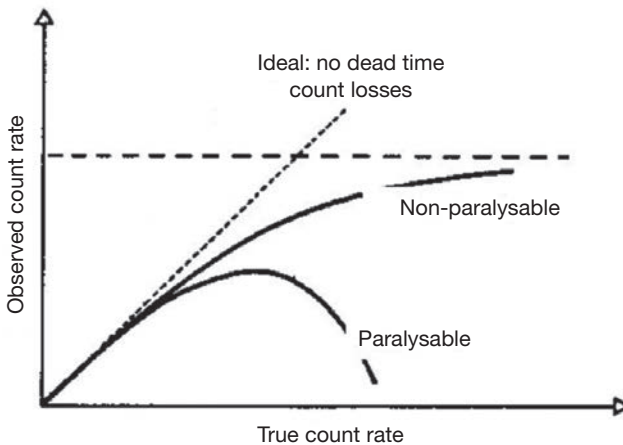


FIG. 10.5. The observed versus true count rates for paralysable and non-paralysable radiation detectors. For paralysable detectors, the observed count rate increases to a maximum value with increasing true count rate (e.g. with increasing activity) and then decreases as the true count rate is further increased. For non-paralysable detectors, the observed count rate also increases with increasing true count rate, asymptotically approaching a maximum value as the true count rate is further increased. In both cases, the maximum observed count rate is directly related to the detector's dead time τ .

10.4. DETECTION AND COUNTING DEVICES

10.4.1. Survey meters

Survey meters, an essential component of any radiation safety programme, are portable, battery operated, gas filled ionization detectors (or, to a much more limited extent, solid state scintillation detectors) used to monitor ambient radiation levels, that is, exposure rates (e.g. in coulombs per kilogram of air per hour ($C \cdot kg^{-1} \cdot h^{-1}$)) or count rates (e.g. in cpm). Among ionization detector survey meters, so-called 'cutie-pies' are relatively low sensitivity ionization chambers

(i.e. are operated at a relatively low potential difference between the anode and cathode) and are designed for use where relatively high fluxes of X rays and γ rays are encountered. The more familiar Geiger counters are operated at a high potential difference (Fig. 10.2), providing a high electron amplification factor and, thus, high sensitivity. Geiger counters are, therefore, well suited for low level surveys, for example, checking for radioactive contamination. Both cutie-pies and Geiger counters are generally calibrated in terms of exposure rate. As an ionization chamber, the cutie-pie's electron signal depends on the energy of the detected X rays or γ rays and is, therefore, directly related to the exposure for all radionuclides. For Geiger counters, on the other hand, signal pulses have the same amplitude regardless of the energy of the incoming radiation. Thus, Geiger counter calibration results apply only to the particular radionuclide(s) used to calibrate the counter (see below). Solid state detectors employ a non-air-equivalent crystal as the detection medium and, thus, cannot measure exposure rates, only count rates.

10.4.2. Dose calibrator

The dose calibrator, used for assaying activities in radiopharmaceutical vials and syringes and in other small sources (e.g. brachytherapy sources), is a pressurized gas filled ionization chamber with a sealed sensitive volume configured in a 'well'-type geometry. While the intrinsic sensitivity of the dose calibrator, as that of other gas filled detectors, is relatively low, the well-type configuration of its sensitive volume provides high geometric efficiency¹, making the overall sensitivity entirely adequate for the relatively high radiopharmaceutical activities (of the order of 10–100 MBq) typically encountered in clinical nuclear medicine. Dose calibrators are equipped with isotope specific push-buttons and/or a potentiometer (with isotope-specific settings provided) to adjust for differences in energy dependent response and to thereby yield accurate readouts of activity (i.e. kBq or MBq) for any radioisotope.

10.4.3. Well counter

Well counters are used for high sensitivity counting of radioactive specimens such as blood or urine samples or 'wipes' from surveys of removable contamination (i.e. 'wipe testing'). Such counting results can be expressed in

¹ The solid angle subtended at the centre of a sphere by the total surface of the sphere is 4π steradians; a steradian is the unit of solid angle. A well-type detector configuration approximates a point source completely surrounded by a detector, yielding a per cent geometric efficiency of 100%, and is, therefore, referred to as a ' 4π ' counting geometry.

terms of activity (e.g. MBq) using the measured isotope specific calibration factor (cpm/MBq) (see Eq. (10.1)). Such devices are generally comprised of a cylindrical scintillation crystal (most commonly, NaI(Tl)) with a circular bore (well) for the sample drilled part-way into the crystal and backed by a PMT and its associated electronics. An alternative design for well counters is the so-called 'through-hole' detection system in which the hole is drilled through the entire crystal. The through-hole design facilitates sample exchange, and because samples are centred lengthwise in the detector, yields a more constant response for different sample volumes as well as slightly higher sensitivity than the well counters. In both the well and through-hole designs, the crystal is surrounded by thick lead shielding to minimize the background due to ambient radiation.

Scintillation counters are often equipped with a multichannel analyser for energy (i.e. isotope) selective counting and an automatic sample changer for automated counting of multiple samples. Importantly, because of their high intrinsic and geometric efficiencies (resulting from the use of a thick crystal and a well-type detector configuration, respectively), well counters are extremely sensitive and, in fact, can reliably be used only for counting activities up to ~100 kBq; at higher activities, and even with dead time corrections applied, dead time counting losses may still become prohibitive and the measured counts inaccurate. Modern well counters often include an integrated computer which is used to create and manage counting protocols (i.e. to specify the isotope, energy window, counting interval, etc.), manage sample handling, and apply background, decay, dead time and other corrections, and, thus, yield dead time-corrected net count rate decay corrected to the start of the current counting session.

10.4.4. Intra-operative probes

Intra-operative probes (Fig. 10.6), small hand-held counting devices, are now widely used in the management of cancer, most commonly to more expeditiously identify and localize sentinel lymph nodes and, thereby, reduce the need for more extensive surgery as well as to identify and localize visually occult disease at surgery following systemic administration of a radiolabelled antibody or other tumour-avid radiotracer. Although intra-operative probes have been used almost exclusively for counting X rays and γ rays, beta (electron and positron) probes constructed with plastic scintillators have also been developed. In addition, small (~10 cm) field of view intra-operative γ cameras have recently become available. Intra-operative γ probes are available with either scintillation or semiconductor (ionization) detectors. Scintillation detector based probes have the advantages of relatively low cost and high sensitivity (mainly because of their greater thickness, ~10 mm versus only ~1 mm in ionization detectors), especially for medium to high energy photons.

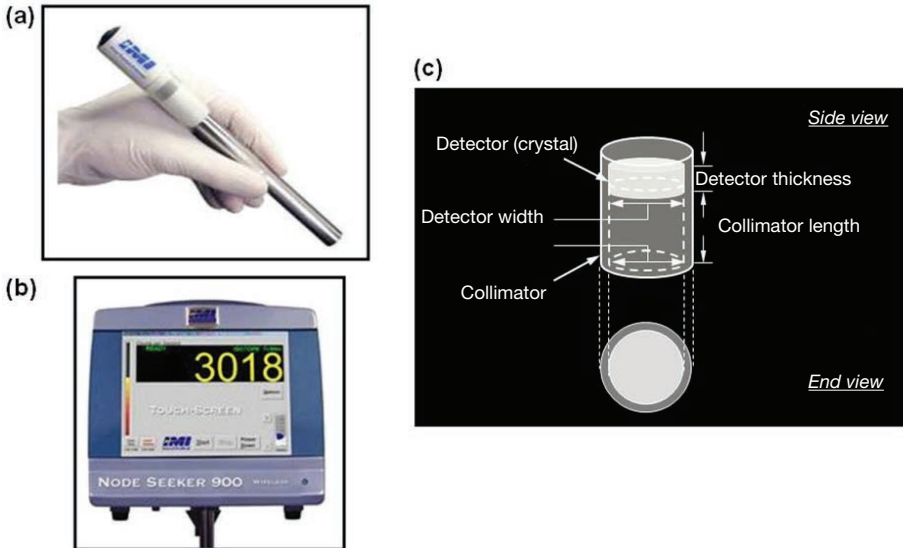


FIG. 10.6. A typical intra-operative probe (Node Seeker 900, Intra Medical Imaging LLC, Los Angeles, CA, United States of America). (a) Hand-held detector. (b) Control and display unit which not only displays the current count rate but also often emits an audible signal, the tone of which is related to the count rate, somewhat analogous to the audible signal produced by some Geiger counters. (c) A diagram of the detector and collimator assembly of a typical intra-operative probe, illustrating that the detector (crystal) is recessed from the collimator aperture. (Courtesy of Intra Medical Imaging LLC, Los Angeles, CA, USA.)

Disadvantages include bulkiness, and relatively poor energy resolution and scatter rejection relative to semiconductor based probes. In some scintillation–detector intra-operative probes, the light signal from the crystal is guided to a remote PMT through a flexible fibre-optic cable, allowing the probe assembly to be made relatively light and compact, and more like a surgical instrument. However, the significant loss of light in the long cable makes it more difficult to separate scatter from unscattered X rays and γ rays.

On the other hand, semiconductor based probes are compact and have excellent energy resolution and scatter rejection. To minimize structural imperfections which degrade energy resolution, semiconductor detectors are made relatively thin (only ~ 1 mm), but at the cost of lower intrinsic sensitivity. The main disadvantage of semiconductor detectors remains their limited thickness and resulting lower sensitivity, especially for medium to high energy X rays and γ rays. Nonetheless, while scintillation detectors can be made thicker and, therefore, more sensitive, semiconductor detectors produce more electrons per X ray and γ ray stopped, and, therefore, have a superior energy resolution. To

date, the few clinical studies directly comparing scintillation and semiconductor intra-operative probes have not provided a clear choice between the two types of probe.

10.4.5. Organ uptake probe

Historically, organ uptake probes have been used almost exclusively for measuring thyroid uptakes and are, thus, generally known as ‘thyroid’ uptake probes.² Thyroid uptake (i.e. the decay-corrected per cent of administered activity in the thyroid) may be measured following oral administration of ¹³¹I-iodide, ¹²³I-iodide or ^{99m}Tc-pertechnetate. The uptake probe is a radionuclide counting system consisting of a wide-aperture, diverging collimator, a NaI(Tl) crystal (typically ~5 cm thick by ~5 cm in diameter), a PMT, a preamplifier, an amplifier, an energy discriminator (i.e. an energy window) and a gantry (stand) (Figs 10.7(a) and (b)). Commercially available thyroid uptake probes are generally supplied as integrated, computerized systems with automated data acquisition and processing capabilities, yielding results directly in terms of per cent uptake.

Each determination of the thyroid uptake includes measurement of the thyroid (i.e. neck) count rate, the ‘thigh’ background count rate (measured over the patient’s thigh and presumed to approximate the count contribution of extra-thyroidal neck activity), the standard count rate (often counted in a neck phantom simulating the thyroid/neck anatomy) and the ambient (i.e. ‘room’) background, with a 1–5 min counting interval for each measurement. Based on the foregoing measurements, and knowing the fraction of the administered activity which is in the standard, the thyroid uptake is calculated as follows:

$$\text{uptake (\%)} = \frac{C_{\text{neck}}/t_{\text{neck}} - C_{\text{thigh}}/t_{\text{thigh}}}{C_{\text{standard}}/t_{\text{standard}} - C_{\text{room}}/t_{\text{room}}} \times F \times 100\% \quad (10.2)$$

where

C is the total counts;

t is the measurement time;

and F is the fraction of administered activity in the standard.

² At one time, organ uptake probes were also used to measure kidney time–activity data for the evaluation of renal function. In addition, organ uptake probes have been adapted to such well counter applications as counting of blood samples and wipes.

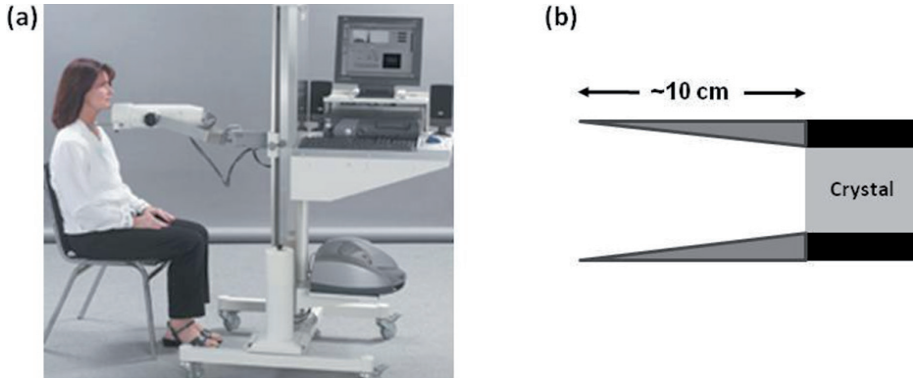


FIG. 10.7. (a) A typical organ ('thyroid') uptake probe system, including an integrated computer, set-up for a thyroid uptake measurement (AtomLab 950™ Thyroid Uptake System, Biodex Medical Systems, Shirley, NY, USA). The rather large neck to collimator aperture distance (typically of the order of 30 cm) should be noted. Although this reduces the overall sensitivity of the measurement of the neck count rate, it serves to minimize the effect of the exact size, shape and position of the thyroid, and the distribution of radioisotope within the gland. (b) A diagram (side view) of the open, or 'flat-field', diverging collimator typically used with thyroid uptake probes. (Courtesy of Biodex Medical Systems, Inc, Shirley, NY, USA.)

By including measurement of a standard activity with each uptake determination, corrections for radioactive decay and day to day variation in system sensitivity are automatic. This approach is sometimes known as the 'two-capsule' method, since one ^{131}I capsule is administered to the patient while a second, identical capsule serves as the standard and is counted with each uptake measurement. Alternatively, the patient capsule itself can be measured immediately before it is administered and then each subsequent uptake value for radioactive decay can be corrected from the time of measurement to the time of administration (by multiplying the right side of Eq. (10.2) by $e^{\lambda\Delta t}$ where λ is the physical decay constant of the administered isotope and Δt is the administration to measurement time interval). This is sometimes known as the 'one-capsule' method. For either the one- or two-capsule method, the fraction of administered activity in the standard is unity. Some centres administer radioiodine as a solution, which is more cost effective, rather than as a capsule. The standard is typically some dilution of the administered solution and the fraction of administered activity in the standard is, therefore, an independently determined value less than unity; for example, if the activity in the standard solution were 1/100th of the value in the administered solution, the value would equal 0.01.

Thyroid uptake measurements are now often performed by region of interest analysis of planar scintigraphic images of the neck and of a standard (i.e. phantom) acquired with a γ camera with parallel-hole collimation.

Organ uptake probes have also been used to measure total body activity, for example, as part of individualized dosimetry based radioiodine treatment of thyroid cancer. For this application, the patient may serve as his or her own standard by measuring the patient's total body count rate shortly (30–60 min) after administration of the radioisotope — to allow it to disperse somewhat throughout the body — but before the patient has voided or otherwise excreted any of the administered activity; in Eq. (10.3) below, this is designated time zero. Whole body measurements are performed with the collimator removed from the probe, the crystal oriented horizontally and at a height above the floor corresponding to the mid-height of the patient, either seated or standing, at a distance of ~ 3 m from the crystal. Further, anterior and posterior (i.e. conjugate-view) measurements are performed by having the patient facing towards and then away from the crystal for the respective measurements. The whole body activity (i.e. the per cent of administered activity in the body) is then calculated based on the geometric mean of the anterior and posterior count rates:

$$\text{Total body activity (\%)} = \frac{\left[\left(\frac{A}{t_A} - \frac{B}{t_B} \right) \times \left(\frac{P}{t_P} - \frac{B}{t_B} \right) \right]^{1/2}}{\left[\left(\frac{A(0)}{t_{A(0)}} - \frac{B(0)}{t_{B(0)}} \right) \times \left(\frac{P(0)}{t_{P(0)}} - \frac{B(0)}{t_{B(0)}} \right) \right]^{1/2}} \times 100\% \quad (10.3)$$

where

A and P are the anterior and posterior total body counts, respectively;

B is the room (background) counts;

t_A , t_P and t_B are the counting intervals for anterior, posterior and room counts, respectively;

and (0) indicates the same quantities at time zero.

As above, the total body activity may be corrected for radioactive decay from the time of measurement to the time of administration (by multiplying the right side of Eq. (10.3) by $e^{\lambda \Delta t}$ where λ is the physical decay constant of the administered isotope and Δt is the administration to measurement time interval).

10.5. QUALITY CONTROL OF DETECTION AND COUNTING DEVICES

QC, which may be defined as an established set of ongoing measurements and analyses designed to ensure that the performance of a procedure or instrument is within a predefined acceptable range, is a critical component of routine nuclear medicine practice. The following is a brief review of routine QC procedures for non-imaging nuclear medicine instrumentation.

Documenting of QC procedures and organized, retrievable records of the results of such procedures are requirements of a sound, compliant QC programme. A written description of all QC procedures, including the acceptable (or tolerance) range of the results of each such procedure and the corrective action for an out of tolerance result, should be included in the facility's procedure manual. For each procedure, the written description should be signed and dated by the facility director, physicist or other responsible individual. For each QC test performed, the following data, as well as the initials or signature of the individual performing the test, should be recorded on a structured and suitably annotated form:

- The test performed;
- The date and time of the test;
- The make, model and serial number of the device tested;
- The make, model, serial number, activity at calibration and date of calibration of any reference source(s) used;
- The result(s) of the test;
- A notation indicating whether the test result was or was not acceptable (i.e. was or was not within the specified tolerance range).

Such records should be archived in chronological order in a secure but reasonably accessible location. It is generally helpful to track the results of QC tests longitudinally (e.g. in the form of a graph of the numerical result versus the date of the test). In this way, long term trends in instrument performance, often imperceptible from one day to the next, may become apparent. Increasingly, of course, such records are maintained in electronic form. It is advisable, however, to also maintain records in hard copy form, both as a backup and for convenient review by regulators and other inspectors.

10.5.1. Reference sources

QC tests of nuclear medicine instrumentation are often performed not with the radionuclides that are used clinically but with longer lived surrogate radionuclides in the form of so-called reference sources. Such standards are

commercially available in various activities and geometries, depending on the application. Importantly, in the USA, the certified activities of such reference sources must be traceable to the National Institute of Standards and Technology (NIST), formerly the National Bureau of Standards. NIST traceability helps ensure the accuracy of the calibrated activity. As such reference sources are long lived, a single standard may be used for months to years, avoiding the need to prepare sources on a daily or weekly basis and, thereby, avoiding possible inaccuracies in dispensing activity as well as the possibility of radioactive contamination. On the other hand, as with all sealed sources, reference sources must be checked for leakage of radioactivity (i.e. ‘wipe-tested’) periodically and an up to date inventory of such standards must be maintained. Reference sources are still radioactive at the end of their useful lifespan and must, therefore, be returned to the vendor or a third party or otherwise disposed of as radioactive waste.

A long lived radionuclide comprising a reference source must match, in terms of the frequency and energy of its X ray and γ ray emissions, the clinical radionuclide for which it acts as a surrogate in order to ensure that instrument responses to the clinical radionuclide and to its surrogate are comparable. Surrogate radionuclides commonly used in nuclear medicine and their physical properties and applications are summarized in Table 10.2.

10.5.2. Survey meter

QC tests of survey meters generally include a daily battery check, with a display indicating whether the voltage supplied by the battery is within the acceptable operating range. In order to confirm that the survey meter has not been contaminated (i.e. yields a reproducibly low exposure or count rate in the absence of radioactivity), the background exposure or count rate should be measured daily in an area remote from radioactive sources within the nuclear medicine facility, if such an area is reasonably accessible. In addition, survey meters should be checked daily for constancy of response by measuring the exposure or count rate of a long lived reference source in a reproducible measurement geometry. Aside from the short term decay of the reference source, the measured day to day exposure or count rates should agree within 10%; if not, the meter should be recalibrated.

Survey meters should be calibrated — that is, checked for accuracy — using suitable long lived reference sources at installation, annually and after any repair. If the source is ‘small’ (compared to the mean free path of its emitted X rays and γ rays within the material comprising the source) and the distance between the source and meter ‘large’ (compared to the dimensions of the source),

then a point-source geometry is approximated and the expected dose rate \dot{D} in air is given by the inverse square law:

$$\dot{D} = \frac{A_0 e^{-\lambda \Delta t} \Gamma_{\delta}}{d^2} \quad (10.4)$$

where

- A_0 is the activity of the reference source at calibration;
- λ is the physical decay constant of the radionuclide comprising the reference source;
- Δt is the time interval between the calibration of the reference source and the current measurement;
- Γ_{δ} is the air kerma rate constant (the subscript δ indicates that only photons with energies greater than δ , typically set at 20 keV, are included) of the radionuclide comprising the reference source;

and d is the distance between the reference source and the meter (Table 10.2).

The dose rates should be measured on each scale and, by appropriate adjustment of the source–meter distance, with two readings (~20% and ~80% of the maximum) on each scale. For all readings, the expected and measured dose rates should agree within 10%.

Many nuclear medicine facilities have their survey meters calibrated by the institutional radiation safety office or by a commercial calibration laboratory. In addition to a calibration report (typically, a one page document) specifying the reference source(s) used, the measurement procedure, and the measured and expected exposure rates, a dated sticker summarizing the calibration results should be affixed to the meter itself.

10.5.3. Dose calibrator

Among routine dose calibrator QC tests³, constancy must be checked daily and accuracy and linearity at least quarterly; daily checks of accuracy are recommended. For the constancy test, an NIST-traceable reference source, such as ⁵⁷Co, ⁶⁸Ge or ¹³⁷Cs (Table 10.2), is placed in the dose calibrator and the

³ At the installation of a dose calibrator, the geometry dependent response for ^{99m}Tc must be measured and volume dependent (2–25 mL) correction factors relative to the ‘standard’ volume (e.g. 10 mL) derived. This procedure is required periodically following installation.

TABLE 10.2. LONG LIVED RADIONUCLIDES COMPRISING REFERENCE SOURCES FOR INSTRUMENTATION QUALITY CONTROL

Radionuclide	Half-life	Physical decay constant λ	Photopeak energy E_γ and frequency of principal X ray or γ ray	Air kerma rate constant Γ_δ ($\text{mGy} \cdot \text{m}^2 \cdot \text{h}^{-1} \cdot \text{GBq}^{-1}$) ^a	Geometry and activity	Quality control application
⁵⁷ Co	272 d	0.00254/d	122 keV (86%)	14.1	Test tube-size rod, ~37 kBq Vial/small bottle, 185–370 MBq	Well counter constancy and accuracy Dose calibrator accuracy and constancy
⁶⁸ Ge ^b	287 d	0.00241/d	511 keV (178%)	129	Test tube-size rod, 37 kBq Vial/small bottle, 185–370 MBq	Well counter constancy and accuracy Dose calibrator accuracy and constancy
¹³⁷ Cs	30 a	0.0231/a	662 keV (86%)	82.1	Test tube-size rod, 37 kBq Vial/small bottle, 185–370 MBq	Well counter constancy and accuracy Dose calibrator accuracy and constancy

^a The air kerma rate constant Γ_δ is equivalent to the older specific γ ray constant Γ .

^b Germanium-68 in a sealed source is in secular equilibrium with its short lived, positron emitting daughter ⁶⁸Ga (half-life: 68 min).

activity reading on each scale recorded; day to day readings should agree within 10%. For the accuracy test (sometimes also known as the energy linearity test), at least two of the foregoing NIST-traceable reference sources are separately placed in the dose calibrator and the activity reading on each scale recorded. For each source, the measured activity on each scale and its current actual activity should agree within 10%.



FIG. 10.8. Set of lead-lined plastic sleeves (Calicheck™ Dose Calibrator Linearity Test Kit, Calicheck, Cleveland, OH, USA) for evaluation of dose calibrator linearity by the shield method. The set is supplied with a 0.64 cm thick lead base, a colour coded unlined sleeve (to provide an activity measurement equivalent to the zero time point measurement of the decay method) and a six colour coded lead-lined sleeve providing attenuation factors nominally equivalent to decay over 6, 12, 20, 30, 40 and 50 h, respectively. (Courtesy of Calicheck, Cleveland, OH, USA.)

The quarterly check of linearity by the so-called ‘decay method’ begins with a high activity (~ 37 GBq), independently calibrated ^{99m}Tc source and its activity is assayed at 12 h intervals over three consecutive days. Over that time, equivalent to twelve half-lives of ^{99m}Tc , the activity decays to ~ 10 MBq. The measured activities are then plotted versus time on a semi-logarithmic graph and the best fit straight line drawn through the data points thus plotted (either ‘by eye’ or using a least squares curve-fitting algorithm). For each data point, the difference between the measured activity and the activity on the best fit straight line at that point should be less than 10%. An alternative approach to checking linearity is the ‘shield method’ in which lead sleeves of increasing thickness are placed in the dose calibrator with a ^{99m}Tc source (Fig. 10.8). By interposing increasing ‘decay-equivalent’ thicknesses (as specified by the manufacturer for

the set of lead sleeves) between the source and the dose calibrator's sensitive volume, a decay-equivalent activity is measured for each sleeve. While the shield method is much faster than the decay method for checking linearity (taking minutes instead of days), an initial decay based calibration of the set of sleeves is recommended to accurately determine the actual decay equivalence of each shield.

10.5.4. Well counter

The routine QC tests for well counters include checks of the photopeak energy window (i.e. energy peaking) if the counter is equipped with a multichannel analyser, background, constancy and efficiency (or sensitivity). Prior to counting samples containing a particular radionuclide, the energy spectrum should be checked to verify that the counter is properly 'peaked', that is, that the radionuclide's photopeak coincides with the preset photopeak energy window⁴. For each photopeak energy window used, the background count rate should be checked daily. Importantly, electronic noise as well as ambient radiation levels, which may be relatively high and variable in a nuclear medicine facility, will produce a non-zero and potentially fluctuating background count rate. Furthermore, even trace contamination of the counting well will produce inaccurately high count rate values. Accordingly, a 'blank' (i.e. an empty counting tube or vial) should always be included to determine the current background count. To check constancy, at least one NIST-traceable reference source (Table 10.2) should likewise be counted each day; day to day net (i.e. gross minus background) count rates should agree within 10%.

In addition, as noted above, for each radionuclide for which a particular well counter is used, the counter should be calibrated — that is, its efficiency (sensitivity) (in cpm/kBq) determined — at installation, annually and after any repair (Eq. (10.1)).

10.5.5. Intra-operative probe

In addition to daily battery and background checks (as done for survey meters), QC tests of intra-operative probes should include a daily bias check for both the primary and any backup battery to verify that bias voltage (or high voltage) is within the acceptable range. As intra-operative probes may not provide a display of the energy spectrum, it may not be possible to visually check

⁴ Isotope specific radionuclide counting or imaging with a scintillation detector is commonly performed using a 20% photopeak energy window, equivalent to an energy range of $E_\gamma \pm 10\%$ where E_γ is the X ray or γ ray energy of the radionuclide.

that the probe is properly peaked, that is, that the photopeak coincides with the preset photopeak energy window. The lower counts or count rates resulting from an inappropriate energy window may, therefore, go unnoticed. Thus, a long lived reference source or set of reference sources (such as ^{57}Co , ^{68}Ge and/or ^{137}Cs (Table 10.2)) should be available for daily checks of count rate constancy; a marked change (e.g. $>\pm 10\%$) in the net count rate from one day to the next may indicate an inappropriate energy window setting or some other technical problem. Ideally, the reference sources should each be incorporated into some sort of cap that fits reproducibly over the probe so that spurious differences in count rates due to variations in source–detector geometry are avoided.

10.5.6. Organ uptake probe

Aside from differences in counting geometry and sensitivity, uptake probes and well counters actually have very much in common and the QC procedures — checks of the photopeak energy window, background, constancy and efficiency — are, therefore, analogous. Importantly, however, efficiency should be measured more frequently — for each patient — than for a well counter, so that the probe net count rates can be reliably converted to thyroid uptakes for individual patients.

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