CHAPTER 8

GENERIC PERFORMANCE MEASURES

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8.1. INTRINSIC AND EXTRINSIC MEASURES

8.1.1. Generic nuclear medicine imagers

The generic nuclear medicine imager, whether a gamma camera, single photon emission computed tomography (SPECT) system or positron emission tomography (PET) scanner, comprises several main components: a detection system, a form of collimation to select γ rays at specific angles, electronics and a computing system to create the map of the radiotracer distribution. This section discusses these components in more detail.

The first stage of a generic nuclear medicine imager is the detection of the γ rays emitted by the radionuclide. In the case of PET, the radiation of interest are the 511 keV annihilation photons that result from the interaction of the positron emitted by the radionuclide with an electron in the tissue. For general nuclear medicine and SPECT, there is one or sometimes more than one γ ray of interest, with energies in the range of <100 to >400 keV.

The γ rays are detected when they interact and deposit energy in the crystal(s) of the imaging system. There are two main types of detector: crystals that give off light that can be converted to an electrical signal when the γ ray interacts (‘scintillators’) and semiconductors, crystals that generate an electrical signal directly when the γ ray deposits energy in the crystal. Scintillation detectors include NaI(Tl), bismuth germanate (BGO) and lutetium oxyorthosilicate (LSO); semiconductor detectors used in nuclear medicine imagers include cadmium zinc telluride (CZT). Radiation detectors are described in more detail in Chapter 6.

When a γ ray interacts in a scintillation crystal, it deposits some or all of its energy. This energy is re-emitted in the form of light with a wavelength dependent on the crystal material but not on the energy of the γ ray. The more energy deposited in the crystal, the greater the intensity of the light emitted. Scintillation crystals are coupled to photomultiplier tubes (PMTs), which serve

to convert the scintillation light into an electrical signal. If scintillation light strikes the photocathode of the PMT, electrons are emitted from the photocathode by the photoelectron effect. The number of photoelectrons emitted depends on the intensity of the scintillation light and, therefore, the energy deposited in the crystal. The energy required to produce a single photoelectron is ~1000 eV, so only a few hundred to a thousand electrons are produced for each γ ray that interacts, well below the number needed to produce a measurable current. The PMT contains approximately ten stages that serve to increase the number of electrons by secondary emission of electrons from these dynodes. The signal at the output of the PMT is a measurable current, the amplitude of which is still proportional to the energy deposited in the crystal.

Semiconductor detectors operate differently: the γ ray still deposits some or all of its energy in the crystal through photoelectric absorption or, more likely, Compton scattering interactions. However, that energy is not re-emitted as scintillation light; instead, it creates electron–hole (e–h) pairs that are then collected by application of an electric field to create a measurable signal. The energy required to create an e–h pair is ~3 eV, so many more charge carriers are created in semiconductor detectors than in scintillators (see Chapter 6).

While the exact implementations vary from system to system, the electronics of nuclear medicine imagers have several common functions: they determine the location of interaction of the γ ray in the detector, calculate the energy deposited in the crystal and ascertain whether that energy falls within a prescribed range of desirable energies, and for PET systems, measure the times that the two annihilation photons interacted and evaluate whether the difference in those times falls within a desired timing window to have both come from the same annihilation event (i.e. from the same positron decay). If an event is determined to be valid, its position (and sometimes the energy and timing information) is sent to the computer to be stored along with the information for the many other valid events.

In order to create an image of the distribution of radiotracer, the measured locations of interaction of the γ rays must be converted to a 2-D or 3-D map through image reconstruction. For 2-D planar imaging with a stationary gamma camera, this can be as simple as displaying the number of events at each detector position. For PET or SPECT imaging, where measurements are made for many views around the subject, the data must be combined through a reconstruction algorithm. These techniques range from analytical methods such as filtered back projection to iterative algorithms where estimates of the distribution are calculated and refined based on a model of the imaging system. Not all events accepted are actually useful events with accurate position, energy and timing information. To obtain quantitative images (i.e. images whose counts are directly

related to the amount of activity at each location), corrections must be applied for these unwanted events as part of the reconstruction process.

Performance measures aim to test one or more of the components, including both hardware and software, of a nuclear medicine imager.

8.1.2. Intrinsic performance

There are two general classes of measurements of scanner performance: intrinsic and extrinsic. Intrinsic measurements reflect the performance of a sub-part of the imager under ideal conditions. For example, measurements made on a gamma camera without a collimator will describe the best possible performance of the detector without the degrading effects of a collimator, although the collimator is essential for clinical imaging. For a PET scanner, intrinsic performance is often determined for a pair of detectors, rather than the entire system. Intrinsic measurements are useful because they reflect the best possible performance and can help isolate the source of any performance degradations observed clinically. However, these measures are typically performed under non-clinical conditions and will not reflect the performance of the nuclear medicine imager for patient studies. Intrinsic measures also tend to be measurements of an isolated characteristic of the system, rather than its impact on imaging studies. They reflect the limits of performance achievable by the detection system and electronics without collimators or image reconstruction.

8.1.3. Extrinsic performance

Extrinsic, or system, performance measures are made on the complete nuclear medicine imager under conditions that are more clinically realistic, although even these measures may not show the full clinical performance of the system. On a gamma camera, extrinsic measurements are made with the collimator in place; for SPECT and PET systems, the performance is often measured on the reconstructed image. The extrinsic performance of a system gives an indication of how well all of the components of the imager work together to yield the final image. As most extrinsic performance measurements attempt to isolate a single aspect of imaging performance (e.g. spatial resolution, count rate performance, sensitivity), the conditions of these measurements generally do not match the conditions encountered in patient imaging studies. However, the results of extrinsic performance measurements are generally good indicators of clinical performance or may provide useful information about system optimization for clinical studies.

8.2. ENERGY RESOLUTION

8.2.1. Energy spectrum

The amplitude of the signal from the detector depends on the energy deposited in the crystal. If the number of measured events with a given amplitude is plotted as a function of the amplitude (Fig. 8.1), the result is an energy spectrum. The shape of the energy spectrum depends on the radiotracer and γ rays emitted through its decay and the characteristics of the detector material, but all energy spectra have common features. There is one (or more than one) peak, called the photopeak, where the γ ray deposited all of its energy in the detector through one or more interactions. There is also a broad, lower energy region that reflects incomplete deposition of the γ ray's energy in the detector and/or Compton scattering of the γ rays in the body with the subsequent loss of energy before detection. Even in the absence of scattering material (i.e. for a point source in air), the photopeak is not a sharp peak but is blurred. This broadening, which depends on the properties of the detector, is due to statistical fluctuations in the detection of photons and conversion of the energy deposited in the crystal into an electrical signal. This effect is larger for scintillation detectors than for semiconductors. With scintillation detectors, there are several steps in the conversion process that are subject to statistical fluctuations, including the conversion of the γ ray's energy into scintillation light, collection of the scintillation light and conversion into photoelectrons at the PMT's photocathode, and multiplication of those photoelectrons at each dynode in the PMT. For semiconductor detectors, statistical uncertainty is introduced in the number of e–h pairs created when the γ ray deposits its energy and in the collection of these pairs.

The goal of nuclear medicine imaging is to map the distribution of radiotracers, so only γ rays that do not interact in the tissue before reaching the detectors are useful; any γ rays that scatter in the body first change their direction and do not provide an accurate measurement of the original radionuclide's location. Unscattered photons are those γ rays with energies in the photopeak. Nuclear medicine imagers accept events whose energies lie in a 'window' around the photopeak energy in order to reduce the contribution of lower energy, scattered γ rays. For PET scanners, a typical energy window is 440–650 keV for LSO detectors; for gamma cameras based on NaI(Tl) detectors, it is 15% of the photopeak energy (e.g. 129.5–150.5 keV for 140 keV γ rays from ^{99m}Tc, and 68–82 keV for the characteristic X rays from ^{201}Tl with a 20% window).

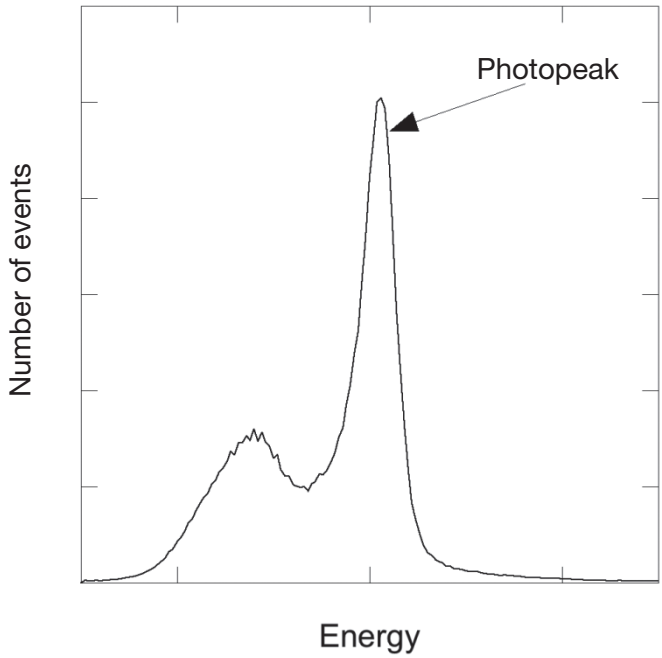

FIG. 8.1. An example of an energy spectrum, defined as the number of measured events with a given amplitude plotted as a function of the amplitude, where the amplitude depends directly on the energy deposited in the crystal.

8.2.2. Intrinsic measurement — energy resolution

The intrinsic ability of a detector to distinguish γ rays of different energies is reflected in its energy resolution. The energy resolution of a detector is defined as the full width of the photopeak at one half of its maximum amplitude, divided by the energy of the photopeak, and is typically expressed as a percentage. A smaller energy resolution value means that the detector is better able to distinguish between two γ rays whose energies are close to each other. The energy resolution depends on the energy of the γ ray approximately as $(\alpha + \beta E)^{1/2}/E$ and, therefore, the energy of the γ ray source must be specified when quoting the energy resolution of a system. The energy resolution worsens at lower energies because fewer photoelectrons are detected (in scintillation detectors) or e–h pairs are created (for semiconductors), so the statistical fluctuations in the measured signal are greater. In addition, the energy resolution of a complete imaging system is typically worse than that of small individual detectors due to slight differences in operating characteristics between detectors.

Only γ rays that have not scattered in the body will provide accurate information about the radiotracer distribution. Accordingly, the energy window is optimal if it includes as many photopeak events as possible, since they are more likely not to have interacted with the tissue, and as few lower energy events as possible, since they are more likely to be the result of one or more Compton scatter interactions in the tissue. As the energy resolution worsens, however, it is necessary to accept more low energy events because the photopeak includes lower energy γ rays. For example, for detection of 511 keV annihilation photons, the lower energy threshold for BGO (15–20% energy resolution) was typically set to 350–380 keV, while that for LSO (12% energy resolution) is 440–460 keV and for $LaBr_3$ (6–7% energy resolution) the lower energy threshold can be set as high as 480–490 keV without loss of unscattered γ rays.

8.2.3. Impact of energy resolution on extrinsic imager performance

The energy resolution is an intrinsic measure of detector performance; it defines the minimum width of the energy window for a given radiotracer. The energy window in turn affects the amount of scattered photons accepted. The ratio of scattered events to total measured events, the 'scatter fraction', is an extrinsic performance characteristic that is of concern, especially for quantitative imaging. In PET systems, for example, the clinical image is assumed to be linearly related to the activity uptake; because scatter adds a smoothly varying background to the image, it degrades the quantitative accuracy of the image and adds to the image noise, even when accurately estimated and subtracted.

There are two major types of scattered event, those where the initial γ ray scattered in the body and those where the γ ray was not completely absorbed in the detector but instead scattered, losing some but not all of its energy. In both cases, the measured energy of the γ ray is lower than the energy of the original photon because some energy is given up to the electron, and the measured position may no longer be related to the original source of the γ ray because the scattered photon does not travel along the same direction as the original γ ray. For typical patient sizes, scattering in the body is much more significant than detector scattering.

The scatter fraction is an extrinsic performance measure that describes the sensitivity of a nuclear medicine imager to scattered events. The measurement involves imaging a line source in a uniformly filled phantom of a specified size at a low activity level, where scattered and unscattered events can be reasonably well differentiated. As the amount of scatter depends on the size and distribution of scattering material in the scanner, the measured scatter fraction cannot be used to infer the amount or distribution of scatter in patient images. However, it is a good indicator of the relative sensitivity of the system to scatter.

The scatter fraction is directly related to the energy resolution of the system in the sense that the energy resolution determines the energy window, in particular the lower energy threshold. This determines the imager's ability to exclude scattered events. However, good energy resolution does not lead to a low scatter fraction unless the energy window used is made appropriately narrow; a scanner with 7% energy resolution will accept approximately as much scatter as one with 12% energy resolution if both systems have the same lower energy threshold. For this reason, measurement of the scatter fraction is a more clinically relevant parameter than the energy resolution.

8.3. SPATIAL RESOLUTION

8.3.1. Spatial resolution blurring

The spatial resolution of a nuclear medicine imager characterizes the system's ability to resolve spatially separated sources of radioactivity. An individual point source of activity does not appear at a single pixel in the image; rather, it is blurred over several pixels, largely due to statistical fluctuations in the detection of the γ rays. Sources whose measured activity distributions overlap cannot be distinguished as distinct sources and instead appear as a single, broad, low contrast source.

In addition to blurring small structures and edges, resolution losses also lead to a decrease in the contrast measure in these structures and at boundaries of the activity distribution. Activity in small structures is blurred into the background and vice versa. Areas of increased or decreased uptake are less easily detected because of this loss of contrast (the 'partial volume effect').

In imagers composed of many small crystals, the spatial resolution of the system is limited by the size of the detector elements. In gamma cameras with a single, large crystal coupled to an array of PMTs, the spatial sampling of the crystal determines the best spatial resolution achievable. The smaller the crystal element or the more finely sampled the detector, the better an event can be localized and the better the spatial resolution will be.

For a given size and sampling, crystals of different materials will have different spatial resolutions. This is because γ rays do not interact at the surface of a crystal but penetrate the crystal before interacting. If a crystal has a low density and low atomic number Z, γ rays will travel further before interacting, compared with a high density, high Z material. The ability to stop γ rays is referred to as the material's stopping power; detectors with higher stopping powers will have more accurate spatial localization than those with low stopping power because there is less inter-crystal scatter. The effect of stopping power becomes more

apparent when γ rays enter the crystal at an oblique angle to the face of the crystal (e.g. near the radial edge of a system comprising a ring of detectors). In that case, the γ rays can completely pass through the entrance crystal before interacting in a neighbouring crystal. The γ ray is then mis-positioned as though it had entered the neighbouring crystal or in some intermediate location, depending on the relative amounts of energy deposited by the two interactions.

Spatial resolution is also affected by the energy of the photon and, for scintillation detectors, the efficiency of collection of the scintillation light by the PMTs. The energy of the γ ray that is deposited in the crystal determines the amplitude of the measured signal, which in turn defines how accurately it can be localized in the detector. The spatial resolution measured in a given crystal with ^{99m}Tc (140 keV) is inferior compared to that which would be measured with a 511 keV photon.

As will be discussed later, the spatial resolution can also depend on the count rate or amount of activity in the scanner. As the count rate increases, there is an increased chance that two events will be detected at the same time in nearby locations in the detector. These events will pile up and appear as a single event at an intermediate location with a summed energy. This can lead to a loss of resolution with increasing activity.

8.3.2. General measures of spatial resolution

There are several ways to characterize the spatial resolution, whether of a detector or of a complete system. The point spread function (PSF) and line spread function (LSF) are the profiles of measured counts as a function of position across the point/line source. Rather than showing the complete profiles, however, it is more convenient to characterize them by simple measures. The full width at half maximum (FWHM) and full width at tenth maximum (FWTM) are useful to describe the widths of the profile although they do not give information about any asymmetry in the response. The equivalent width was defined as a way to combine the FWHM and FWTM into a single parameter and describe the shape of the profile in a simple way; it is defined as the width of a box function with a height equal to the maximum amplitude of the profile and an area equal to the total number of counts in the profile above 1/20 of its maximum amplitude. Reducing the PSF or LSF to a few parameters carries with it a loss of information about the spatial response of the imager; for example, LSFs or PSFs can have very different shapes and still have the same FWHM.

The modulation transfer function (MTF) is one way to more completely characterize the ability of a system to reproduce spatial frequencies. The MTF is calculated as the Fourier transform of the PSF and is a plot of the response of a system to different spatial frequencies. High spatial frequencies correspond to

fine detail and sharp edges, while low spatial frequencies correspond to coarse detail. The better the response at high frequencies, the smaller the structures that can be resolved. A flat response across all spatial frequencies means that the system most accurately reproduces the object. As it is difficult to compare imaging performance based on the MTF, however, the FWHM and FWTM are used to characterize spatial resolution.

8.3.3. Intrinsic measurement — spatial resolution

The intrinsic spatial resolution is a measure of the resolution at the detector level (or detector pair level for PET) without any collimation. It defines the best possible resolution of the system, since later steps in the imaging hardware degrade the resolution from the detector resolution. On gamma cameras, the intrinsic resolution is determined using a bar phantom with narrow slits of activity across the detector. On PET systems, the intrinsic resolution is measured as a source is moved between a pair of detectors operating in coincidence. The FWHM and FWTM of profiles of detected counts as a function of position are taken as measures of the intrinsic spatial resolution. In both cases, the intrinsic spatial resolution sets a limit on the resolution but does not translate easily into a clinically useful value because other components of the imager impact the resolution in the image.

8.3.4. Extrinsic measurement — spatial resolution

The spatial resolution of a nuclear medicine imager depends on many factors other than just the detectors. The linear and angular sampling play a significant role: to preserve the intrinsic resolution, the imager should be sampled every 0.1 × FWHM. Under-sampling leads to small structures being missed in the image. For single-photon imagers, a collimator is used to limit the direction of γ rays incident on the detector. Collimators are designed for specific purposes (e.g. sensitivity or resolution) and/or specific radionuclides. As the hole size and spacing of a collimator will affect the spatial sampling, each collimator will lead to different system spatial resolution.

The reconstruction processing performed to create tomographic images in SPECT or PET also affects the image resolution. Reconstruction algorithms are generally chosen to preserve as much fine detail and edge information as possible, while keeping image noise sufficiently low so that it is not confused with actual structure. The parameters of reconstruction can, therefore, change with the imaging study and with the number of events measured.

The spatial resolution is not constant throughout the imaging field of view (FOV). For PET systems, the resolution does not vary significantly with location

of the source between two detectors in a detector pair, but the system's radial resolution often degrades as the source is moved radially outwards from the centre of the scanner. For gamma cameras, the resolution degrades as the source is moved away from the detector face. For this reason, system spatial resolution measurements are performed with the source at different locations in the imaging FOV.

Extrinsic measures of spatial resolution are made under more clinically realistic conditions and include the effects of the collimator (for single photon imaging) and reconstruction processing. The extrinsic spatial resolution is typically measured with a small point or line source of activity of a sufficiently low amount such that effects seen at high count rates (i.e. mis-positioning of events) are negligible. Measurements of system spatial resolution can be performed in air or with scattering material added. A stationary source is positioned at specified locations throughout the nuclear medicine imager's FOV. The spatial resolution is determined from the images, including any reconstruction or processing steps, by drawing profiles through the source. No spatial smoothing or other post-processing is performed. In addition, any resolution modelling or resolution recovery techniques applied during clinical reconstruction are not used in the measurement of extrinsic resolution. The extrinsic spatial resolution is distinguished from the intrinsic resolution because it includes many effects not seen with the intrinsic resolution: collimator blurring, linear and angular sampling, reconstruction algorithm, spatial smoothing, and impact of electronics.

While the extrinsic resolution measurement reflects the resolution of the complete imaging system, the spatial resolution achieved in patient images is typically somewhat worse than the extrinsic spatial resolution. The spatial sampling is finer than occurs clinically because the pixel size is typically smaller than that used for patient studies in order to sample the PSF or LSF sufficiently. For imagers that reconstruct the data, the reconstruction algorithm in the performance measurement is often not the technique applied to clinical data; an analytical algorithm such as filtered back projection is generally specified for tomographic systems to standardize results between systems. Another key determinant of the clinical resolution is noise in the data that necessitates noise reduction through spatial averaging (smoothing), which blurs the image. For data with high statistics, a sharp reconstruction algorithm can be applied, and the resulting image has good spatial resolution. For more typical nuclear medicine studies, where the number of detected events is limited, some form of spatial smoothing is applied, with the resulting blurring of fine structures.

8.4. TEMPORAL RESOLUTION

8.4.1. Intrinsic measurement — temporal resolution

As the activity in the FOV increases, events arrive closer to each other in time until the imager cannot distinguish individual events. The timing resolution, or resolving time, is the time needed between successive interactions in the detector for the two events to be recorded separately. The timing resolution is largely limited by the decay time of the crystal. For scintillators, the decay time can be as high as 250–300 ns or as low as 20–40 ns, depending on the detector material. Typically, the scintillation light does not decay with a single time constant but with a combination of fast (nanosecond) and slow (microsecond) components. For semiconductor detectors, the decay time is much smaller. In addition to the detector decay time, the various components of the electronics can contribute to the loss of temporal resolution. The timing resolution is generally of less interest than its impact on the count rate performance of the system.

The timing of events is critical for PET, where two annihilation photons must be detected within a timing window to be recorded as a valid event. The timing window must be set wide enough to measure valid coincidence events but not so wide that many coincidences between uncorrelated annihilation photons (‘random coincidences’) are accepted. Coincidence timing electronics are carefully designed so that a detector’s signal is processed as quickly as possible, rather than waiting for the entire scintillation light to be measured. This allows the coincidence timing window to be set to <10 ns, limited by the time of flight of the two annihilation photons across the imager’s diameter. For a ring diameter of 90 cm, the minimum coincidence time window would be 6 ns.

Recent developments in PET technology allow for the difference in times of arrival (‘time of flight’) of the two annihilation photons to be measured. For time of flight systems, the coincidence time window is still 4–6 ns but the time of flight difference can be measured with a resolution of 300–600 ps. This time of flight information is used in reconstruction to localize the events. The timing resolution is measured with a low-activity source of activity by recording a histogram of the number of events as a function of time difference.

8.4.2. Dead time

The consequence of a finite timing resolution is a loss of counts measured at higher activities. When two photons arrive within the resolving time of the detector, the two photons are seen by the electronics as a single event. One or both of the events may be lost, and the events are also mis-positioned in space. The random nature of radioactive decay means that there is always a possibility that

two events will arrive within the resolving time of the detector; this possibility increases as the activity in the imager increases.

There are two kinds of dead time: non-paralysable and paralysable (see also Chapter 6). Non-paralysable dead time arises when an event causes the system to be unresponsive for a period of time, so that any later events that arrive during that time are not recorded. For paralysable dead time, the second event is not only not recorded but also extends the period for which the electronics are unresponsive. At moderate count rates, paralysable and non-paralysable dead times are the same; it is only at high count rates that the two types of dead time differ (see Fig. 8.2). It can be seen that systems with non-paralysable dead time saturate at high count rates, while those with paralysable dead time peak and then record fewer events as the activity increases. This leads to an ambiguity in the measured count rate: the same observed count rate corresponds to two different activity levels. The system dead time performance of nuclear medicine scanners is typically intermediate between paralysable and non-paralysable dead time because some components have paralysable dead time while other components have non-paralysable dead time.

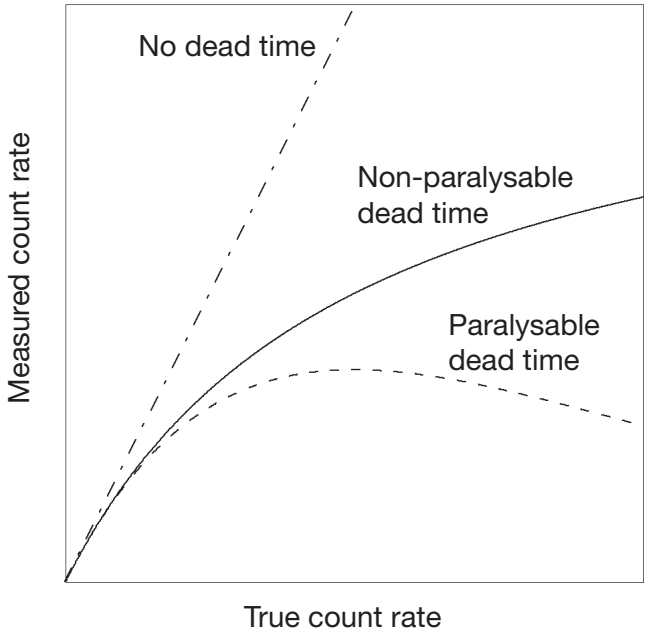

FIG. 8.2. System dead time as a function of count rate.

With increased dead time, additional activity injected in the patient does not lead to a comparable improvement in image quality or reduction in image noise. Dead time losses depend on the single event rate, coincidence count rate (for PET), and the analogue and digital design characteristics of the nuclear medicine imager. Dead time losses can depend on the activity distribution, especially for PET because of the different single photon and coincidence rate relationship with source distribution. They also depend on the radioisotope because dead time results from all γ rays that interact in the detector, not just the photons that fall within the energy window. For imaging studies with a large dynamic range (e.g. cardiac scans), count rate performance is critical.

To correct for event losses due to dead time, a correction based on a decaying source study is often applied to clinical data. The dead time correction will generally correct for the loss of counts, so that the number of counts in the image is independent of the count rate; it does not, however, compensate for the higher image noise that arises because fewer events are actually measured.

8.4.3. Count rate performance measures

The generic measurement of count rate performance involves determining the response of the nuclear medicine imager as a function of activity presented to the system. Typically, this requires starting with a high amount of activity and acquiring multiple images over time as the activity decays. The energy window is set at low activity levels and is not changed at higher activities to accommodate a shift in the photopeak due to pile-up effects. By comparing the observed events with the counts that would be expected after decay correction of events detected at low activities, the system dead time can be determined as a function of activity level. It is especially important to determine the maximum measurable count rate, since higher activities would result in no increase and perhaps a decrease in detected counts. While most count rate performance measures call for starting with a high activity and imaging as the activity decays, if too high an activity is used at the beginning of the measurement, the detector may show effects of saturation during later measurements at lower activities. Therefore, the amount of activity at the beginning of the study must be sufficient to measure the peak count rate but not be so high as to saturate the system for a significant period.

Intrinsic count rate performance measurements are performed with a source in air and without any detector collimation. This is typically performed only on gamma cameras. The system, or extrinsic, count rate performance is measured with the complete system, including any collimation or detector motion, and a distributed source with scattering material (e.g. a cylindrical phantom of specified dimensions or a source placed within scattering material). The scatter adds low

energy photons that contribute to pile-up and dead time that are not present in the intrinsic measurement.

For PET, random coincidences also increase as the activity increases; whereas the true coincidence rate would increase linearly with activity in the absence of dead time losses, the random coincidence rate increases quadratically with activity, so that their impact becomes greater at higher count rates. The activity where the random rate equals the true event rate is of importance, in addition to the activity and count rate at which the true count rate saturates or peaks. A global measure of the impact of random coincidences and scatter on image quality is given in the noise equivalent count rate (NECR) defined as:

$$\mathrm{NECR} = \frac{T^2}{T + S + kR} \tag{8.1}$$

where T, S and R are the true, scatter and random coincidence count rates, respectively, and k is a factor that is equal to one if a smooth estimate of random coincidences is used and two if a noisy estimate is used. This parameter does not include reconstruction effects or local image noise differences but can be useful in determining optimal activity ranges.

For systems that correct for dead time, it is important to apply dead time correction and to reconstruct the data in addition to looking at the count rates. The quantitative accuracy of the dead time correction is determined by looking at a large region of interest in decay-corrected, reconstructed images; the counts in the region of interest should be independent of activity level. It is also important to examine the images at high activities for artefacts that may arise due to spatially-varying mis-positioning effects or inaccuracies in various corrections with increased activity.

8.5. SENSITIVITY

8.5.1. Image noise and sensitivity

Images from nuclear medicine devices are typically noisy because the amount of activity that can be safely injected and/or the scan duration without patient discomfort or physiological changes in activity distribution is limited. The number of detected events for a given amount of activity in the imaging system's FOV is an important performance characteristic because a more efficient imager can achieve low image noise with lower injected activity than a less efficient system. Noise in the image can affect both visual (qualitative) image quality

and quantitative accuracy, especially in areas of low uptake or low contrast. The relative response of a system to a given amount of activity is reflected in its sensitivity.

The sensitivity of a system is determined by many factors. The geometry of the imager, especially the solid angle of the detectors, as well as any collimation will determine how many photons reach the detectors. The stopping power and depth of the detectors will impact how many of these photons are detected. In addition, the radionuclide's energy, coupled with the imager's energy resolution and energy window, affect the number of accepted events. Finally, the number of counts measured in a given time for a fixed amount of activity depends on the source distribution and its position in the imager.

8.5.2. Extrinsic measure — sensitivity

All performance measurements of sensitivity are extrinsic; for single photon imaging, in particular, the collimator is a major source of loss of events, so it is more clinically interesting to know the sensitivity of the system with a particular collimator.

As noted above, the number of observed counts depends greatly on the activity distribution. For this reason, any measurement of sensitivity is performed under prescribed conditions that do not attempt to replicate patient activity distributions. The source configurations and definitions of sensitivity vary widely, however. For planar imaging, a shallow dish source without intervening scatter material is used, and the sensitivity is reported as a count rate per activity. For SPECT, a cylindrical phantom is filled uniformly with a known activity concentration, and the sensitivity is reported as a count rate per activity concentration. For whole body PET scanners, a line source that extends through the axial FOV is imaged with sequentially thicker sleeves of absorbing material, and the data are extrapolated to the count rate one would measure without any absorber; the sensitivity is then reported as a count rate per unit activity. Small animal PET systems use a point source in air centred in the scanner, and the count rate per activity, as well as the absolute sensitivity (in per cent) are reported. None of these sensitivity measurements can be used to predict the number of events that will be observed for patient studies; however, systems with higher sensitivity will generally record more events from a patient activity distribution than those with lower sensitivity.

8.6. IMAGE QUALITY

8.6.1. Image uniformity

The uniformity of response of a nuclear medicine imager across the FOV is important for both qualitative and quantitative image quality. All PMTs of a given type do not respond exactly the same way, and a correction for this difference in gain is applied before the image is formed. Collimators can also have defects that lead to non-uniformities in the image. For tomographic scanners, corrections for attenuation and unwanted events such as scatter can also affect the uniformity of the image.

Intrinsic uniformity is measured without a collimator by exposing the detector to a uniform activity distribution (e.g. from a distant, uncollimated point source). Intrinsic uniformity is measured at both low and high count rates, where mis-positioning effects become more pronounced. The extrinsic system uniformity is determined with a collimator in place (for single photon imaging), and images are processed or reconstructed as for clinical studies. In both cases, sufficient counts must be detected, so that image noise is low. Quantitative assessment of image uniformity includes variation of pixel counts in small regions across the FOV. However, because simple metrics of non-uniformity such as this do not provide a complete assessment of what the eye perceives in the image, a visual analysis is also important.

8.6.2. Resolution/noise trade-off

Most performance measurements are carried out under non-clinical conditions to isolate an aspect of the imager's performance. To include more of the effects seen in clinical data, some performance standards call for a measurement of image quality. The activity distribution is a series of small structures (e.g. spheres of varying diameters) in a background activity typical of the activity levels seen in patient studies. The activity is imaged for a clinically relevant time, so that the noise level in the data is comparable to that in typical patient studies. The data are processed in the same manner as clinical data. The resulting image, then, is a better representation of the resolution and noise seen clinically. Data analysis consists of such measures as sphere to background contrast recovery, noise in background areas and/or signal to noise ratio in the spheres. While still a simplistic and non-clinical distribution, the measurement gives a more relevant indication of clinical resolution/noise performance.

8.7. OTHER PERFORMANCE MEASURES

There are many other performance measures that reflect a given aspect of a nuclear medicine imager. For planar systems, the spatial linearity, or spatial distortion of the measured position of photons compared to the actual position, is important for good image quality. A number of nuclear medicine imaging systems incorporate anatomical (e.g. computed tomography or magnetic resonance imaging) imagers into the scanner, and the images from the different modalities must be registered spatially. Another area where spatial registration is necessary is in single photon systems where multiple energy windows are used, and the images acquired in the different windows must be overlaid to form the image. Quantitative linearity and calibration is an important measurement for systems such as PET scanners that aim to relate pixel values to activity concentrations.