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Problems in detection and measurement in nuclear medicine

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Abstract. Nuclear Medicine studies are performed with a variety of types of radiation measurement instruments, depending on the kind of radiation source that is being measured and the type of information sought. For example, some instruments are designed for in vitro measurements on blood samples, urine specimens, and so forth. Others are designed for in vivo measurements of radioactivity in patients. All these instruments have special design characteristics to optimize them for their specific tasks, as described in this study; however, some considerations of design characteristics and performance limitations are common to all of them. An important consideration for any radiation measurement instrument is its detection efficiency. Maximum detection efficiency is desirable because one thus obtains maximum information with a minimum amount of radioactivity. Also important are instrument's counting rate limitations. There are finite counting rate limits for all counting and imaging instruments used in nuclear medicine, above which accurate results are obtained because of data losses and other data distortions. Non penetrating radiations, such as ß particles, have special detection and measurement problems. In this study, some of these general considerations have been discussed.

1 Introduction

Radioisotopes have made their unique contributions to medicine because it is possible to detect the disintegration of individual nuclei and hence to locate submicroscopic quantities of a given material in body tissues or fluids. The physical amount of radioactive tracer required to follow, for example, a metabolic process is so small that it does not alter the process itself [1].

The extreme sensitivity of radiation detection equipment is a cardinal factor in the tracer procedures of nuclear medicine. Hence, the choice and use of nuclear instrumentation plays a vital part in the value and accuracy of the results obtained in radioisotope tests [1].

2 Materials and Methods

Nuclear medicine studies are performed with a variety of radiation measurement instruments, depending on the kind of radiation source that is being measured and the type of information sought. The practice of in vivo counting now frequently faces the problem of the detection and the quantitative assessment of low energy photon emitting radionuclides in the body.

In general, it is desirable to have as large a detection efficiency as possible, so that a maximum counting rate can be obtained from a minimum amount of activity. Detection efficiency is affected by several factors, including the following:

a) The geometric efficiency, which is the efficiency with which the detector intercepts radiation emitted from the source. This is determined mostly by detector size and the distance from the source to the detector.

b) The intrinsic efficiency of the detector, which refers to the efficiency with which the detector absorbs incident radiation events and converts them into potentially usable detector output signals. This is primarily a function of detector thickness and composition and of the type and energy of the radiation to be detected.

c) The fraction of output signals produced by the detector that are recorded by the counting system. This is an important factor in energy-selective counting, in which a pulse-height analyzer is used to select for counting only those detector output signals within a desired amplitude (energy) range.

d) Absorption and scatter of radiation within the source itself, or by material between the source and the radiation detector. This is especially important for in vivo studies, in which the source activity generally is at some depth within the patient [2].

It has been shown in a previous study [3] that currently used concepts such as efficiency and background have to be employed in a more precise way when the measurement of low energy photons is considered. The counting efficiency associated with a counting geometry and the point efficiency is preferably used instead of intrinsic efficiency, which does not consider the anisotropy of the detectors in use.

Too often, the efficiency is considered as the key parameter to describe a counting device so that the detector with the highest volume is often selected in several applications. This is not correct in lower energy photon

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range: a volume increase or an increased number of detectors in an array always leads to an increase of the continuum. This is proportional to the total volume of the detector, but the efficiency is only increased when the added volume is effectively active in detection of the photons exiting the body. If the added volume only increases the continuum, the detection limits are increased. In other words, a detection volume that does not collect the examined photons, increases the continuum without bringing information from the investigated sources. It is better to look for a low detection limit in the examined region than for a high efficiency in full energy range. This means that the detector's size can advantageously be tailored in accordance with the applications [4].

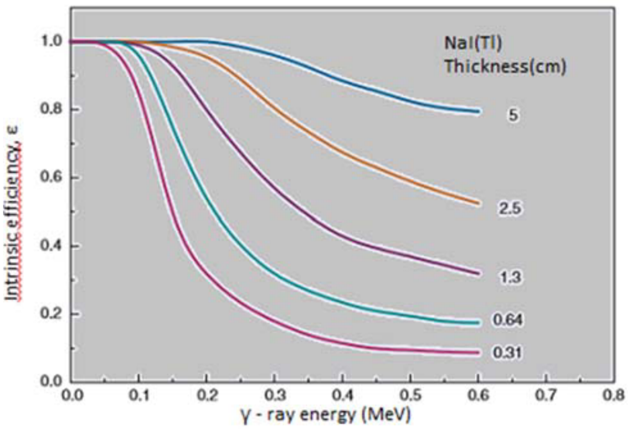

Figure 1. Intrinsic efficiency versus γ-ray energy for NaI(Tl) detectors of different thicknesses.

The size of the detector can be tailored to each application in order to improve the response in terms of the ratio "Efficiency/ detection limit": the thickness can be adapted to the energy range in order to decrease the detection limits (Fig. 1). Extra thickness increases the continuum without bringing information concerning the examined photopeak. A detector thickness can be considered adapted to the energy when 50 to 90% of the incident photons are absorbed in the detector. The same effect can be considered when the diameter is increased in the measurement of a local deposition. The use of small detector arrays instead of a large single one will provide an opportunity to examine the spectrum from each detector separately, to localize the deposition, to correct calibration factor according to the burden geometry and to repeat the burden calculation procedure with a better detection limit. (Fig. 2)

Figure 2. Examples of detector profiles with different complications for the computations of total detection efficiency.

Radiation striking the sensitive portion of a detector may be wholly or partially absorbed or it may pass right through the detector. If a Geiger tube is used, the signal from the detector is independent of the energy of the radiation, so that partial or total absorption produce the same results. In a scintillation detector the size of the detector output pulse is a function of the energy lost by the gamma ray in the crystal. Therefore, for scintillation detectors it is necessary to know both the total intrinsic efficiency (that is, the fraction of the incident rays detected) and the efficiency for total absorption, called the photofraction (Fig. 3).

For a scintillation detector using a sodium iodide crystal, both the total and the photofraction efficiency are determined by the energy of the radiation and the size of the crystal [1].

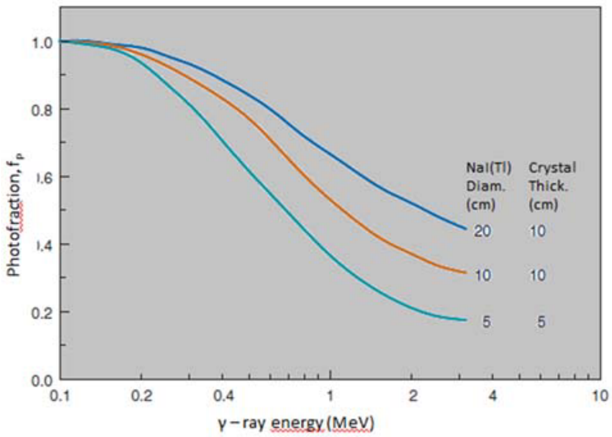

Figure 3. Photofraction versus γ-ray energy for cylindrical NaI(Tl) detectors of different sizes.

Images of the distribution of radioactive material in organs of patients are formed by looking at many small areas, either one after the other (scanning), or simultaneously (a camera technique). The quality, and hence the diagnostic value of the resulting image or scan depends to a large extent on the size and sharpness of outline of each of the areas or picture elements and on the statistical accuracy of the data obtained from each picture element. Obviously, the spatial definition, often called resolution, is improved if the size of each picture element is reduced, just as a printed picture looks better if a finer screen is used on the printing plate. Making the picture element smaller means that fewer counts will be recorded for that area in a given time unless the dose to the patient is increased. Thus, a compromise must be reached between spatial definition and the statistical validity of the information.

Determining the absolute efficiency (percentage of disintegrations detected) of a radiation detector in a given physical setup is a complex problem because of the many factors of geometry and detector performance that must be considered. For this reason, most radiation measurements are relative rather than absolute; that is, they involve obtaining count-rate data on an unknown sample (or patient) and on a standard of the same radioisotope under as nearly identical conditions as possible [2]. For example, a dose is measured before administration to a patient, a subsequent blood sample is counted later at the same position relative to the detector, and the ratio of the two net count- rates represent the clinical data. In order to be able to make both the standard and the unknown measurements as similar as possible, one must understand and control all factors affecting the counting efficiency.

The first factor that must be considered is the portion of the radiation produced in the sample or organ which interacts with other atoms in the source and never gets outside. This process is called self- absorption. Obviously, the standard and the unknown should have similar self-absorption properties.

The attenuation by the tissues is indeed so important that the measurement is affected by a serious lack of accuracy; this can vary between 100 and 700% [1]. A heterogeneous deposition can be difficult to quantify or can even be undetectable [2,3]. This difficulty leads to reconsideration of the counting strategy and also to defining precisely the concepts that are used. This is necessary if radionuclides such as ^{125}I, ^{241}Am and ^{67}Ga are to be assessed in vivo with a reasonable sensitivity.

Background radiation is present at all times in all places. It comes in part from cosmic radiation and in part from naturally occurring radioactive material incorporated in the building. For example, granite contains detectable amounts of uranium daughter products. In addition, background radiation may also come from nearby sources of radioactive material such as a cobalt-60 therapy installation, a radium safe, a supply of therapeutic or multiple diagnostic doses of radiopharmaceuticals, or patients who have received radioisotope therapy doses [1].

For this reason the detector should be well shielded on all sides except the one facing the patient or the sample being measured. For clinical work the shielding should be at least the equivalent of 3/4 inch of lead for 1-inch diameter scintillation detectors and should go up from there to 2 inches of lead or more for 3-inch diameter scintillation detectors. It is very important that this shielding against the general background extend to the back of the sensitive portion of the detector, since background radiation can come from all directions and can be scattered back even if one principal component, such as cosmic radiation, comes mainly from one direction [1].

The background count-rate of an unshielded scintillation detector varies with the volume of the crystal while its counting efficiency of radiation of a given energy coming from a point source varies with the frontal area of the crystal. Thus, there is no point in choosing a crystal which is thicker than required for the almost total absorption of the radiation being measured. A 1-inch thick crystal is adequate for measuring ^{131}I if all detected rays are counted, while a 2-inch thick crystal are used (2-inch diameter or more) better rations of efficiency to limit the counted rays to those falling in the photopeak (total absorption) [1].

Detectors must be shielded not only against background radiation, but also against radiation coming from parts of the patient's body other than the one under study at the moment. If a whole organ, such as a thyroid or kidney is being measured, then the front opening should be conical and subtend a solid angle just large enough to enable the detector requires a 36°collimator to "see" a large thyroid at a distance of 20cm. At a distance of 35 cm the same detector needs a 20° collimator. This type of collimator is called a flat field collimator because it has rather uniform sensitivity across its opening.

The background is an important parameter: it depends on the energy range of the measured photons. In the measurement of low energy photon emitters, consideration of background is crucial because any shielding produces, by fluorescence, a shift of the continuum towards the low energy region. For this reason, shielding can be avoided and the counting can be carried out in many places without shielding when the background is first controlled. This technique allows much longer counting periods, but depends on the required detection limits and on the examined energies. The environment has an effect on the continuum and then on the detection limits of a counting system. If the source is covered by inactive material, as for example the thyroid is covered by neck tissue, then absorbed in the covering material must also be taken into account. It is possible that a gamma ray may be only partially absorbed in the source or its covering matters and that it may emerge with reduced energy and a change in direction as a scattered ray. For this reason, the standard must be arranged to have scattering conditions similar to those of the unknown, or all scattered radiation must be eliminated from the measurement by using an energy discriminator.

Of all of the radiation emerging from the source only a fraction will be directed toward the detector. This fraction is determined by the solid angle subtended by the detector with respect to the source. It is ruled by the “inverse square law,” since doubling the distance between a point source and detector reduces the solid angle by a factor of 4[1].

Not all of the radiation within the subtended solid angle reaches the sensitive portion of the detector because of absorption in the air and in the detector cover. This is important primarily in the case of beta and low energy gamma and x radiation.

Because of their relatively short ranges in solid materials, beta particles create special detection and measurement problems. These problems are especially severe with low-energy beta particle emitters, such as ^{3}H and ^{14}C. The preferred method for assay of these radionuclides is by liquid scintillation counting techniques; however, these techniques are not applicable in all situations, such as when surveying a bench top with a survey meter to detect ^{14}C contamination.

A survey meter can be used to detect surface contamination by beta particle emitters provided it has an entrance window sufficiently thin to permit the beta particles to enter the sensitive volume of the detector. Efficient detection of low energy beta emitters requires a very thin entrance window, preferably fabricated from a low-density material. A typical entrance window for a survey meter designed for ^{3}H and ^{14}C detection is 0.03 mm thick Mylar (~1.3 mg/cm^2 thick). Mica and beryllium also are used. Such thin windows are very fragile, and usually they are protected by an overlying wire screen. Beta particles that are more energetic (e.g., from ^{32}P) can be detected with much thicker and more rugged entrance windows; for example, 0.2 mm-thick aluminum (~50mg/cm^2) provides approximately 50% detection efficiency for ^{32}P.

GM and proportional counters sometimes are used to assay the activities of beta emitting radionuclides in small trays (planchets) or similar sample holders. Two serious problems arising in these measurements are self-absorption and backscattering. Self-absorption depends on the sample thickness and the beta particle energy. For ^{14}C and similar low energy beta emitters, self-absorption in a sample thickness of only a few mg/cm^2 is sufficient to cause a significant reduction of counting rate. Backscattering of beta particles from the sample and sample holder tends to increase the sample counting rate and can amount to 20% to 30% of the total sample counting rate in some circumstances. Accurate assay of beta emitting radioactive samples by external particle counting techniques requires careful attention to sample preparation. If only relative counting rates are important, then it is necessary to have sample volumes and sample holders as nearly identical as possible.

Bremsstrahlung counting can be employed as an indirect method for detecting beta particles using detectors that normally are sensitive only to more penetrating radiations such as x-rays and gamma rays. Bremsstrahlung counting also was employed in some early studies using ^{32}P for the detection of brain tumors and still used occasionally to map the distribution of ^{32}P labeled materials administered for therapeutic purposes. Bremsstrahlung counting is effective only for relatively energetic beta particles and requires perhaps 1000 times greater activity than a gamma ray emitter because of the very low efficiency of bremsstrahlung production.

Detection efficiencies can be determined experimentally using calibration sources. A calibration source is one for which the activity or emission rate is known accurately. This determination is made by the commercial supplier of the source.

3 Conclusions

Radiation measurement systems are subject to various types of malfunctions that can lead to sudden or gradual changes in their performance characteristics. For example, electronic components and detectors can fail or experience a progressive deterioration of function, leading to changes in detection efficiency, increased background, and so forth. To ensure consistently accurate results, quality assurance procedures should be employed on a regular basis for all radiation measurement systems. These would include (1) daily measurement of the system's response to a standard radiation source (e.g., a calibration "rod standard" for a well counter or a "check source" for a survey meter) (2) daily measurement of background levels; and (3) for systems with pulse-height analysis capabilities, a periodic (e.g., monthly) measurement of system energy resolution.

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