GUIDELINES FOR THE CALIBRATION OF LOW ENERGY PHOTON SOURCES AND BETA-RAY BRACHYTHERAPY SOURCES[1]

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1 INTRODUCTION

With the development of improved methods of implanting brachytherapy sources in a precise manner for treating prostate cancer and other disease processes, there has been a tremendous growth in the use of low energy photon sources, such as ^{125}I and ^{103}Pd brachytherapy seeds. Low energy photon sources have the advantage of easier shielding and also lowering the dose to normal tissue. However, the dose distributions around these sources are affected by the details in construction of the source and its encapsulation more than other sources used for brachytherapy treatments, such as ^{192}Ir. With increasing number of new low energy photon sources on the market, care should be taken with regard to its traceability to primary standards. It cannot be assumed, that a calibration factor for an ionization chamber, that is valid for one type of low energy photon source automatically is valid for another source even if both would use the same isotope. Moreover, the method used to calculate the dose must also take into account the structure of the source and the encapsulation. The dose calculation algorithm that is valid for one type of low energy source may not be valid for another source even if in both cases the same radionuclide is used. Simple "point source" approximations, i.e. where the source is modeled as a point, should be avoided, as such methods do not account for any details in the source construction.

In this document, the dose calculation formalism adopted for low energy photon sources is that recommended by the American Association of Physicists in Medicine (AAPM) as outlined by Task Group-43 (TG-43) [1]. This method accounts for the source and capsule geometry. The AAPM recommends brachytherapy photon sources to be specified in terms of 'Air Kerma Strength' that is also used in the formalism mentioned above. On the other hand, the International Commission on Radiation Units and Measurements (ICRU) recommends that the specification be done in terms of Reference Air Kerma Rate [2,3]. In this document the latter recommendation is adopted. Both of the quantities give the same numerical value for the source strength and differ only in the units they are expressed[2]. The recommended dose calculation method is discussed further in the text.

Sealed beta-ray sources for brachytherapy treatments have been in use for few decades already. An example is application of ^{90}Sr/^{90}Y planar sources for ophthalmic brachytherapy treatments. For these types of treatments, a precise dose distribution within the eye globe is needed. Modern diagnostic techniques permit the determination of all volumes of interest in the eye, i.e. tumor and critical structures such as optic disc and iris with a high precision. It is therefore of importance to optimize the treatment by limiting the dose to these critical structures.

A relatively new and rapidly developing field in brachytherapy is the use of beta-ray sources for prevention of restenosis, i.e. re-closing of artery, following coronary and peripheral artery interventional procedures such as angioplasty, atherectomy and stent implantation.

The dosimetry of beta-ray sources for therapeutic applications is particularly difficult due to the short distances involved, being at the millimeter range, and the high dose gradients at such short distances. Further difficulties are caused by the non-uniform distribution of

[1] This follows a Consultants' Meeting held during 1-5 November 1999 at the IAEA in Vienna. The scientific secretary of the meeting was Mr. H. T. Tölli.

[2] The units of Air Kerma Strength and the Reference Air Kerma Rate are m^2Gy/h and Gy/h, respectively.

activity in the source itself, causing a highly irregular dose distribution.
The aim of this report is to recommend methods for calibration of low energy photon sources and beta-ray sources used in brachytherapy treatments and to propose suitable detectors for this purpose. Dose calculation methods are given both for the photon sources and beta-ray sources covered in this report.

The present report has been developed in close collaboration with the ICRU Report Committee on this subject. The ICRU is planning to publish a report on the calibration of the type of sources discussed here. The present report is to a large extent based on that report [4].

2 SPECIFICATION OF BRACHYTHERAPY SOURCES

The following discussion is limited to the specification of brachytherapy sources from the point of view of what is needed to achieve traceability of calibrations. There are also other necessary and useful specifications for using the sources in clinical applications.

2.1 LOW ENERGY PHOTON SOURCES

The recommended quantity for specification of brachytherapy gamma sources is the reference air kerma rate, K_R, defined by the ICRU [2,3] as the kerma rate to air, in air, at a reference distance of one meter, corrected for air attenuation and scattering.

2.2 BETA-RAY SOURCES

The recommended quantity for specification of beta-ray sources is the reference absorbed dose rate in water at a reference distance from the source. The reference distance differs from one type of source to another. For planar and concave sources, the reference distance is 1 mm from the centre of the source, whereas for seeds and line sources it is 2 mm in the transverse direction from the source's longitudinal axis. For balloon, shell and stent sources the reference distance is 0.5 mm measured from the surface of the source.
It must be recognized that measurements at these short distances are a difficult task. The distances are chosen from the point of view of the low penetration of the beta-rays and the relevance to clinical applications.
The contained activity can be used as a supplementary specification for beta-ray seed, wire, balloon, shell and stent sources.

2.3 OTHER IMPORTANT QUANTITIES

Whereas the reference air kerma rate and the reference absorbed dose rate are sufficient to yield traceability in the source calibration, it is of importance that other parameters are specified as well. To be able to make use of the published theoretical spectral information of brachytherapy sources, a useful specification is the purity of the source, i.e. a statement on the maximum percent amount of any contaminants in the source. The following sections give some quantities for beta-ray sources that are of importance in clinical applications.

2.3.1 Beta-ray plaque sources

2.3.1.1 Depth dose

As a further specification, the relative central axis depth dose curve in water should be given, preferably in numeric form, for each type of source.

2.3.1.2 Source uniformity

Source uniformity is specified as the uniformity of the absorbed dose rate measured at a depth of 1 mm in a water-equivalent medium. A map of uniformity or few dose profiles across the source should be available as part of source specification.

Source uniformity of plaque sources can be quantified by a parameter, which is equal to the percentage difference of the maximum and minimum values of relative absorbed dose rate over a specified area of the source[3]. The value of this parameter should not exceed 20 %.

2.3.2 Beta-ray seed and wire sources

2.3.2.1 Contained activity

The importance of contained activity as a source specifier is in comparison between model predictions and dosimetry

[3] A more precise definition is given in the ICRU Report [4].

measurements. Monte Carlo calculations predict dose per history, where a history represents the interactions undergone by a single emitted photon or electron. The number of histories can be related to contained activity by disintegration probabilities and branching ratios for complicated decay structures. Thus it can be said that Monte Carlo models predict dose rate per unit contained activity. When one wants to compare the predictions of a model to dose rate measurements with a particular source, one can only do so with knowledge of the contained activity for the source in question.
Contained activity for a beta-ray source can best be determined from a destructive measurement, which involves dissolving a source in a liquid medium that captures all of the contained activity into an aqueous solution [5]. By a suitable dilution of this solution, contained activity can be determined with a high degree of accuracy (1 to 2% at 1 σ) by the liquid scintillation technique.

A contained activity calibration of a seed or wire beta-ray source can be transferred to a well-type ionization chamber resulting in a method to specify such sources in terms of contained activity rather than reference absorbed dose rate. The preferred use of this quantity is, however, to convert contained activity to reference absorbed dose rate using well-established reference absorbed dose rate per unit activity constants for particular source types. These constants are obtained using a combination of Monte Carlo calculations and careful absorbed dose rate measurements.

2.3.2.2 Source uniformity

It has been recommended [6] that the uniformity of seed and line sources be evaluated in terms of absorbed dose rate at a distance of 2 mm from the source center both longitudinally and perpendicular to the source axis (equatorial) in a tissue-equivalent medium. For longitudinal uniformity it is recommended that over the central 2/3 of the active length of the source a deviation from maximum to minimum dose rate be no greater than 20% relative to the average dose rate over this length. Equatorial deviations should be no greater than 20 % relative to the average over all angles.

2.3.3 Beta-ray liquid- or gas filled balloons, shell and stent sources

For these sources the recommended calibration quantity is the reference absorbed dose rate measured at a distance of 0.5 mm from the source surface. For stent sources, which are highly non-uniform even at this depth, there is a lack of guidance as to whether the maximum or average dose rate is the quantity of interest. Particularly for stents and volume sources, the quantity contained activity (see above) takes on increased importance and may become the preferred quantity for source specification. Since the absorbed dose rates from stents are so low, there are practical difficulties with absorbed dose rate measurements with all but the most sensitive detector systems.

2.3.4 Obsolete quantities for photon sources

Quantities such as equivalent mass of radium and apparent activity, A_{app}, are considered obsolete and are not recommended for the specification of brachytherapy photon sources. However, these quantities are widely used in the brachytherapy community. In particular, A_{app} is often used by vendors for source strength specification. It is also frequently employed in older brachytherapy treatment planning systems. In such cases, when a conversion from one quantity to another is necessary, a consistent set of conversion factors must be used [7].

A_{app} is defined as a quantity that is mathematically derived from the reference air kerma that is traceable to the appropriate standard. It cannot be experimentally determined independently of reference air kerma rate [8]. The apparent activity is related to the reference air kerma rate by

$$A_{app} = \frac{r_{ref}^2 K_R}{(\Gamma_\delta)_K} \qquad (1)$$

where $(\Gamma_\delta)_K$ is the air kerma rate constant and r_{ref} is the reference distance of one meter. The value of the air kerma rate constant depends on the construction of the source and its encapsulation as well as the photon energy.

The problem in the use of A_{app} is apparent from the above equation. Different values of

$(\Gamma_\delta)_K$ will give different apparent activities. For many brachytherapy sources, a number of air kerma rate constants have been published. Failure to uniformly define and apply $(\Gamma_\delta)_K$ could cause significant confusion and unnecessary treatment delivery errors. The apparent activity is not the contained activity and will differ depending on the construction of the source. The use of A_{app} should cease as soon as possible.

Table I give a summary of the recommended quantities for specification of brachytherapy sources discussed in this report. Included are also recommended working standards for calibration.

TABLE I. SPECIFICATION OF BRACHYTHERAPY SOURCES AND THE RECOMMENDED WORKING STANDARDS AT SSDLs AND HOSPITALS FOR CALIBRATION

Source type	Primary quantity	Distance specified	Measured from	Supplementary quantity	Working standard
Photon seed and line	Reference air kerma rate	1 m	Source	None	Well type ionization chamber
Beta plane and concave	Reference absorbed dose rate	1 mm	Surface	None	Calibrated source
Beta seed and line	Reference absorbed dose rate	2 mm	Centre	Contained activity	Well type ionization chamber
Beta balloon, shell & stent	Reference absorbed dose rate	0.5 mm	Surface	Contained activity	Well type ionization chamber

3. SOURCE DATA

3.1 PHOTON SOURCES

Some data for low energy photon sources used in brachytherapy applications are given in this section. More extensive description, including constructional details and the type of clinical application are given in the forthcoming ICRU Report [4], upon which the present report is based.

The half-lives of ^{125}I and ^{103}Pd are given in Table II.

TABLE II. HALF-LIVES OF LOW ENERGY PHOTON SOURCES DISCUSSED IN THIS REPORT.

Isotope	Half-life (Days)
^{125}I	59.41
^{103}Pd	16.99

In Tables III and IV are shown some ^{125}I and ^{103}Pd sources with a calibration available at a Primary Standards Dosimetry Laboratory (PSDL). Currently, the only PSDL that have established primary standards for the low energy photon sources is the National Institute of Standards and Technology (NIST), USA.

TABLE III. SOME ^{125}I SOURCES WITH A PSDL CALIBRATION AVAILABLE.

Manufacturer	Model(s)
Nycomed Amersham	6711, 6702
North American Scientific / Mentor	MED3631-A/M
International Isotopes Inc.	IS-12500, IS-12501
Bebig / Uromed	Symmetra
Best Industries	2301
Mills Biopharmaceuticals, Inc.	125SH
Syncor Pharmaceuticals, Inc.	Pharmaseed BT-125-1

TABLE IV. SOME ^{103}Pd SOURCES WITH A PSDL CALIBRATION AVAILABLE

Manufacturer	Model(s)
Theragenics	Theraseed 200
North American Scientific / Mentor	MED3633
Best Industries	2335
International Brachytherapy	1031L

The dosimetric characteristics of low energy sources, such as ^{125}I and ^{103}Pd, are very sensitive to the details of encapsulation geometry and source internal structure due to self-absorption and filtration effects. Significant dosimetric differences between different seed models containing the same radionuclide may result from relatively minor differences in design specifications or in manufacturing processes. It is therefore important to individually evaluate the dosimetric characteristics of each new low energy (less than 50 keV), photon-emitting brachytherapy source product.

In the Symmetra source the activity of ^{125}I is distributed on a ceramic rod. The rod is coated with gold for enhanced visibility on radiographs. On the other hand, in the MED3631-A/M source, the two radiographic markers are made of silver and copper. With respect to the low energy emitted by the sources, and the differences in the source constructions, it is easy to understand that the average photon energy emitted differ from one source to another. Clearly, it is not possible to use a common air kerma rate constant, $(\Gamma_\delta)_K$, for determination of the apparent activity. Figure 1 shows an example of measured photon energy spectra from three different ^{125}I sources. Note the presence of the Ag-peaks for some sources.

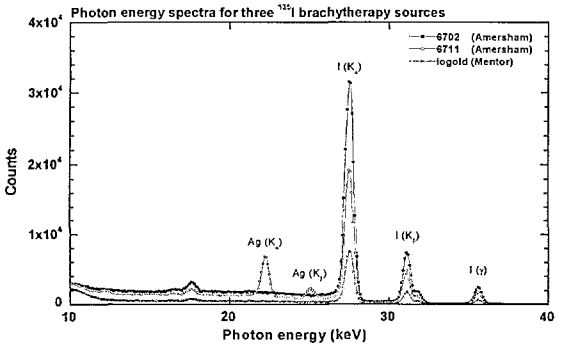

FIGURE 1. Photon energy spectra of three different ^{125}I sources measured at 40 cm distance with a high purity HPGe detector.

Presently there are a number of new sources appearing on the market. The procedure prior the clinical use of these sources has been outlined by Williamson [9]. It is inappropriate to apply the different constants and functions (i.e. dose-rate constants, radial dose functions, anisotropy functions, anisotropy factors, geometry functions)[4] published in the TG-43 Report [1] for currently available ^{125}I (Amersham models 6711 and 6702) and ^{103}Pd (Theraseed 200) interstitial sources to other low energy seed products. Prior to approval of new sealed brachytherapy sources a PSDL calibration should be obtained and dosimetric measurements need to be made and published [9]. The dosimetric characteristics of each new product should be evaluated. At least one and preferably two experimental studies of the dose distribution using an appropriate phantom should be completed. At least one study must include absolute dose-rate measurements, and, in addition, a Monte Carlo simulation by an independent investigator should be made which includes calculation of the dose rate constant, i.e. the dose at a distance of 1 cm per unit Reference Air Kerma Rate. These dosimetric studies should be compared with each other and relevant data from the literature. Taken together, the dose measurements and Monte Carlo calculations should encompass a sufficient range of distances and polar angles that dose-rate constants, radial dose functions, anisotropy functions, anisotropy factors and anisotropy constants can be unambiguously estimated. In addition, a rigorous system of verifying constancy and accuracy of the vendor's source calibration should be maintained.

3.2 BETA-RAY PLAQUE SOURCES

Physical data on beta-ray sources are given in Table V. For $^{188}W/^{188}Re$, $^{90}Sr/^{90}Y$ and $^{106}Ru/^{106}Rh$ the emissions of the short-lived daughter are in equilibrium with the long-lived parent. Further, in these cases, only the beta energy of the daughter is of importance, because the relatively low energy beta particles of the parent are absorbed by the source encapsulation.

TABLE V. PHYSICAL DATA ON BETA-RAY SOURCES.

Beta emitter	Maximum energy (MeV)	Average energy (MeV)	Half life (days)
^{133}Xe	0.346	0.100	5.243
^{32}P	1.71	0.695	14.26
$^{188}W/^{188}Re$	2.12 (^{188}Re)	0.766 (^{188}Re)	69.4 (^{188}W)
$^{90}Sr/^{90}Y$	2.28 (^{90}Y)	0.933 (^{90}Y)	10512 (^{90}Sr)
$^{106}Ru/^{106}Rh$	3.54 (^{106}Rh)	1.42 (^{106}Rh)	373.6 (^{106}Rh)

Clinical planar sources of $^{90}Sr/^{90}Y$ have 4 to 9 mm active diameters (10 to 13 mm physical diameters) [10]. The concave $^{90}Sr/^{90}Y$ sources have an active diameter of 6 to 18 mm with a 10 or 15 mm radius of curvature. For $^{106}Ru/^{106}Rh$, only concave sources have been available, with 10 to 23.5 mm active diameters and 12 to 14 mm radii of curvature. Examples of typical ophthalmic plaques are shown in figures 2 and 3.

[4] See Section 6

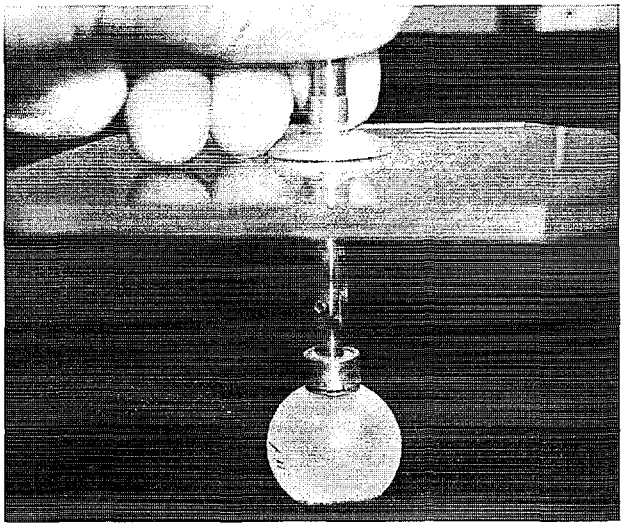

FIGURE 2. $^{90}Sr/^{90}Y$ eye plaque applied to a plastic eye phantom

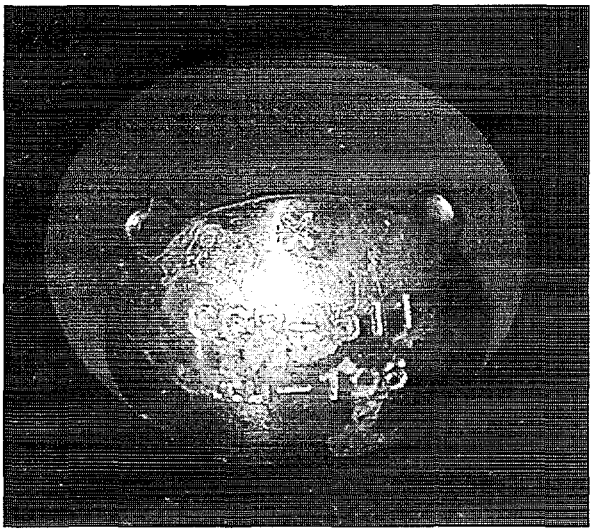

FIGURE 3. $^{106}Ru/^{106}Rh$ ophthalmic applicator with two suture holes.

3.3 BETA-RAY SEED AND WIRE SOURCES

In intravascular brachytherapy applications lesions in the coronary arteries are treated with either beta particles or photons. The lesions are usually on the order of 2 to 4 cm in length in arteries with diameters of 3 to 5 mm. This requires line sources of a very narrow diameter, less than 1 mm. Typical geometries include encapsulated line sources mounted on the end of wires that can be used to insert and remove the wires to and from the treatment point. Line sources may also be constructed from linear arrays of "seeds" which can be delivered to the lesion site either manually or pneumatically. Isotopes being used for these sources include ^{32}P, $^{90}Sr/^{90}Y$, ^{90}Y, and $^{188}W/^{188}Re$. The physical length of these sources varies but is generally 2 to 4 cm to adequately cover the lesions. The stepping of shorter wire sources is being investigated to treat longer lesions.

3.4 BETA-RAY BALLOON, SHELL AND STENT SOURCES

As with seed and line sources, radioactive liquid or gas filled balloons are being investigated for use in treating coronary and peripheral artery lesions. Isotopes being considered include ^{32}P, ^{188}Re (liquids) and ^{133}Xe (gas). Physical data for these isotopes are included in Table V. Balloon lengths being used conform to standard sizes of angioplasty balloons, which range from 2 to 4 cm in length and 2.5 to 3.5 mm in diameter. One concern with the use of such sources is the possibility of a balloon rupture that would release the radioactive fluid within the blood stream, or even worse, the creation of a gas bubble if there is a burst of a gas-filled balloon. There is also the concern for contamination, which is why short half-life sources are preferred for this application. In addition there is the presence of the radioactive medium throughout the length of the catheter, and the corresponding difficulty in assessing the amount of activity in the balloon versus what remains in the catheter.

An alternative approach to the delivery of dose by a balloon source is to use a balloon with a radioactive coating, which results in a cylindrical shell source. The only isotope that has been employed in this manner is ^{32}P. The advantage of such a source is that the activity is located very near the target, and thus less contained activity is required to achieve the desired dose rate than in a volume or a line source. However, since the encapsulation of such a source is minimal there are concerns for source integrity.

A special case of shell sources is a radioactive stent, the only current examples of which also employ ^{32}P. Like a normal non-radioactive stent, the stent source is deployed as a permanent implant, which makes it attractive as a source to interventional cardiologists. Quite modest activities on the order of 1 μCi have been shown to be effective in animal studies, however results in human clinical trials have

been so far disappointing and higher activities are being investigated. Since the activity is distributed on the surface or within the structure of the mesh-like stent, the dose distribution in the vicinity of the source is highly non-uniform.

4 CALIBRATION OF BRACHY-THERAPY SOURCES AT PSDLs

4.1 REFERENCE STANDARDS

4.1.1 Photon sources: Wide Angle Free Air Chamber (WAFAC)

Currently only NIST can provide reference air kerma rate calibrations for low energy photon sources. The calibration is accomplished with the WAFAC system developed by Loevinger [11]. In the new calibration procedure, the characteristic x-rays from the Titanium[5] capsulation are filtered out. These x-rays, having an energy of only 4.5 keV, does not have any effect on the dose in tissue at typical treatment distance of about 1 cm. On the other hand, they have a significant effect of approximately 10 % on the calibration signal. Prior to January 1999, when the WAFAC system was taken into use, the characteristic x-rays were not filtered out. This has therefore resulted in a change in brachytherapy ^{125}I source calibrations. The WAFAC is being used to establish calibrations for the many new ^{125}I and ^{103}Pd sources that are introduced in the market. These standards are then transferred to SSDLs (in the USA, Accredited Dosimetry Calibration Laboratories (ADCLs) in order to calibrate well type chambers for users. The WAFAC system is reviewed in detail in the ICRU Report [4].

4.2 BETA-RAY SOURCES: EXTRA-POLATION CHAMBER

The extrapolation chamber is a primary standard for the determination of absorbed dose rate of beta-ray sources. Constructional details and operational performance of extrapolation chambers are given in the ICRU Report [4]. By suitable construction, it can be used for all other type of beta sources except concave plaque sources.

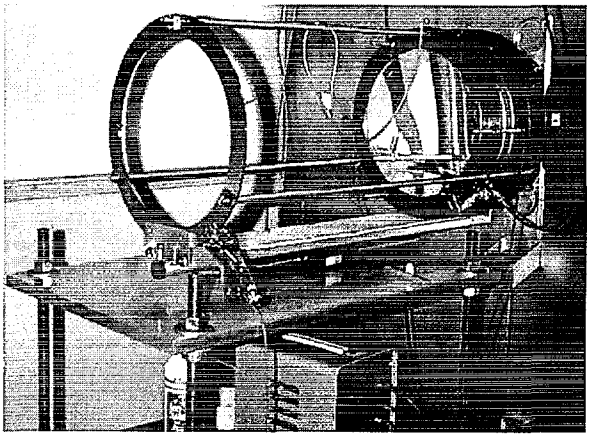

FIGURE 4. WIDE ANGLE FREE-AIR CHAMBER AT NIST.

The concave sources cannot be accurately calibrated by the extrapolation chamber due to the geometry, which does not allow to place the source close enough to the chamber. For concave plaque sources, therefore, recourse must be made to the calibrated detector approach.

The extrapolation chamber is basically an air-filled plane parallel chamber where the distance between the high-voltage and collecting electrodes (air gap) can be varied. The absorbed dose rate is determined from current measurements at a series of air gaps; the current values as a function of air gap are fitted to determine the slope of this data at the limit of zero air gap. The absorbed dose rate in water is then given by the Bragg-Gray relationship

$$D_w = \frac{(W/e)\cdot S_{air}^{water}}{\rho_0 a}(\Delta I/\Delta \ell)_{\ell\to 0} k_{back} \qquad (2)$$

where (W/e) is the average energy in joules needed to produce one coulomb of charge of either sign in dry air (33.97 $\pm$ 0.05 JC^{-1}), S_{air}^{water} is the ratio of the mean mass collision stopping power of water to that of air, ρ_0 is the density of air at the reference temperature and pressure (T_0, p_0), a is the area of the collecting electrode, $(\Delta I/\Delta \ell)_{\ell\to 0}$ is the rate of change of current (normalized to a reference temperature and pressure) with extrapolation chamber air-gap thickness as the thickness approaches zero, and k_{back} is a correction factor that accounts for the difference in backscatter from the collecting electrode compared to that of water.

Of critical importance is the area of the collecting electrode used, because accurate

[5] The most common capsule material used in low energy photon sources

knowledge of this area is needed for determining the dose rate from the measured currents, and this is the area that the measured dose rate will be effectively averaged over. It is also important that the area of the collecting electrode be smaller than the radiation field being measured, so that the measurement averaging area is determined by the collecting electrode rather than by the radiation field.

For accurate measurement of reference absorbed dose rate of beta-ray planar applicators, a collecting electrode diameter of about 4 mm is recommended to compromise between the requirement of point-like measurement and the uncertainty of the determination of the collection volume. Because of the effect of the divergence of the radiation field, it is recommended that the range of air gaps used be kept below 0.2 mm, with a sufficient number of air gaps employed to establish the functional character of the current versus air gap dependence. Other requirements on the extrapolation chamber and the measurement technique, including details of various correction factors, are discussed in the ICRU Report [4].

The extrapolation chamber can also be used to determine reference absorbed dose rate from a beta-ray emitting seed or wire source [12]. For these measurements the source is inserted in a hole in a tissue-equivalent plastic block with the center of the source at a distance of 2 mm from the block surface. At this depth, the radiation field from a seed or wire source is such that a collecting electrode diameter of 1 mm can be used to measure absorbed-dose rate. There are problems with this method, mainly due to an unacceptable large uncertainty (±7.5% at 1σ) which must be assigned to the measurement because of uncertainties in

- The effective collecting area of the extrapolation chamber
- The divergence effect of the small source/ collector geometry.

For this reason, it may be that the calibrated detector approach, described in the following section, should be used for the calibration of beta-ray brachytherapy seed and wire sources.

An additional possible future approach to the determination of a reference quantity for seed and wire sources is an in-air measurement with an extrapolation chamber, such as those used for measurement of protection-level beta-ray reference radiation fields. For this determination, the quantity dose rate in tissue at a depth of 0.07 mm, D(0.07mm), is measured at a distance in air from the source, positioned on a low-scatter support. The measurement is performed at a large distance, e.g. 30 cm in air with an extrapolation chamber with a large, e.g. 30 mm diameter collecting electrode with the same techniques and corrections as are applied to the measurement of protection-level beta-ray radiation fields. The measured quantity, absorbed dose rate to tissue at 7 mg/cm^2 measured at 30 cm in air, is related to reference absorbed dose rate at 2 mm in water from the same source via the conversion factor Λ_β which is defined as:

$$\Lambda_\beta = \frac{D_w(2\text{mm})}{D(0.07\text{mm},300\text{mm})} \qquad (3)$$

Values of Λ_β must be determined for each source type to be calibrated using this method, usually by a combination of measurements and Monte Carlo model calculations.

4.3 WORKING STANDARDS

For routine calibrations of brachytherapy sources at the PSDLs, the complicated and time-consuming measurements by WAFAC or extrapolation chambers are not always feasible. As working standards for routine calibrations, suitable calibrated detectors are applied also at the PSDL level. For low energy photon sources, and beta particle sources used for intravascular brachytherapy, the well type ionization chamber is the recommended working standard instrument. For beta-ray plaque sources, several possibilities are available as described in the following section. The general considerations and practical guidance on measurements do not differ from that which is appropriate for calibrations at SSDL level, and this is discussed in detail in Section 5.

5 CALIBRATION OF BRACHY-THERAPY SOURCES AT SSDLs AND HOSPITALS

Accurate measurements by the extrapolation chamber technique require careful construction of the chamber and exact consideration of a number of factors. The same is true for the development and use of special free air

chambers (WAFAC). Therefore these primary techniques are neither relevant nor feasible for application at the SSDLs. Instead, the use of a suitable calibrated detector, as a reference and working standard of the SSDL must be considered. For low energy photon sources and for beta-ray sources used in intravascular brachytherapy, this is a calibrated well type ionization chamber. For other sources, other calibrated detectors may be used at the SSDLs. At the user level in hospitals, similar equipment can be used for QA checks of the manufacture source calibrations.

Other techniques mentioned later can be useful for other source parameter characterization. For example, TLDs in phantoms have been used extensively for the measurement of absorbed dose rate for photons.

For beta-rays, in principle any detector whose output can be related to absorbed dose or dose rate can be used to determine reference absorbed dose rate of beta-ray brachytherapy sources. However, due to the low penetration of beta particles, the detector needs to approximate as much as possible an ideal point-like detector. The most important characteristic of a beta particle detector is its thickness. In order to reduce the energy dependence to a minimum, it should be as thin as possible. For good lateral spatial resolution, the area should be as small as possible. However, both these requirements come at the expense of sensitivity, and therefore compromises must be made for real-world detectors. Some detector systems which approach the required properties are radiochromic film, thin plastic scintillators, thin thermoluminescence dosimeters (TLDs), diode detectors, diamond detectors, thin alanine photo-stimulated luminescence (PSL) systems and radiochromic gel dosimeters. There are number of practical and technical characteristics of the detector systems which are independent of the sources to be calibrated. These characteristics for a few detector systems are summarized in Appendix A. A number of other characteristics, where the suitability of the detector is dependent on the sources to be measured, are summarized in Appendixes B to D. The characteristics of a number of detectors are also discussed in detail in the ICRU Report [4].

5.1 CALIBRATION OF LOW ENERGY PHOTON SOURCES

The calibration of photon reference sources at the PSDL allows calibration of SSDL well type ionization chambers. These chambers are then used to calibrate hospital well type ionization chambers so that the sources provided by the manufacturer can be measured. The free-in air calibration technique, which is mentioned in IAEA TECDOC-1079, is not recommended for the following reasons:

- The sources are of low intensity,
- An appropriate calibration factor for an air cavity chamber at these energies is difficult to establish.

These two requirements will lead to a large uncertainty, approximately 7% or even greater.

Since many seeds (e.g. 100 per implant) are used for prostate treatments, it becomes a great deal of work to measure each individual seed. AAPM TG-56 [13] therefore suggests that at least 10% of all seeds be measured before they can be used clinically. This is an interim procedure until measurement equipment becomes available to make the measurement of many individual seeds in an efficient manner. Ranges are established by manufacturers, who sort the seeds in groups of similar air kerma rate. The seed to seed variation of reference air kerma rate, measured by the user, of seeds purchased from such a group, may show a deviation of as high as 10% from the average of the batch. It is suggested that the mean of the measured reference air kerma rate agrees with the manufacturer's value to within 3% and the variation of the seeds measured is within ±5% of the mean [13].

As of January 1, 1999, the revised primary standard for ^{125}I sealed sources was implemented. The source strengths changed for the sources that were in use for a number of years, namely Amersham 6711 and 6702 ^{125}I seeds. As a result, calibration of brachytherapy well type ionization chambers that were calibrated for these sources prior to 1999 will change accordingly. Compared to seeds marketed prior to this date, calibration values will numerically decrease by 10.3% as in Equation 4 below. For both models of ^{125}I seeds which this change effects (6711 and 6702)

$K_{R,85std}$ and $K_{R,99std}$ are related by[6]:

$$K_{R,99std} = K_{R,85std} \cdot 0.897 \quad (4)$$

Since 1 January 1999, calibration factors based upon the new standard from NIST have been provided. Nycomed Amersham instituted the new standard on 26 July 1999. For ^{103}Pd, Theragenics has not yet adopted the new standard.

With the revision of the primary standard, corresponding adjustments must be made to the pre-1999 calibration factors used with well type ionization chambers to verify vendor calibrations. The multiplicative calibration factors to convert the reading to reference air kerma rate, $N_{SK,\ 85std}$, have to be modified. To verify seed calibrations traceable to the new standard, either a new correction factor must be obtained from an SSDL or the old factor (pre-1999) needs to be modified as follows:

$$N_{K,99std} = N_{K,85std} \cdot 0.897 \quad (5)$$

The product of the instrument reading, the revised factor $N_{Sk,99std}$, and other corrections independent of the calibration standard (e.g., temperature and pressure corrections), will now represent the Reference Air Kerma Rate, $K_{R,99std}$, traceable to the revised standard. Note that these factors do not apply to other similar sources from other manufacturers. Independent calibrations must be obtained for each manufacturer seed and if there are any changes in seed construction, a new factor will need to be obtained.

It should be noted that pressurized well type ionization chambers used in the Nuclear Medicine Department are not recommended for brachytherapy measurements due to the following reasons:

- The chambers measure only in units of activity, which is a derived quantity
- The chambers have settings for given radionuclides but not brachytherapy sources
- Without close control, the general use of the chamber may result in contamination from nuclear medicine procedures
- Since the gas may leak from the pressurized volume, the response may change over time
- The thick walls required for the pressurization may absorb part of the radiation to be measured. Since this results in a high-energy dependence, small variations in the relative peak intensities are unduly emphasized.

[6] '99std' and '85std' refers to the years, 1999 and 1985 when the standards at NIST were taken into use.

5.2 CALIBRATION OF BETA-RAY SOURCES

The measurement of reference absorbed dose rate with the calibrated detector should be carried out in a water phantom whenever possible. When this is not possible or convenient (cf. column 7 of the Table in Appendix A), as in the case of some radiochromic film, TLDs, alanine and other water-sensitive detectors, recourse must be made to water-equivalent plastics. Water-equivalent epoxies (e.g. Solid water™, WT1), A-150 tissue equivalent plastic or polymethyl methacrylate (PMMA) can be used as water-equivalent plastics. Polystyrene, however, is recommended as the best substitute for water for these energies of electrons.

Since most of the available detectors are not ideal point-like detectors, as a quality assurance procedure the calibration measurements should be confirmed by measurements with another detector or by Monte Carlo calculations whenever possible.

5.2.1 Beta-ray plaque sources

For the determination of dose rate at the reference distance of 1 mm, measurements along the axis of the source (perpendicular to the source plane for planar sources) should be carried out. Starting from the "zero distance" where the detector is in contact with the source surface, or as close to the surface as possible, measurements should include a point where the effective point of measurement of the detector is at the distance of 1 mm or close to it. The accurate distances for measurements in water can be ensured by using a gauge with accurately known thickness (uncertainty of thickness less than 0.05 mm) between the source and the detector. For all other distances, the detector should be moved with a micrometer-driven holding system that enables relative movements with a precision of at least 0.05 mm. For the same measurements in solid phantoms, the spherical caps (to accurately

touch the surface of concave sources) and plates of different thickness should be machined with a tolerance of less than or equal to 0.05 mm.

The absorbed dose rate at the reference distance of 1 mm should be determined from the measurement results, either directly, by curve fitting, or by accurate interpolation of values close to the reference point of 1 mm. For non-water measurements, the density of the phantom material must be considered in the specification of the measurement depth.

The central axis depth dose curve relative to the absorbed dose rate at the reference distance of 1 mm should be compared with the reference curve given in Table VI (reproduced from the ICRU Report [4]). To the first approximation, the relative depth dose values obtained, down to about 5 mm depth, are expected to conform within about 10 % to the reference curve values.

The reference data in Table VI is the average data from measurements with several detectors and confirmed by close agreement with Monte Carlo calculated data. For the purposes of interpolation of these averages, the following equation may be used:

$$\frac{D(z,r_0)}{D(z_0,r_0)} = \exp(a_s + b_s z + c_s z^2 + d_s z^3 + e_s z^4 + f_s z^5) \quad (6)$$

where z is the depth, expressed in mm of water equivalence. The values of the coefficients of this function are given in Table VII for three plaque source geometries.

TABLE VI. RELATIVE AXIAL DEPTH-DOSE DISTRIBUTION IN WATER FOR A PLANAR ^{90}Sr SOURCE AND FOR A PLANAR AND CONCAVE ^{106}Ru SOURCES.

Depth (mm)	^{90}Sr/^{90}Y planar	^{106}Ru/^{106}Rh planar	^{106}Ru/^{106}Rh concave
0.0	1.752	1.351	1.115
0.5	1.342	1.165	1.069
1.0	1.000	1.000	1.000
1.5	0.734	0.855	0.915
2.0	0.533	0.727	0.824
3.0	0.272	0.515	0.644
4.0	0.127	0.353	0.484
5.0	0.052	0.233	0.353
6.0	0.018	0.148	0.249
7.0	--	0.090	0.170
10.0	--	0.019	0.043

TABLE VII. Coefficients of the fitted relative depth-dose functions of beta-ray sources

Coefficient	^{90}Sr/^{90}Y planar	^{106}Ru/^{106}Rh planar	^{106}Ru/^{106}Rh concave
a_s	0.5608	0.3008	0.1089
b_s	-0.4913	-0.2928	-0.05458
c_s	-0.09887	-0.007527	-0.06305
d_s	0.03619	-0.0001728	0.008861
e_s	-0.007232	-0.0002206	-0.0007853
f_s	0.0004487	0.00001792	0.00002589

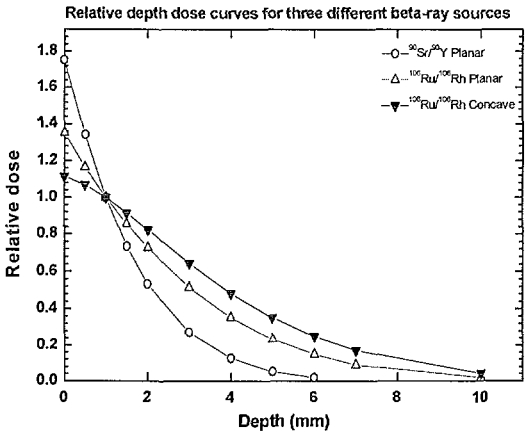

FIGURE 4. Relative central depth dose curves for three beta-ray sources normalized at 1mm.

5.3 INTRAVASCULAR SOURCES

Because of the small source dimensions, close distances, very divergent radiation fields and very high absorbed dose rate gradients from these sources, measurements with calibrated detectors present serious challenges. Generally speaking all these problems are lessened with increasing distance from the source, however this comes at the price of much reduced signal due to the steeply sloped depth dose curves from these sources. For near geometry (<5 mm) measurements, care must be taken to account for dose deposition profiles both in the vertical as well as the lateral dimensions of the detection element. Knowledge of the proper corrections to make in these fields requires *a priori* knowledge of the expected three-dimensional absorbed dose profile, which often is not specified for newer source designs. For this reason, measuring reference absorbed dose rate from intravascular brachytherapy sources with calibrated detectors is not recommended for users and should be approached with

extreme caution even by experienced SSDLs. The preferred method for both SSDLs and users is the use of a good-quality well type ionization chamber with a calibration for the particular source geometry in question traceable to a PSDL.

5.4 CORRECTION FACTORS

Due to the finite size of all available detectors, and the presence of covering material or other constructional elements of some detectors, the following corrections should be considered for accurate measurements at a point:

- Correction for offset depth due to covering material and finite thickness of the detector.
- Correction for the effective point of measurement of the sensitive volume of the detector.
- Correction for geometry for measurements in contact with concave sources.
- When the measurements are carried out in water-equivalent plastics instead of water, the depth of measurement should be scaled to the corresponding depth in water.

5.4.1 Correction for offset depth

The offset depth is the estimated separation between the detector surface and the centre of the detector. It is equal to the thickness of the covering material plus half of the thickness of the sensitive volume of the detector. Offset depths of a few commercially available detectors are given in Appendix E. The values in this Table should be regarded as nominal only; there may be individual differences in these values for a given type of detector, and the value may also differ from the nominal value derived from the specifications given by the detector manufacturer. It is recommended that the covering thickness be checked by radiography for each individual detector.

5.4.2 Correction for effective point of measurement

Using the detector centre as the effective point of measurement is only valid for detectors in fields with a linear dose gradient across the dosimeter. The actual effective point of measurement of a detector of finite thickness is the depth of an infinitely thin detector that gives the same dose rate as that averaged over the detector of finite thickness. If the relative central axis depth dose function, D(z, r=0), is available, the average relative dose $D_{avg}(t,z)$ across a dosimeter of thickness t with its surface at depth z is given by

$$D_{avg}(t,z) = \frac{\int_z^{z+t} D(z, r=0)dz}{t} \qquad (7)$$

The effective point of measurement, for the given source, detector and depth, can then be obtained from this by determining the root (value of z) for the depth dose function which gives D_{avg}. For the detectors given in Appendix C, the maximum shift of the effective point from the centre is about 0.2 mm, corresponding to an error in dose about 9 %; for most of the cases, the error would be much smaller.

5.4.3 Correction for detector geometry

When a concave source is in contact with a rigid cylindrical detector, the detector surface does not touch the source surface except right on the edges of the detector. This creates a geometrical "offset" for the "zero" distance measurement, which depends on the radius of curvature of the source, R, and on the physical diameter of the detector, d. This geometrical offset, k_G, can be calculated by

$$k_G = R - (R^2 - d^2/4)^{1/2} \qquad (8)$$

For example, for the PTW diamond detector, with R=12 mm and d=7.1 mm, k_G = 0.54 mm.

5.4.4 Scaling from water equivalent plastic to water

Beta dose distributions can be approximately scaled from one medium to another, as described in the ICRU Report [4]. For point sources in infinite media, the dose rate, $D_m(r_m\rho_m)$, at a distance r_m corresponding to an areal density of $r_m\rho_m$ (in g/cm^2) in the medium, is related to the dose rate in water, at the same areal density, $r_w\rho_w$, but scaled, by

$$D_m(r_m\rho_m) = (\eta_{m,w})^3(\rho_m/\rho_w)^2 D_w(\eta_{m,w}\, r_w\rho_w) \qquad (9)$$

where $\eta_{m,w}$ is the scaling factor of the medium relative to water and ρ_w and ρ_m are the densities of water and the medium respectively. It should be noted that the scaling factor has the nature of a ratio and thus $\eta_{m,w} = 1/\eta_{w,m}$. The scaling factors $\eta_{m,w}$ for the water-equivalent plastics recommended in this guide are given in Table

VIII.

Table VIII. Scaling factor $\eta_{m,w}$ for water-equivalent plastics recommended in this guide.

Plastic	Density (g/cm^3)	Scaling factor, $\eta_{m,w}$, relative to water
A-150 tissue-equivalent	1.127	0.968
Polystyrene	1.05	0.938
PMMA	1.19	0.949
WT1, ("solid water")	1.02	0.957

An alternative approach to scaling for non-point-like geometries is to carry out Monte Carlo simulations of the same source in the two different media. Scaling is then calculated from a comparison of the depth doses in the two media.

5.5 CALIBRATION OF THE DETECTOR

5.5.1 Well type ionization chambers for calibration of low energy photon sources and intravascular brachytherapy sources

The preferred method of calibration is against the WAFAC (for the PDSL) and well chambers (for the SSDL and at the radiotherapy centres). The use of well type chambers and their characteristics have been published by the IAEA [7]. This reference includes the manner in which these chambers should be used. In brief, the SSDL should obtain a calibrated ^{125}I, ^{103}Pd or intravascular brachytherapy source for the various models desired and then calibrate the well ionization chamber in their laboratory. Alternatively, the SSDL can have their well ionization chamber calibrated for the source types they desire. The SSDL then will calibrate the user's well ionization chamber for the sources. The procedures for these calibrations and maintaining checks on them are given in TECDOC-1079 [7]. As a quality assurance check, a ^{137}Cs or other long half-life source should be measured periodically to monitor the long-term stability of the chamber.

It is very important that the design of the source holder be consistent in each step through the PSDL to the SSDLs to the users. The positioning of the source within the well ionization chamber volume must be well specified and reproducible. The centre of the source shall be located at the calibration point as defined by the PSDL or the SSDL.

5.5.2 Detectors for beta-ray plaque sources

The preferred method of calibration of the detector is against extrapolation chamber measurements of reference absorbed dose rate in the field of a relevant planar beta-ray reference source at a PSDL. The uniformity of dose rate over the area of the planar reference source as given by the uniformity parameter should be better than 10% but shall in no case exceed 20 % [4]. This calibrated source can then become the secondary standard of the SSDL to calibrate other detectors.

When a suitable calibrated planar reference source is not available, the calibration of the detector can be carried out in a high-energy photon (usually a ^{60}Co) or electron beam, where the dose rate is determined by measurements with an air kerma- or absorbed dose to water-calibrated ionization chamber. There are many hazards associated with this technique. Consideration must be given for the possible effects of dose rate or the dependence of energy and radiation type on the response of the detector (see Appendix A). The effective point of the detector must be placed at the depth in phantom where absorbed dose to water is specified.

5.6 CALIBRATION UNCERTAINTY

5.6.1 Low energy photon sources

The overall uncertainty for calibrating well type ionization chambers for low energy photon sources is 1.2 % at 1σ.

5.6.2 Beta-ray sources

Since the uncertainty of measurements with calibrated detector/source systems is dominated by the uncertainty in the primary calibration of the planar source, calibrations with any of the systems shown in Appendices C and D, all exhibit approximately the same degree on uncertainty. An example uncertainty analysis is given in Table IX. The estimated combined uncertainty for measurements with calibrated detectors is 8 to 10 % for planar and concave beta-particle ophthalmic sources, and even

higher for intravascular brachytherapy sources.

TABLE IX. UNCERTAINTY ANALYSIS FOR A CALIBRATED DETECTOR SYSTEM

Component	Type A (%)	Type B (%)
Calibration of beta-particle planar reference source	0.4	6
Response of calibration films exposed to standard source		3
Response of films exposed to source under test		3
Combined uncertainty (quadratic sum)	7.4	

5.7 TRACEABILITY OF SOURCE CALIBRATIONS

The standards applied and the traceability of calibrations at different levels is summarized in Appendix F.

5.7.1 Low energy photon sources

An ^{125}I or ^{103}Pd photon source is calibrated at a PSDL with the WAFAC which is the primary standard. The SSDL then can use a calibrated source from the PSDL to calibrate their well ionization chamber with a direct traceability to the primary standard. Thereafter the SSDL can use this type of source to calibrate a user well ionization chamber.

5.7.2 Beta-ray sources

When a planar reference source is calibrated with an extrapolation chamber there will be a direct traceability to a primary standard. This planar source can then serve as the secondary standard source at the SSDL to be used to calibrate other detectors. This is the recommended method to establish traceability. When the detector is calibrated against absorbed dose to water measurements by ionization chambers at high-energy photon or electron beams, the traceability is that of the calibration of the ionization chamber.

6. CALCULATION OF DOSE CLOSE TO LOW ENERGY PHOTON SOURCES

Due to the recent change in the NIST standards for low energy photon sources some parameters in the dose calculation formalism must be changed accordingly. Moreover, the dosimetry constants for ^{125}I proposed by TG-43 results in calculated dose rates that may be reduced as much as 17 % from former values [14] recommended by the AAPM. It must be therefore strongly emphasized to use TG-43 in the dose rate calculation. The dose rate constants for ^{125}I given in TG-43 and in Table X below, apply only to the source models 6711 and 6702 (cf. Table III). These must not be used with any other ^{125}I brachytherapy sources due to possible differences in encapsulation and source construction.

The relation between the Air Kerma Strength, S_K, and the Reference Air Kerma Rate, K_R, is given by

$$S_K = r_{ref}^2 \cdot K_R \qquad (10)$$

where r_{ref} is the reference distance of 1 meter. Because the numerical value of the reference distance is unity, the numerical values of S_K and K_R must be equal. This means that the same formalism, without any changes in the numerical values of the constants and factors given by TG-43 can be used irrespective whether the source is calibrated in terms of Air Kerma Strength or Reference Air Kerma Rate. In other words, these two quantities are interchangeable.

The dose rate can be calculated using the formalism:

$$\dot{D}(r,\theta) = K_R \cdot \Lambda \cdot \frac{G(r,\theta)}{G(r_0,\theta_0)} \cdot g(r) \cdot F(r,\theta) \qquad (11)$$

where:

K_R is the Reference Air Kerma Rate,

Λ is the dose rate constant, i.e. the dose rate in water at a distance of 1 cm on the transverse axis per unit Reference Air Kerma Rate.

$G(r,\theta)$ is the geometry factor accounting for the variation of relative dose rate due to the spatial distribution of the activity within the source. The reference point, (r_0,θ_0), is chosen to lie on the transverse bisector of the source at a distance of 1 cm of its center, i.e. $r_0 = 1$ cm and $\theta_0 = \pi/2$

$g(r)$ is the radial dose function accounting for the effects of absorption and scatter in the medium along the transverse axis of the source

$F(r,\theta)$ is the anisotropy function, which accounts for the anisotropy of dose rate distribution around the source, including the effects of absorption and scatter in the medium.

Suggested procedures both for adopting the TG-43 dosimetry protocol and for implementing the revised NIST air-kerma strength standard were published in [15].

As was discussed previously, the change in the standard for ^{125}I was to eliminate the contribution of the titanium 4.5 keV x-rays. Since the dose is specified at 1 cm, the 4.5 keV x-rays do not contribute to the dose at 1 cm in tissue, since they are effective only to about 1 mm in tissue. However, in air they do contribute to the measurement. For this reason they must be eliminated from the primary measurement. Because in tissue the 4.5 keV x-rays only affect the dose to 1 mm, use of the TG-43 formalism with the current values for g(r) may underestimate the dose for endovascular cases.

The values for the dose rate constant, Λ, for two ^{125}I brachytherapy source are given in Table X.

TABLE X. Dose rate constants for two ^{125}I interstitial brachytherapy sources.

Source model	Λ
6711	0.98
6702	1.04

It is again emphasized that the dose rate constants in Table X applies only to the models indicated and when used together with the TG-43 formalism.

Values of the other constants and functions in equation 11 are given in [1].

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APPENDIX A.

TABLE. Summary of suitability of different detectors for brachytherapy dosimetry: Characteristics that are source independent

Detector	Availability	Long term stability	Dose linearity	Dose rate dependence	Dependence on environnemental conditions	Use in water	Real-time measurement	Cost
Radiochromic film	Good	Fair	Fair	Good	Poor	Fair	Poor	Fair
TLD(LiF)	Good	Poor	Poor	Good	Fair	Poor	Poor	Fair
Plastic scintillator	Poor	Fair/ Poor	Good	Fair	Fair	Fair	Fair	Poor
Diode	Fair	Poor	Fair	N/A	Fair	Fair	Fair	Fair
Alanine	Fair	Fair	Fair	Fair	Poor (?)	Poor	Poor	Poor
PSL	Fair	Poor	Good	N/A	Poor	Poor	Poor	Poor
Diamond	Poor	Fair	Poor	Poor	Fair	Fair	Fair	Poor
Parallel-plate ion chamber	Fair	Fair	Poor	N/A	Poor	Fair	Fair	Fair/ Poor
Polymer gels	Fair / Poor	Fair	N/A	N/A	Poor	Fair	Poor	N/A

APPENDIX B.

TABLE. Summary of the suitability of different detector systems for the calibration of low energy photon sources

Detector	Size/spatial resolution		Water equivalence	Sensitivity	Reproducibility	Dose rate dependence	Energy dependence	Directional dependence
	Lateral	Depth						
Radiochromic film	Good	Good	Good	Poor	Poor	N/A	Poor	Fair
TLD (LiF)	Poor	Fair	Fair	Fair	Fair	N/A	Poor	Poor
Plastic scintillator	Poor	Fair	Good	Fair	Fair	N/A	Poor	Fair
Diode	Fair/ Poor	Fair	Poor	Fair	Fair	N/A	Poor	Poor
Alanine	Poor	Poor	Good	Poor	Fair	N/A	Fair	Fair
PSL	Good	Good	Poor	Good	Fair	N/A	Poor	N/A
Diamond	Poor	Fair	Fair	N/A	N/A	N/A	Poor	N/A
Parallel-plate ion chamber	N/A	N/A	N/A	N/A	N/A	N/A	Good	N/A
Polymer gels	Fair	Fair	Good	N/A	N/A	N/A	Good	Fair

APPENDIX C.

TABLE. Summary of the suitability of different detector systems for the calibration of beta-ray ophthalmic applicators

Detector	Size/spatial resolution		Water equivalence	Sensitivity	Reproducibility	Dose rate dependence	Energy dependence	Directional dependence
	Lateral	Depth						
Radiochromic film	Good	Good	Good	Poor	Poor	N/A	Good	Fair
TLD (LiF)	Poor	Fair	Fair	Fair	Fair	N/A	Good	Poor
Plastic scintillator	Good	Fair	Good	Fair	Fair	N/A	Good	Fair
Diode	Fair/ Poor	Good	Poor	Good	Fair	N/A	Fair	Poor
Alanine	Poor	Fair	Good	Poor	Fair	N/A	Fair	Fair
PSL	Good	Good	Poor	Good	Fair	N/A	Fair	N/A
Diamond	Poor	Fair	Good	Good	Fair	Poor	Fair	Fair
Parallel-plate ion chamber	Poor	Fair/ Poor	Fair	Poor	Good	N/A	Fair	Poor
Polymer gels	Fair	Fair	Good	N/A	N/A	N/A	Fair	N/A

APPENDIX D.

TABLE. Summary of the suitability of different detector systems for the calibration of beta-ray seed and line sources

Detector	Size/spatial resolution		Water equivalence	Sensitivity	Reproducibility	Dose rate dependence	Energy dependence	Directional dependence
	Lateral	Depth						
Radiochromic film	Good	Good	Good	Poor	Poor	N/A	Good	Fair
TLD (LiF)	Poor	Fair	Fair	Fair	Fair	N/A	Good	Poor
Plastic scintillator	Poor	Fair	Good	Fair	Fair	N/A	Fair	Fair
Diode	Fair/ Poor	Fair	Poor	Fair	Fair	N/A	Poor	Poor
Alanine						N/A	N/A	N/A
PSL	Good	Good	Poor	Good	Fair	N/A	N/A	N/A
Diamond	N/A	N/A	N/A	N/A	N/A	N/A	N/A	N/A
Parallel-plate ion chamber	N/A	N/A	N/A	N/A	N/A	N/A	N/A	N/A
Polymer gels	N/A	N/A	N/A	N/A	N/A	N/A	N/A	N/A

APPENDIX E.

TABLE. Characteristics of a few commercially available detector systems.

Detector	Effective thickness (mm)	Covering thickness (mm)	Offset depth mm (mg/cm^2)	Measurement diameter (mm)	Physical diameter (mm)
Radiochromic Film 6-8 μm emulsion layer on 0.1 mm PTP* backing	0.007	0	0 (0.6)	Diameter of laser beam for absorbance measurements	Selectable
Radiochromic Film 16-18 μm emulsion layer on 0.1 mm PTP* backing	0.0017	0	0 (0.12)	Diameter of laser beam for absorbance measurements	Selectable
LiF:Mg,Ti cylindrical pellets Type MTS-N	0.3	0	0.15 (39.6)	5	5
LiF:Mg,Ti cylindrical pellets Type MTS-N	1.0	0	0.5 (132)	5	5
Alanine: L-α-alanine crystals mixed with paraffin binder, by A. Weiser Messtechnik	1.2	0	0.6 (66.4)	4.9	4.9
Plastic scintillator of Essen type (PTW)	0.4	0.2 (polyethylene)	0.4 (39.2)	1	6
PTW diamond detector	0.3	0.65 (polystyrene)	0.8 (103)	4	7.1

*PTP: polyethylene terephthalate

APPENDIX F.

TABLE. Traceability of calibrations and calibration checks for brachytherapy sources

Step	Photon sources, long-lived nuclides *All clinical sources to be calibrated*	Photon sources, short-lived nuclides *All or random sample, min. 10 % of clinical sources to be calibrated*		Beta-ray sources *All clinical sources to be calibrated*		
	^{137}Cs, (^{60}Co)	^{192}Ir	^{125}I, ^{103}Pd	^{90}Sr-^{90}Y, ^{106}Ru-^{106}Rh, ^{32}P		
				Planar Sources	Concave sources	Seed sources
Reference standard at PSDL	Spherical graphite cavity chamber (LDR). Free in-air measurements	Spherical graphite cavity chamber, free in-air measurement (LDR). Interpolative calibration by free in-air measurements (HDR).	WAFAC (titanium x-rays excluded)	Extrapol. chamber	Calibrated detector +planar reference source	Calibrated detector + planar reference source
Working standard at PSDL	Large volume ionization chamber, free in-air measurements +reference source	Well type ionization chamber	Well type ionization chamber	Extrapol. chamber or calibr. Detector +calibrated planar source	Calibrated detector	*Well type ionization chamber* +ref. source
Standards at IAEA* laboratory, SSDL or ADCL and supplier's laboratory	Well type ionization chamber +reference source (LDR & HDR)	*Ionization chamber with Interpolative calibration factor (HDR).* Well type ionization chamber (HDR & LDR).	Well type ionization chamber	Calibrated planar source +Calibrated detector	Calibrated planar source +Calibrated detector	Well type ion. Chamber +reference source
Hospital user	Well type ionization chamber +reference source	Well type ionization chamber	Well type ionization chamber	Calibrated Detector	Calibrated detector	Well type ion. Chamber +ref. source

* Currently the IAEA provides well type chamber calibrations for LDR ^{137}Cs quality only