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Calibration of photon and beta ray sources used in brachytherapy

*Guidelines on standardized procedures at
Secondary Standards Dosimetry Laboratories (SSDLs) and hospitals*



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FOREWORD

It has generally been recognized that international harmonization in radiotherapy dosimetry is essential. Consequently, the IAEA has given much effort to this, for example by publishing a number of reports in the Technical Reports Series (TRS) for external beam dosimetry, most notably TRS-277 and more recently TRS-398. Both of these reports describe in detail the steps to be taken for absorbed dose determination in water and they are often referred to as 'dosimetry protocols'. Similar to TRS-277, it is expected that TRS-398 will be adopted or used as a model by a large number of countries as their national protocol.

In 1996, the IAEA established a calibration service for low dose rate (LDR) ^{137}Cs brachytherapy sources, which is the most widely used source for treatment of gynecological cancer. To further enhance harmonization in brachytherapy dosimetry, the IAEA published in 1999 IAEA-TECDOC-1079 entitled "Calibration of Brachytherapy Sources. Guidelines on Standardized Procedures for the Calibration of Brachytherapy Sources at Secondary Standard Dosimetry Laboratories (SSDLs) and Hospitals". The report was well received and was distributed in a large number of copies to the members of the IAEA/WHO network of SSDLs and to medical physicists working with brachytherapy.

The present report is an update of the aforementioned TECDOC. Whereas TECDOC-1079 described methods for calibrating brachytherapy sources with photon energies at or above those of ^{192}Ir , the current report has a wider scope in that it deals with standardization of calibration of all the most commonly used brachytherapy sources, including both photon and beta emitting sources. The latter sources have been in use for a few decades already, but their calibration methods have been unclear. Methods are also described for calibrating sources used in the rapidly growing field of cardiovascular angioplasty. In this application, irradiation of the vessel wall is done in an attempt to prevent restenosis after cardiovascular interventions. The present report includes a description of suitable detector systems that can be used for the calibration.

It must be emphasized that for safe use of brachytherapy a comprehensive quality assurance (QA) programme should be developed at the radiotherapy center using this modality. A QA programme cannot rest on a source calibration alone, but in addition it should address all the different steps included in the treatment process. Such a programme is described in IAEA-TECDOC-1040, "Design and Implementation of a Radiotherapy Programme: Clinical, Medical Physics, Radiation Protection and Safety Aspects". As summarized in the present report, omission of a QA programme may have serious consequences for a patient undergoing brachytherapy treatment.

The parts of this publication describing the calibration of low energy photon sources and beta ray sources have been written in close collaboration with members of the International Commission on Radiation Units and Measurements (ICRU). Their valuable contributions to this report are acknowledged.

The IAEA staff members responsible for this publication were H. Töllli and A. Shanta of the Division of Human Health.

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1. INTRODUCTION

Brachytherapy uses encapsulated radioactive sources to deliver a high dose to tissues near the source. An important part of a general quality assurance (QA) programme for brachytherapy dosimetry is the source calibration. For some brachytherapy sources, vendors assign large uncertainties to their stated calibration values, in some cases up to $\pm 10\%$. End-user calibration of brachytherapy sources is necessary, not only to check vendor stated calibration but to ensure traceability to internationally accepted standards. Quoting from the report of Task Group 40 (TG-40) of the American Association of Physicists in Medicine (AAPM), “Each institution planning to provide brachytherapy should have the ability to independently verify the source strength provided by the manufacturer” [1].

This report discusses calibration techniques of the most commonly used gamma and beta ray brachytherapy sources. For ^{137}Cs low dose rate (LDR) source calibrations, the IAEA Dosimetry Laboratory maintains reference sources calibrated at a PSDL. These sources can be used to calibrate well type ionization chambers maintained at the SSDLs. In Section 7 of this report ^{137}Cs LDR calibrations are discussed in detail.

Included in this report is also a description of suitable detector systems to be used in the calibrations. The Primary Standards Dosimetry Laboratories (PSDLs) make use of the most accurate and sensitive calibration systems, which in many cases cannot be used at SSDL or at a hospital. In the evaluation of uncertainties of the source calibration at different steps in the calibration chain, it is important to know the component uncertainties contributed at each level. Thus, in order to make a meaningful assessment of the total uncertainty in the calibration of a particular source at the radiotherapy center, the user needs to know the corresponding uncertainties at the SSDL and the PSDL to which the calibration is traceable. It is therefore important to have knowledge of the calibration systems used at each level. The present report gives a description of the systems used at the PSDLs and recommends suitable detectors for calibration for SSDLs and hospitals. In particular at the hospital level, the detector should be reliable and easy to use. For photon emitting brachytherapy sources, and for some of the beta ray sources, the detector of choice is the well type ionization chamber. Even though these chambers are usually of rather robust construction, quality control measures must be performed in order to assess the performance of these chambers, as is done with all other types of ionization chambers. Such a programme is described in Section 9.

From the point of view of traceability to primary standards, calibration of ^{192}Ir high dose rate (HDR) sources is a special case. At the time of preparing this report, no PSDL maintains standards for this radiation quality. At several PSDLs, however, intensive research is underway to establish such standards. Today, the calibration of these sources is based on an interpolation technique in which established standards for external beams are used. This technique yields a link for the calibration of ^{192}Ir HDR sources to primary standards that is not as robust as is the case for instance for ^{137}Cs LDR brachytherapy sources.

2. CHARACTERIZATION OF BRACHYTHERAPY SOURCES

The following section gives the recommended quantities for brachytherapy source specification. A discussion is also given of other quantities that are useful in clinical applications.

2.1. Specification

2.1.1. Gamma ray sources

The recommended quantity for the specification of the gamma sources is the reference air kerma rate, defined by the ICRU [2, 3, 4] as the kerma rate to air, in air, at a reference distance of one meter, corrected for air attenuation and scattering¹. For needles, tubes and other similar rigid sources, the direction from the source centre to the reference point shall be at right angles to the long axis of the source. The SI unit of reference air kerma rate is Gy·s⁻¹ but for the purposes of source specification it is more convenient to use μGy·h⁻¹ for LDR brachytherapy sources, progressing to μGy·s⁻¹ and mGy·h⁻¹ for HDR applications.

2.1.2. 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.

2.2. Summary of recommended calibration quantities

Table I summarizes the quantities recommended for specification of brachytherapy sources. The recommendations are in agreement with those given by the ICRU [4].

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

¹ At present there is an ongoing discussion on the proper definition of reference air kerma rate. The definition given in this publication agrees with that given in the ICRU Reports 38 and 58.

2.3. Other important quantities

Whereas the reference air kerma rate and the reference absorbed dose rate are sufficient to yield traceability of the source calibration, it is important that other quantities 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 useful 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 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, determined at a depth of 1 mm in a water-equivalent medium, over a specified area of the source². The value of this parameter should not exceed 20 % [4].

A map of uniformity or a few dose profiles across the source should be available as part of source specification.

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 the comparison between model predictions using Monte-Carlo techniques and dosimetry 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 be determined from a destructive measurement, which involves dissolving the 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, with a relative uncertainty between 1 to 2% ($k=1$) 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

² A more precise definition is given in Ref. [4].

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 centre 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 [4]. 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.

2.4. **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 (cf. Section 9).

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 [7]. The apparent activity is related to the reference air kerma rate, K_R , by

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

where $(\Gamma_\delta)_K$ is the air kerma rate constant³ 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.

³ The index δ in the air kerma rate constant indicates that only photons with energies greater than δ are considered. Photons with energies below this limit are assumed to be absorbed in the source or in the capsule.

The problem in the use of A_{app} is obvious from the above equation since different values of $(\Gamma_{\delta})_K$ will give different apparent activities. Because a number of air kerma rate constants have been published for many brachytherapy sources, 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.

3. SOURCE DATA

3.1. Photon sources

Some data for low energy photon sources used in brachytherapy applications are given in this Section. A more extensive description, including constructional details and the type of clinical application are given in [4].

TABLE II. DATA ON GAMMA EMITTING SOURCES DISCUSSED IN THIS REPORT

Isotope	Half-life (days)	Traceability to a PSDL
^{125}I	59.41	Yes
^{103}Pd	16.99	Yes
^{192}Ir	73.831	Only for LDR, none for HDR
^{137}Cs	11019.6	Yes
^{60}Co	1925.4	Yes

The column on traceability in Table II does not mean that e.g. all ^{125}I brachytherapy source calibrations are traceable to a PSDL. There is a tremendous growth in the market for new sources, in particular for low energy photon emitters, and not all of them have traceable calibrations. Currently, the only PSDL that has established primary standards for the low energy photon sources is the National Institute of Standards and Technology (NIST), USA.

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.

Because of the low energy of the photons emitted by this radioisotope, and the differences in the source constructions, it is easy to understand that the average energy of the emitted photons differs from one source to another. Clearly, it is not possible to use a common air kerma rate constant, $(\Gamma_{\delta})_K$, for these sources 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.

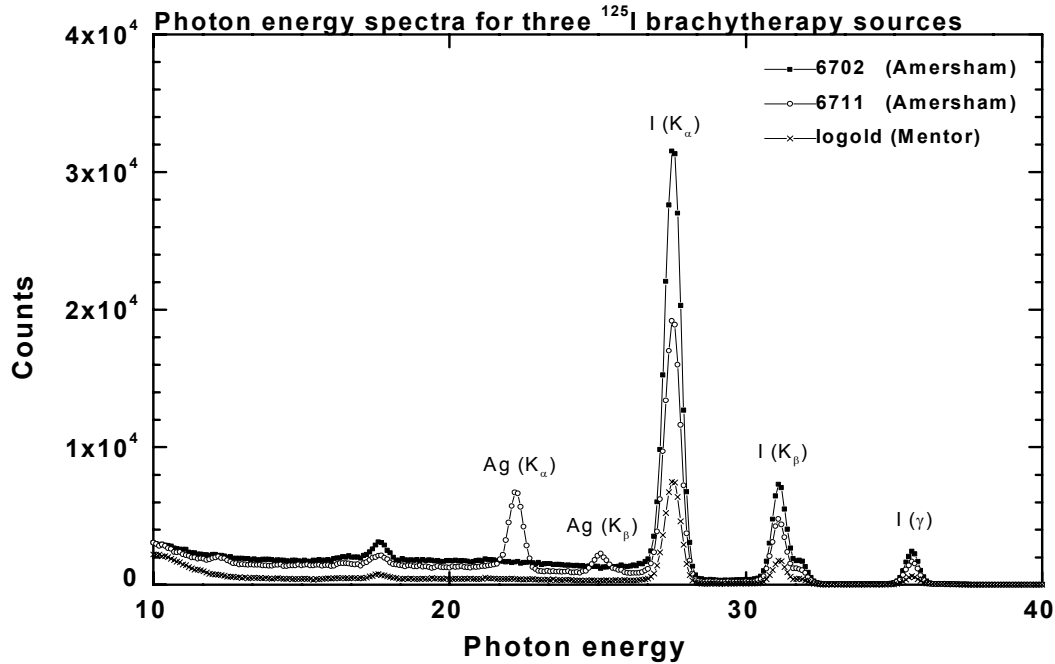


FIG.1. Photon energy spectra of three different ^{125}I sources measured at 40 cm distance with a high purity HPGe detector.

Presently there are a number of new low energy photon sources appearing on the market. It is inappropriate to apply the different constants and functions (i.e. dose rate constants, radial dose functions, anisotropy functions, anisotropy factors, geometry functions) published in the TG-43 Report [8] for currently available ^{125}I (Amersham models 6711 and 6702) and ^{103}Pd (TheraSeed 200) interstitial sources. Prior to approval of new sealed brachytherapy sources for clinical use, a PSDL calibration should be obtained and the dosimetric characteristics of the source should be determined [9]. 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 rate 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 such 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 III. For $^{188}\text{W}/^{188}\text{Re}$, $^{90}\text{Sr}/^{90}\text{Y}$ and $^{106}\text{Ru}/^{106}\text{Rh}$ 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 III. PHYSICAL DATA ON BETA RAY SOURCES

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

Clinical planar sources of $^{90}\text{Sr}/^{90}\text{Y}$ have 4 to 9 mm active diameters (10 to 13 mm physical diameters) [10]. The concave $^{90}\text{Sr}/^{90}\text{Y}$ sources have an active diameter of 6 to 18 mm with a 10 or 15 mm radius of curvature. For $^{106}\text{Ru}/^{106}\text{Rh}$, 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 Figure 2.

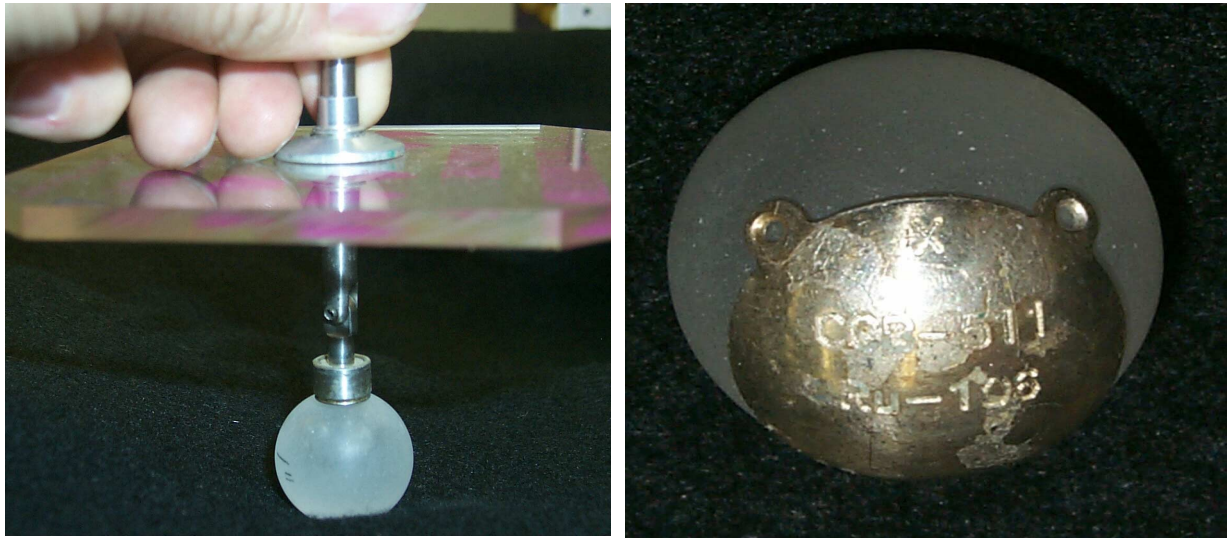


FIG. 2. Left: $^{90}\text{Sr}/^{90}\text{Y}$ eye plaque applied to a plastic eye phantom. Right: $^{106}\text{Ru}/^{106}\text{Rh}$ ophthalmic applicator with two suture holes.

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 which are inserted and removed 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 , $^{90}\text{Sr}/^{90}\text{Y}$, ^{90}Y , and $^{188}\text{W}/^{188}\text{Re}$. The physical length of these sources varies but is generally 2 to 6 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 III. 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 a concern about 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 of 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. PRIMARY STANDARDS

4.1. Reference standards

4.1.1. Standards for ^{137}Cs , ^{60}Co , ^{192}Ir (LDR): Spherical cavity chamber

The calibration of ^{137}Cs and ^{60}Co sources in terms of air kerma is usually based on measurements with a series of graphite cavity ionization chambers that serve as primary standards of exposure (later air kerma rate) [11]. At NIST (formerly National Bureau of Standards, NBS) sources of ^{137}Cs and ^{60}Co were calibrated in air at distances less than 1 m, and readings were corrected for air attenuation, build-up and room scatter. These calibrated sources were then used as working standards for the calibration of unknown sources of the same type using the replacement method with a 2.8 litre aluminium ionization chamber at distances between 0.5 and 1 m. The details of this calibration procedure as well as a list of the types of sources, which can be calibrated, are given in [12].

LDR ^{192}Ir seeds were calibrated at NIST [13] using the same graphite cavity ionization chambers as are used for the longer lived sources described above. The procedure was to use a

large collection of LDR sources to provide adequate signal for the relatively insensitive cavity chamber used. The calibrated seeds were then measured individually in a 3.44 litre aluminium re-entrant ionization chamber which transferred the calibration to this device. The working standard is thus the re-entrant chamber. This calibration was only made for two types of LDR seeds. Additional types of seeds would require an additional calibration of the type described by Loftus [13].

4.1.2. High dose rate ^{192}Ir

At the present time, primary standards for HDR ^{192}Ir sources are not available. The average energy of a ^{192}Ir brachytherapy source falls in an energy gap between the standards that have been established at PSDLs. The traceability to primary standards is maintained in this case by calibrating a suitable ion chamber in the reference fields for X radiation (ISO narrow series) [14], ^{137}Cs and ^{60}Co -radiation. The chamber is then given a calibration factor for the ^{192}Ir spectrum. This procedure as performed at the PSDL of Germany, Physikalisch-Technische Bundesanstalt (PTB) [15], comprises the evaluation of the entire calibration function of the ionization chamber between 20 keV and ^{60}Co radiation, and a subsequent interpolation for the ^{192}Ir emission lines weighted with their emission probability. As the response of the ionization chamber is determined by using X radiation instead of monoenergetic radiation, the response function obtained with X rays has to be deconvoluted to a monoenergetic function prior to the interpolation of the calibration factor. A prerequisite for the deconvolution method is the knowledge of the spectral photon distribution of the X radiation used for the calibration [15]. The uncertainty of this procedure depends strongly on the variation with photon energy of the response function of the ionization chamber used. For a 1000 cm³ LS-01 chamber the overall relative uncertainty is 1.5% (k=1). This calibration was only made for one type of HDR source; additional source types would require an additional calibration.

4.1.3. Low-energy photon sources: Wide angle free air chamber (WAFAC)

Currently only NIST can provide reference air kerma rate calibrations for ^{125}I and ^{103}Pd low energy photon sources. The calibration is accomplished with the WAFAC system developed by Loevinger [16]. In the new calibration procedure, the characteristic X rays from the Titanium⁴ capsulation are filtered out. These X rays, having energy of only 4.5 keV, do not have any effect on the dose in tissue at a typical treatment distance of about 1 cm. On the other hand, the characteristic X rays have a significant effect (approximately 10%) on the calibration signal.

Prior to January 1999, when the WAFAC system was taken into use, the effect of characteristic X rays on the calibration was not accounted for. 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 being introduced into the market. These calibrated standard sources are then transferred to SSDLs (in the USA, Accredited Dosimetry Calibration Laboratories (ADCLs)) in order to calibrate well type ionization chambers for users. The WAFAC system is reviewed in detail in Ref. [4].

⁴ The most common capsule material used in low energy photon sources.

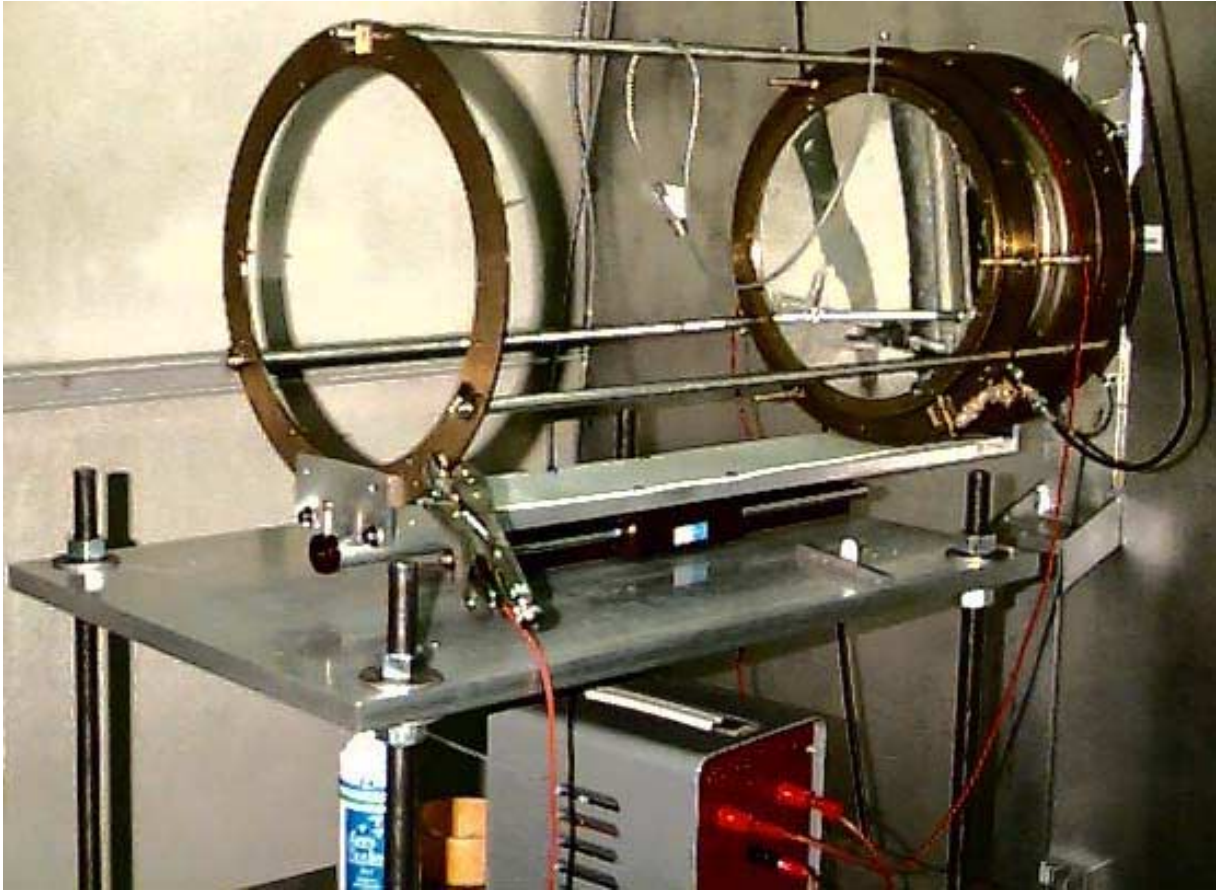


FIG. 3. Wide angle free-air chamber at NIST.

4.1.4. Beta ray sources: Extrapolation 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 Ref. [4]. By suitable construction, it can be used for all other types of beta sources except concave plaque sources.

The concave sources cannot be accurately calibrated by the extrapolation chamber due to their geometry, which does not allow the placement of 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

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

where (W/e) is the average energy in joules needed to produce one coulomb of charge of either sign in dry air ($33.97 \pm 0.05 \text{ JC}^{-1}$), 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, a is the

area of the collecting electrode, $(\Delta I / \Delta \lambda)_{\lambda \rightarrow 0}$ is the rate of change of corrected 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, a , of the collecting electrode used, because accurate 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 as a compromise between the requirement of point-like measurement and the uncertainty of the determination of the collection volume. Without correction for 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 or less. A model to account for the effects of divergence is under development [17]. A sufficient number of air gaps should be 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 Ref. [4].

The extrapolation chamber has been used at NIST to determine reference absorbed dose rate from a beta ray emitting seed or wire source [18]. For these measurements the source is inserted in a hole in a tissue-equivalent plastic block with the centre 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 unacceptably large uncertainty, approximately $\pm 7.5\%$ ($k=1$) that must be assigned to the measurement due to:

- 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 Section 8, should be used for the calibration of beta ray brachytherapy seed and wire sources.

4.2. Working standards

For routine calibrations of brachytherapy sources at the PSDLs, the complicated and time-consuming measurements by cavity chambers, WAFAC or extrapolation chambers are not always feasible. As working standards for routine calibrations, suitable calibrated detectors are applied also at the PSDL level. As described above, a working standard for ^{60}Co and ^{137}Cs sources may be a large volume ionization chamber (see 4.1.1). For ^{192}Ir and low energy photon sources, and beta particle sources used for intravascular brachytherapy, the well type ionization chamber is the accepted working standard instrument. For beta ray plaque sources, several possibilities are available as described in Section 8. The general considerations and practical guidance on measurements do not differ from those which are appropriate for calibrations at the SSDL level.

5. CALIBRATION AT THE SSDL AND HOSPITAL LEVEL

5.1. Establishment of standards for photon and intravascular sources

5.1.1. Traceability in calibrations at SSDLs

At SSDLs the recommended detector for the calibration of all brachytherapy photon sources, as well as for the calibration of intravascular beta sources, is an appropriately calibrated well type chamber. For SSDLs the preferred method for traceability in the source calibrations is to have the well type chamber calibrated against the primary standard at the PSDL. However, due to practical reasons, calibrations at an ADCL or the IAEA Dosimetry Laboratory, which are traceable to a PSDL, can be used as an alternative. This calibration should be carried out for each radionuclide and source type to be used.

5.1.2. Traceability in calibrations at hospitals

It is recommended that for brachytherapy photon and intravascular beta ray sources be calibrated with an appropriately calibrated well type chamber. For traceability, from the hospital to the SSDL, the well type chamber should be calibrated at the SSDL (or ADCL).

5.2. Maintenance of standards for photon sources and intravascular sources

For the maintenance of standards, well type chambers should be recalibrated regularly (for recommended calibration intervals, see Section 7.1.6). Recalibrations at ^{137}Cs quality can be made at the appropriate PSDL or the IAEA Dosimetry Laboratory. For regular recalibrations for all radionuclides, it may not be possible for the SSDLs to send their well type chambers to the appropriate PSDL, although this is desirable. In this case, the traceability can be maintained by recalibration of the well type chambers with long-lived (e.g. ^{137}Cs) sources and additional constancy checks at lower energies. In particular, for ^{192}Ir quality, the maintenance can be made as described below.

5.3. Maintenance of standards for ^{192}Ir quality

It was mentioned above that the SSDL's well type chamber should be recalibrated for all qualities for which it is used. However, this approach may not be possible for all SSDLs and an alternative method for ^{192}Ir is described here⁵.

The complicated energy spectrum of ^{192}Ir HDR includes about 40 energies falling approximately between 50 keV and 700 keV and with an average energy of 397 keV. Checks of the well type chambers response at the 'end' energies, i.e. 50 keV and 700 keV should be performed. If the response of the chamber does not change significantly with time at these energies, it may be concluded that the chamber's calibration factor for ^{192}Ir HDR remains unchanged. In practice, it is possible to use ^{241}Am (average energy 60 keV) and Cs-137 (average energy 661 keV) check sources for this purpose.

It is realized that the method is not of the highest metrological quality, but is the best possible with regard to the constraints that may be present for some SSDLs. It can be noted that the situation is similar to that in external beam dosimetry; ionization chambers are often calibrated for Co-60 quality only but are frequently used at other qualities.

⁵ Alternatively, the free in air measurement technique described in Section 6 can be used to calibrate an ^{192}Ir HDR source which in turn can be used to calibrate a well type chamber. This method can also be considered as a useful redundancy check of the well type chamber calibration.

The conclusions made above are supported by results obtained at the ADCL, at the University of Wisconsin, USA [19]. The well type chambers, as described in this report, have proven their stability with time. Upon calibration every 2 years with HDR ^{192}Ir sources, some types of chambers (Standard Imaging HDR 1000+) have been shown to have the same calibration factor to within $\pm 0.5\%$ [19]. Three such chambers that have been tracked over a ten-year period and have exhibited calibration factors which have remained constant to within $\pm 0.3\%$. The chamber-to-chamber variation of the ratio of HDR ^{192}Ir source calibration to ^{137}Cs and ^{60}Co has shown constancy to within $\pm 0.5\%$ [20]. For these reasons a practical solution for checking the HDR ^{192}Ir source calibration factor on a well ionization chamber is for the physicist to monitor the response of the chamber throughout its lifetime by bracketing the HDR ^{192}Ir source average energy of 397 keV with the suggested sources, ^{137}Cs and ^{241}Am . The same principle can be applied to the checking of the well type chamber calibration factor for low energy photon sources, provided a long-lived low energy check source is also used for constancy checking of the chamber. For beta-particle brachytherapy sources, constancy checks may be performed with either the photon sources, or a long lived beta-particle source (e.g. ^{90}Sr).

5.4. Establishment and maintenance of standards for planar and concave beta sources

For beta ray planar and concave sources used in ophthalmic brachytherapy, a standard may be established by the calibration of a long-lived planar source (e.g. ^{90}Sr) at the PSDL level. This calibrated source then serves as a local standard for the calibration of other devices, as outlined in Section 8. Unless there is evidence of obvious damage to the source, or changes to the output (other than by decay) the calibration need not ever be repeated at the PSDL. The substitution method should be used for the calibration of unknown planar sources. For concave sources, methods as outlined in [21] should be used.

5.5. Guidance on constancy limits for well type chambers

The stability of the output of a well type chamber should be checked at least 4 times per year. As a practical guidance on the constancy of the well type chamber's response, the following limits can be used; if the calibration factor from the ^{137}Cs re-calibrations, and the periodic constancy checks, remain the same to within $\pm 1\%$ for high-energy photon sources, or within $\pm 1.5\%$ for low-energy photon sources, or within $\pm 3\%$ for beta-particle sources, it can reasonably be assumed that the calibration factor for other sources has not changed.

If during the constancy checks of the well type chamber, it is observed that its response changes by more than the limits given above, a recalibration is recommended.

5.6. Electrometers to be used

The range of measured current in calibrating brachytherapy source is very wide depending on which type of ionization chamber is used. The following recommendations cover both well type chamber measurements and calibrations using free in-air measurement technique. A relatively high current in the nA-range characterizes well type chambers, whereas in the free in-air measurements (see Section 6), the current is usually quite low in the pA-range.

The requirements of IEC 60731 may be used as a guide to the desired characteristics of the electrometers used for these measurements. In addition they shall be capable of measuring currents as high as 200 nA for the high dose rate sources and have a signal resolution of 0.1%. For the low dose rate sources the signal resolution should be 10 fA or less; this may be

achieved for some electrometers by a charge resolution of 0.2 pC when used in the charge integration mode. It may be necessary to have two electrometers to cover the full range of brachytherapy sources to be calibrated. Caution should be exercised as some electrometers commonly used for external beam measurements can saturate at currents well below 100 nA.

6. CALIBRATION USING FREE IN-AIR MEASUREMENT TECHNIQUE

6.1. General

This section describes a method for calibrating a ‘high-energy’ photon source using a free in-air calibration technique. The method cannot be used for ^{125}I or ^{103}Pd due to the low energy of the photons emitted from these brachytherapy sources. Some reasons for the unsuitability are:

- The uncertainty in the air kerma calibration factor for an air cavity chamber at these low photon energies is unacceptably high.
- In general, low energy photon source does not have sufficient high reference air kerma rate for in air measurements. In combination with a possibly high leakage current, such measurements are subject to a large uncertainty.
- Air humidity may affect the attenuation of the low energy photons, thus affecting the measured current more than what is the case with e.g. in measurements with ^{192}Ir brachytherapy sources.

For long-lived nuclides, it is possible to maintain a source as the working standard. Subsequent re-calibration of well type ionization chambers can therefore be made using the calibrated source. This is the method of choice e.g. for ^{137}Cs LDR sources. For short-lived nuclides on the other hand, maintenance of a source as a standard is not possible and frequent calibrations of new sources are therefore necessary. The aim of the free in-air calibration of a brachytherapy source is to calibrate a source, which in turn can be used to calibrate a well type chamber. Another aim may be as a redundancy check on the calibration of a well type ionization chamber.

6.2. Formalism for reference air kerma rate

The reference air kerma rate is a quantity specified at the distance of 1 m. The direct measurement at 1 m, however, is not always practical due to low signals and the possible high leakage currents of the ionization chambers used. The reference air kerma rate, K_R , may be determined from measurements made free in-air using the equation:

$$K_R = N_K \cdot (M_u/t) \cdot k_{\text{air}} \cdot k_{\text{scatt}} \cdot k_n \cdot (d/d_{\text{ref}})^2 \quad (3)$$

where

- N_K is the air kerma calibration factor of the ionization chamber at the actual photon energy;

- M_u is the measured charge collected during the time t and corrected for ambient temperature and pressure, recombination losses and transit effects during source transfer in the case of afterloading systems;
- k_{air} is the correction for attenuation of the primary photons by the air between the source and the chamber;
- k_{scatt} is the correction for scattered radiation from the walls, floor, measurement set-up, air, etc.;
- k_n is the non-uniformity correction factor, accounting for the non-uniform electron fluence within the air cavity;
- d is the measurement distance i.e. the distance between the centre of the source and the centre of the ionization chamber;
- d_{ref} is the reference distance of 1 m.

It should be noted that the equation (3) yields the reference air kerma rate at the day of measurement. If the reference air kerma rate at another day is required, an additional correction for the source decay is necessary.

6.3. Ionization chambers to be used

For HDR sources, ionization chambers with volumes greater than 0.5 cm^3 can be used (e.g. Baldwin-Farmer 0.6 cm^3 chamber). For LDR sources, ionization chambers of higher volumes, up to about 1000 cm^3 may be needed to obtain sufficient signal. For very large chambers, the uncertainty of the non-uniformity correction factor increases [22] making the use of the chamber non-feasible. For ^{192}Ir calibrations, it is recommended to use chambers that have a variation of the air-kerma calibration factor of less than 5% between ^{60}Co and 60 keV.

6.4. Air kerma calibration of ionization chambers

Converting the ionization chamber reading, M_u , in Equation (3), to the reference air kerma rate requires the chamber to be calibrated in terms of air kerma at the actual photon energy of the brachytherapy source. For ^{60}Co and ^{137}Cs source calibrations, the calibration is done in external photon beams at these qualities.

The calibration of the ionization chamber for ^{192}Ir is less straightforward because none of the PSDLs has established standards for the use of thimble ionization chambers for ^{192}Ir HDR and only few have standards for some ^{192}Ir LDR sources. It is therefore necessary to obtain the air kerma calibration factor for the ionization chamber using an indirect method. Using the interpolation method based on the determination of the response function of the chamber, as described in Section 4.1.2, one PSDL (PTB) can provide a calibration factor for HDR ^{192}Ir . Another option for the SSDL is a simpler method based on the technique developed by Goetsch et al. [23]. It was originally developed for calibration of ionization chambers for subsequent use in ^{192}Ir HDR dosimetry, but can also be used in ^{192}Ir LDR source calibrations.

The principle proposed by Goetsch is to calibrate the chamber at an appropriate X ray quality and at ^{137}Cs , or at ^{60}Co if a ^{137}Cs beam is not available. With the knowledge of the air kerma calibration factors at these two energies, the air kerma calibration factor for ^{192}Ir is obtained

by interpolation. This method requires the total wall thickness to be the same at each calibration quality [23].

The air kerma weighted average energy of an ^{192}Ir brachytherapy source is 397 keV [24, 25]. A typical X ray beam that can be used for calibration at the SSDLs is 250 kV, added filtration of 1.0 mm Al and 1.65 mm Cu, and a HVL of 2.50 mm Cu. Beams similar to this should be used for the lower energy portion of the determination of the air kerma calibration factor. Primary laboratories can provide air-kerma calibration factors for these beam qualities and secondary standard chambers can be calibrated at these energies. Alternatively the SSDLs may opt to calibrate the secondary standard chambers at these energies themselves. In this report, the X ray beam referred to is the 250 kV beam above.

6.4.1. ^{137}Cs calibration point

The ionization chamber wall must be thick enough to block all electrons emanating from the source or capsule, and to provide charged particle equilibrium for the highest energy secondary electrons present in the ^{137}Cs beam. The required total wall thickness (inner wall and cap) needed is 0.36 g/cm^2 .

Air kerma calibration factors, N_K , for both the ^{137}Cs and X ray beam must be determined with the build up cap (equivalent wall) in place for both beams. The measured calibration factors give the air kerma per unit charge for the chamber including the attenuation of the cap. Due to the interpolation technique [23], the attenuation of the cap and scattering effects of the chamber wall must be taken into account. Thus, the factor called A-wall, A_w , is introduced.

The response of the chamber alone, N_{ch} , is given by:

$$N_{\text{ch}} = N_K A_w \quad (4)$$

The calibration factor for ^{192}Ir can then be obtained by interpolation between the N_{ch} factors for the two bracketing energies from the following equation:

$$A_{w,\text{Ir}} N_{K,\text{Ir}} = [A_{w,250\text{kV}} N_{K,250\text{kV}} + A_{w,\text{Cs}} N_{K,\text{Cs}}]/2 \quad (5)$$

where $N_{K,\text{Ir}}$, $N_{K,250\text{kV}}$ and $N_{K,\text{Cs}}$ are the air kerma calibration factors for ^{192}Ir , 250 kV X rays and ^{137}Cs qualities, respectively, and $A_{w,\text{Ir}}$, $A_{w,250\text{kV}}$ and $A_{w,\text{Cs}}$ are the corresponding A-wall factors. If $N_{K,250\text{kV}}$ and $N_{K,\text{Cs}}$ do not differ by more than 10%, which usually is the case, then the equation for $N_{K,\text{Ir}}$ can be written as [23]:

$$N_{K,\text{Ir}} = (1 + x) [N_{K,250\text{kV}} + N_{K,\text{Cs}}]/2 \quad (6)$$

where $x = 0.0037 \cdot (t/9.3 \cdot 10^{22})$ for a wall thickness of t electrons/cm².

If a total wall thickness of 0.36 g/cm^2 is not available, a ^{60}Co build up cap can be used instead. The difference in the $N_{K,\text{Ir}}$ calibration factor using these two different wall thickness is about 0.5%.

6.4.2. ^{60}Co calibration point

In the event that there is no ^{137}Cs beam energy at the SSDL, a ^{60}Co beam may be used as the high energy point using the appropriate build up cap and wall thickness for ^{60}Co , 0.5 g/cm^2 .

This thickness must be used also in the calibration in the 250 kV X ray beam. The method for determination of the $N_{K, Ir}$ calibration factor is similar to that described above except that the relative weighting of the air kerma calibration factors is different.

The weighted interpolation factors are given by the following equations:

$$f_{w, 250kV} = \left| \frac{\overline{h\nu}_{Ir} - \overline{h\nu}_{Co}}{\overline{h\nu}_{Co} - \overline{h\nu}_{250kV}} \right| = 0.8 \quad \text{and} \quad f_{w, Co} = \left| \frac{\overline{h\nu}_{Ir} - \overline{h\nu}_{250kV}}{\overline{h\nu}_{Co} - \overline{h\nu}_{250kV}} \right| = 0.2 \quad (7)$$

where $\overline{h\nu}_{Ir}$ and $\overline{h\nu}_{Co}$ are the air kerma weighted average energies of ^{192}Ir gamma rays and ^{60}Co gamma rays, respectively, and $\overline{h\nu}_{250kV}$ represents the effective energy⁶ (131 keV) of the 250kV X ray beam. This results in the following equation for $N_{K, Ir}$ with the weighted air kerma values [26]:

$$N_{K, Ir} = (0.8 \cdot A_{w, 250kV} N_{K, 250kV} + 0.2 \cdot A_{w, Co} N_{K, Co}) / A_{w, Ir} \quad (8)$$

Table IV includes A_w factors for different ionization chambers. If the chamber in use is not listed in the table, then A_w can be set to 1.000 for each energy in Equation (8), and the calibration factor is determined with

$$N_{K, Ir} = 0.8 \cdot N_{K, 250kV} + 0.2 \cdot N_{K, Co} \quad (9)$$

In particular, for ^{192}Ir LDR source calibrations using the free in-air measurement technique, a large volume ionisation chamber must be used. Such chambers are not included in Table IV and consequently, equation (9) is to be used in such cases.

With the use of Equation (9), the uncertainty in the air kerma calibration factor for ^{192}Ir increases by approximately 0.5%.

The interpolation method based on the response function of the chamber, as applied at PTB and described in Section 4.1.2 is an inherently more accurate method than the above method of Goetsch since no assumptions are made as to the response function of the chamber but rather it is measured and calculated for the energies of each line in the ^{192}Ir spectrum. Thus the uncertainties using the Goetsch method will be somewhat higher depending on the chamber type; there are insufficient studies to estimate the differences in uncertainty.

6.5. Correction factors for free in-air measurements

To obtain the reference air kerma rate with the least possible uncertainty necessitates a cautious performance of the free in-air measurements and the use of up-to-date correction factors. In this section the various correction factors are discussed in detail, for reference air kerma rate determination of ^{192}Ir LDR and HDR sources, and ^{137}Cs and ^{60}Co sources.

⁶ Strictly speaking, effective energy is defined only for narrow spectra and is the monoenergetic energy which has the same attenuation coefficient as the narrow spectrum.

TABLE IV. MONTE-CARLO CALCULATED A_w FACTORS FOR DIFFERENT IONIZATION CHAMBERS FOR 250 kV X RAY, ^{192}Ir AND ^{60}Co . THE UNCERTAINTIES (ONE STANDARD DEVIATION) ARE $< 0.1\%$ VALUES FROM [27]

Ionization chamber	Air cavity length/radius (mm)	Wall material/thickness gcm^{-2}	Build-up cap material/thickness gcm^{-2}	A_w 250kV	A_w ^{192}Ir	A_w ^{60}Co
Capintec 0.65 cm ³ PR-06C Farmer	22.3 / 3.2	C552 / 0.050	C552 / 0.924	0.998	0.980	0.984
Capintec 0.65 cm ³ PR-06C Farmer	22.3 / 3.2	C552 / 0.050	Polyst. / 0.537	0.997	0.986	0.990
Capintec 0.65 cm ³ PR-06C Farmer	22.3 / 3.2	C552 / 0.050	PMMA / 0.547	0.992	0.984	0.989
Capintec 0.6 cm ³ PR-05P AAPM	23.8 / 3.3	Graphite / 0.046	PMMA / 0.625	0.995	0.986	0.986
Exradin 0.5 cm ³ A2 (2mm cap)	11.4 / 4.8	C552 / 0.176	C552 / 0.352	0.986	0.978	0.984
Exradin 0.5 cm ³ A2 (4mm cap)	11.4 / 4.8	C552 / 0.176	C552 / 0.712	0.989	0.973	0.976
Exradin 0.5 cm ³ P2 (4mm cap)	11.4 / 4.8	Polyst./0.105	Polyst. / 0.420	0.986	0.982	0.988
Exradin 0.5 cm ³ T2 (4mm cap)	11.4 / 4.8	A150 / 0.113	A150 / 0.455	0.983	0.979	0.985
Exradin 0.65 cm ³ Farmer A 12	24.2 / 3.1	C552 / 0.088	C552 / 0.493	0.999	0.988	0.991
NE 0.6 cm ³ Farmer 2505	24 / 3.0	Tufnol / 0.075	PMMA / 0.545	0.997	0.989	0.990
NE 0.6 cm ³ Farmer 2505/A	24 / 3.0	Nylon 66 / 0.063	PMMA / 0.545	0.996	0.984	0.989
NE 0.6 cm ³ Farmer 2505/3A	24 / 3.2	Graphite / 0.065	PMMA / 0.551	0.998	0.989	0.989
NE 0.6 cm ³ Farmer 2505/3B	24 / 3.2	Nylon 66/0.041	PMMA / 0.551	0.995	0.990	0.989
NE 0.6 cm ³ Farmer 2571	24.1 / 3.15	Graphite / 0.065	Delrin / 0.551	0.999	0.989	0.988
NE 0.6 cm ³ Farmer 2571	24.1 / 3.15	Graphite / 0.065	PMMA / 0.550	0.998	0.989	0.989
NE 0.6 cm ³ Farmer 2581	24.1 / 3.2	A150 / 0.040	PMMA / 0.584	0.986	0.988	0.987
NE 0.6 cm ³ Farmer 2581	24.1 / 3.2	A150 / 0.041	Polyst. / 0.584	0.991	0.990	0.991
PTW 1.0 cm ³ 23 331 rigid	22 / 3.95	PMMA / 0.060	PMMA / 0.345	0.997	0.992	0.993
PTW 0.6 cm ³ Farmer 30 001	23 / 3.05	PMMA / 0.045	PMMA / 0.541	0.997	0.990	0.990
PTW 0.6 cm ³ Farmer 30 002	23 / 3.05	Graphite / 0.079	PMMA / 0.541	0.993	0.989	0.989

Uncollimated brachytherapy sources are typically measured at distances which are shorter than those used in the calibration of collimated teletherapy beams. At these short distances, air kerma measurements are extremely sensitive to positional uncertainties. Therefore, the calibration requires some device (a jig) of low-density plastic to hold the chamber and the source in precise position during the calibration. Any such mounting device unavoidably compromises between the need for mechanical rigidity and the desire to minimize scatter. While corrections for scatter can be determined, they should be kept at a minimum. Both of these issues contribute a major part to the overall calibration uncertainty.

6.5.1. Measurement distances

Increasing the distance decreases the uncertainty in the calibration distance and the effect of the finite size of the ionization chamber. However, this improvement results in a reduced signal and an increased relative importance of room and equipment scatter. There are four effects that contribute to the uncertainties in calibration of brachytherapy sources using an ionization chamber. These effects expressed as a function of distance between the source and the chamber (SCD) are:

- Uncertainty due to the effects of the chamber size. The uncertainty in the non-uniformity correction factor decreases with increasing SCD;
- scatter, which as a percentage of the total signal increases with increasing SCD;
- positional uncertainty, which follows the inverse square law and thus decreases with increasing SCD;
- leakage current relative to the ionization reading, the effect of which increases with increasing SCD.

The measurement distance should be selected so that the combined uncertainty due to the above effects will be minimized. This would generally be the distance where the various correction factors, when combined in quadrature, have a minimum value. For a combination of ^{192}Ir HDR source and a Farmer-type chamber, the optimum distance has been shown to be 16 cm [28]. With the possible exception of the scattered radiation, it can be noted that the different contributions listed above have only minor energy dependence. Thus, the optimum distance for ^{60}Co and ^{137}Cs HDR source calibrations should be approximately the same as that for a ^{192}Ir HDR source. It must be pointed out that the non-uniformity correction factors used in this report are calculated assuming point source geometry. Thus, in all free in-air measurements, in HDR as well as LDR, the distances used must be large enough so that the source can be considered as a point source. Furthermore, the inclusion of the inverse square relation in Equation (3) implies that sufficiently large distances must be used. A practical criteria is that the distance between the chamber centre and the centre of the source must be at least 10 times the length of the source in order to ensure that the error introduced due to the point source approximation is less than 0.1%.

It is recommended in this report that measurements should be made at multiple distances and the reference air kerma rate should be determined from the measurements made at each distance. This procedure will give redundancy and large variations in the K_R , as determined from the different measurements, are indications of bad experimental conditions. For HDR source calibrations, the measurement distances can be selected around the optimum distance (e.g. between 10 cm and 40 cm). For LDR sources, with the use of large volume ionization chambers, measurement distances between 50 cm and 100 cm are appropriate.

6.5.2. The scatter correction factor

To maintain the contribution of scattered radiation at a minimum, the source and chamber should be placed in the centre of the room and well above the floor (at least 1 m from any wall or floor). All measurements should preferably be carried out using the same jig position within the room.

Two methods have been used to determine the scatter correction: the multiple distance method [23] and the shadow shield method [22, 25, 29]. In the former method, the air kerma rate due to scattered radiation is assumed to be constant over the measurement distances.

In the shadow shield method, a cone of a high Z material is placed between the source and the chamber in order to prevent the primary photons from reaching the chamber. The ratio of the measured charge with and without the shield in place can be used to calculate the scatter correction factor. The height of the cone must be large enough to provide sufficient attenuation and should not be placed too close to the chamber due to possible scattering from

the cone. Therefore, the multiple distance method is recommended for measurements which involve short distances, and in particular in ^{137}Cs and ^{60}Co HDR source calibrations.

In the multiple distance method, readings are made at a series of distances with carefully measured separations. If a linear scanner is used, the readings are to be taken by scanning continuously in one direction through the sequential distances in order to avoid any backlash errors in the scanner mechanism. The readings made at the different distances reflect the inverse square law differences between them, and an assumed constant amount of scatter.

It is essential in this method that the changes in distance be precise and accurate, in order to derive the correction c that yields the “true” centre-to-centre source to chamber distances, d' . The distance for a reading is expressed by the following equation:

$$d' = d + c \quad (10)$$

where

- d' is the center-to-center source chamber distance accounting for the offset c in the distance;
- d is the apparent center-to-center source chamber distance;
- c is the offset in the set-up distance (c can be positive or negative).

The contribution of scatter radiation to the air kerma rate, K_s , is included in the measured air kerma rate, $K(d')$. Therefore the air kerma value due to the primary photons only, $K_p(d')$, is given by

$$K_p(d') = K(d') - K_s \quad (11)$$

Combining the Equations (10) and (11) yields:

$$K_p(d') = (K(d') - K_s) (d+c)^2/(d')^2 \quad (12)$$

for any distance. The air kerma due to the primary photons varies as the inverse of the square of the distance, and therefore, measurements at three distances can be used to determine the three unknowns, $K_p(d')$, K_s , and c . For redundancy preferably five or seven distances, e.g. in HDR brachytherapy source calibrations 10, 15, 20, 25, 30, 35 and 40 cm should be used. The seven distances redundantly determine the scatter contribution and factor c since there are 3 unknowns with 35 solutions. A computer generated solution then can be used to average the solutions. Thus, the scatter correction k_{scatt} can be determined as follows:

$$k_{\text{scatt}} = 1 - K_s/K(d') = 1 - K_s/(N_K \cdot M_u \cdot k_n) \quad (13)$$

where the measured charge M_u has been corrected for ambient conditions. The determined values of c should not vary by more than ± 1 mm. If there is a large variation in c values when the redundant solutions are made, it is indicative of an error made in the measurement process. In such cases, the entire process should be carefully reviewed and the measurements repeated.

The shadow shield method has mainly been used to determine the scatter correction factor at a distance of 1 m. Table V shows the results of a few experimental determinations of the scatter correction using the shadow shield method. The results suggest that the room size may not be critical for this factor.

TABLE V. SCATTER CORRECTION FACTORS DETERMINED WITH THE SHADOW SHIELD METHOD AT 1 m DISTANCE FROM AN ^{192}Ir SOURCE

Author		k_{scatt}	Chambers	Room size m × m × m
Verhaegen et al.	[22]	0.940	NE 2551 and Exradin A6	4 × 4 × 4
Verhaegen et al.	[22]	0.975	PTW LS-10	4 × 4 × 4
Petersen et al.	[30]	0.940	Exradin A5	6 × 6 × 3
Drugge	[25]	0.940	Exradin A5 and NE 2530/1	3.5 × 5 × 3.5
Piermattei et al.	[29]	0.928	Exradin A4	—
Piermattei et al.	[29]	0.941	Exradin A6	—

In ^{192}Ir dosimetry it has been shown that the scatter correction factors obtained with the two methods are in a good agreement [22, 25].

6.5.3. The non-uniformity correction factor

In the measurements of brachytherapy sources free in-air, the non-collimated geometry, with high divergence of the incident photons, differs from the geometry of the collimated photon beams such as those external beams used for calibrating the chamber. There will be a marked variation in the photon fluence over the different parts of the chamber.

The electrons entering the air cavity are mainly generated in the inner wall of the chamber. Due to the non-uniform photon fluence over the wall, the generation of electrons from the wall varies significantly from place to place in the wall. The net result of this is a non-uniform electron fluence in the air cavity of the chamber.

In order to take into account this non-uniformity, to convert the measured charge or current into air kerma rate at the measurement distance, it is necessary to apply a non-uniformity correction factor, k_n . This factor depends on the

- shape and dimensions of the ionization chamber (spherical, cylindrical, internal radius and length);
- measurement distance and the source geometry ('point source', line source, etc.);
- material in the inner wall of the chamber;
- energy of the photons emitted from the source.

The most widely used non-uniformity correction factors are those given by Kondo and Randolph [32]. In their theory, the electron fluence in the air cavity of the ionization chamber is assumed to be isotropic. The theory was later extended by Bielajew [31] who included a more realistic angular distribution of electron fluence in the air cavity of the chamber. In contrast to the isotropic theory, this anisotropic theory predicts the wall material and a photon energy dependence in the non-uniformity correction factor. The relationship between the two theories is given by

$$A_{\text{pn}}(d) = A_{\text{pn}}^{\text{KR}}(d) + \omega A'_{\text{pn}}(d) \quad (14)$$

where $1/A_{\text{pn}}^{\text{KR}}(d)$ is the non-uniformity correction factor obtained from the isotropic theory of Kondo and Randolph and $1/A_{\text{pn}}(d)$ is the non-uniformity correction factor according to the anisotropic theory of Bielajew. $A'_{\text{pn}}(d)$ takes into account the anisotropic electron fluence within the air cavity and the degree of anisotropy is given by the energy and material

dependent factor ω . Thus, the theory by Bielajew predicts an energy and inner wall material dependence in the non-uniformity correction factor. In contrast, the theory by Kondo and Randolph is independent of both these factors.

It is recommended in this report that the factor $1/A_{pn}(d)$ according to the theory by Bielajew be used for determination of k_n . Thus,

$$k_n = 1/A_{pn}(d) \quad (15)$$

For cylindrical ionization chambers, it has been shown that the non-uniformity correction factor obtained with the anisotropic theory is, for commonly used chamber wall materials, quite insensitive to the ω -values [33]. Table VI gives values of ω for some commonly used inner wall materials. For materials that are not included in the table, a good approximation is to use the value for that material with similar dosimetric properties as that listed in Table VI. For example, the ω value for C552 plastic can be taken to be the same as that for graphite, i.e. 0.992. It should be noted that the wall material referred to is the material in the inner wall of the ionization chamber, not the material in the build up cap.

TABLE VI. MATERIAL- AND PHOTON ENERGY DEPENDENT FACTORS, ω

Inner wall material	ω
A-150	1.066
PMMA	1.014
Graphite	0.992

The values in Table VI were calculated for an unfiltered ^{192}Ir source. As shown for graphite (the inner wall material of an NE2571 chamber) in Figure 4, the non-uniformity correction factor has only minor energy dependence. Other wall materials listed in Table VI show similar behaviour. Without loss of accuracy, these values can therefore be used in ^{137}Cs and ^{60}Co calibrations.

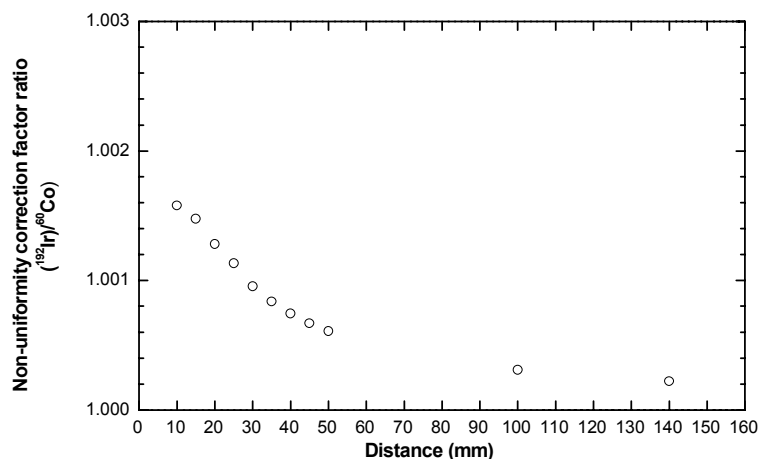


FIG. 4. Ratio of non-uniformity correction factor for an NE2571 ionization chamber at ^{192}Ir and ^{60}Co qualities.

The parameters, $A_{pn}^{KR}(d)$ and $A'_{pn}(d)$, for the calculation of the non-uniformity correction factor for cylindrical chambers are given in Tables VII and VIII as a function of the cylindrical chamber's shape factor, $\sigma=R_c/L_c$, and the distance factor, $\alpha=R_c/d$. In these formulas, R_c is the chamber's internal radius, L_c is the internal half-length of the chamber and d is the measurement distance.

TABLE VII. VALUES OF FACTORS $A_{pn}^{KR}(d)$ FOR CYLINDRICAL IONIZATION CHAMBERS. R_c AND L_c ARE THE CHAMBERS' INTERNAL RADIUS AND HALF-LENGTH

Distance factor $\alpha=R_c/d$	Shape factor $\sigma=R_c/L_c$								
	0.05	0.10	0.25	0.50	0.70	0.80	1.00	2.00	4.00
0.000	1.0000	1.0000	1.0000	1.0000	1.0000	1.0000	1.0000	1.0000	1.0000
0.005	0.9967	0.9992	0.9999	1.0000	1.0000	1.0000	1.0000	1.0000	1.0000
0.010	0.9869	0.9967	0.9995	0.9999	1.0000	1.0000	1.0000	1.0001	1.0001
0.050	0.7854	0.9273	0.9878	0.9980	0.9998	1.0003	1.0008	1.0015	1.0015
0.100	0.5546	0.7863	0.9541	0.9921	0.9992	1.0010	1.0031	1.0059	1.0061
0.200	0.3349	0.5586	0.8524	0.9694	0.9963	1.0035	1.0123	1.0238	1.0250
0.300	0.2401	0.4263	0.7476	0.9359	0.9908	1.0067	1.0268	1.0551	1.0586
0.400	0.1892	0.3468	0.6615	0.8980	0.9831	1.0099	1.0460	1.1019	1.1103
0.500	0.1584	0.2960	0.5966	0.8629	0.9755	1.0142	1.0698	1.1676	1.1864
0.600	0.1388	0.2628	0.5508	0.8370	0.9732	1.0235	1.1002	1.2576	1.2985
0.700	0.1266	0.2421	0.5226	0.8263	0.9842	1.0457	1.1443	1.3809	1.4681
0.800	0.1206	0.2326	0.5146	0.8416	1.0233	1.0971	1.2200	1.5592	1.7406
0.900	0.1235	0.2398	0.5429	0.9166	1.1364	1.2284	1.3864	1.8736	2.2432

TABLE VIII. VALUES OF FACTORS $A'_{pn}(d)$ FOR CYLINDRICAL IONIZATION CHAMBERS. R_c AND L_c ARE THE CHAMBERS' INTERNAL RADIUS AND HALF-LENGTH

Distance factor $\alpha=R_c/d$	Shape factor $\sigma=R_c/L_c$								
	0.05	0.10	0.25	0.50	0.70	0.80	1.00	2.00	4.00
0.000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000
0.005	-0.0014	-0.0012	-0.0009	-0.0005	-0.0003	-0.0002	-0.0001	0.0002	0.0004
0.010	-0.0027	-0.0024	-0.0017	-0.0009	-0.0005	-0.0004	-0.0001	0.0005	0.0007
0.050	-0.0056	-0.0093	-0.0083	-0.0047	-0.0027	-0.0019	-0.0007	0.0024	0.0036
0.100	-0.0032	-0.0103	-0.0148	-0.0093	-0.0055	-0.0039	-0.0014	0.0047	0.0072
0.200	-0.0011	-0.0062	-0.0203	-0.0179	-0.0115	-0.0086	-0.0036	0.0093	0.0147
0.300	-0.0006	-0.0036	-0.0190	-0.0242	-0.0180	-0.0143	-0.0071	0.0136	0.0229
0.400	-0.0003	-0.0023	-0.0159	-0.0274	-0.0241	-0.0205	-0.0122	0.0173	0.0323
0.500	-0.0002	-0.0016	-0.0130	-0.0279	-0.0285	-0.0261	-0.0186	0.0194	0.0433
0.600	-0.0002	-0.0012	-0.0106	-0.0267	-0.0309	-0.0302	-0.0250	0.0188	0.0563
0.700	-0.0001	-0.0009	-0.0088	-0.0247	-0.0314	-0.0324	-0.0303	0.0138	0.0712
0.800	-0.0001	-0.0007	-0.0073	-0.0224	-0.0306	-0.0328	-0.0338	0.0036	0.0851
0.900	-0.0001	-0.0006	-0.0062	-0.0202	-0.0290	-0.0321	-0.0354	-0.0100	0.0869

The anisotropic non-uniformity correction factors for Farmer-type chambers (internal length 24.1 mm, internal radius 3.15 mm, e.g. NE2571, NE2581) at different distances from the source are given in Table IX [34]. For the calculation of the factors in Table IX, the cone (a deviation from cylindrical geometry) at the tip of the chamber has been taken into consideration, resulting in values which are slightly different from those that would be derived from Tables VII and VIII otherwise.

TABLE IX. NON-UNIFORMITY CORRECTION FACTORS FOR FARMER-TYPE IONIZATION CHAMBERS (INTERNAL RADIUS 3.15 mm, LENGTH 24.1 mm)

Distance (cm)	k_n
10.0	1.009
15.0	1.005
20.0	1.004
25.0	1.003
30.0	1.002
40.0	1.002
50.0	1.001

For spherical ionization chambers, $\omega = 0$, and the non-uniformity correction factors given by Kondo and Randolph can be directly applied. The $A_{pn}(d)$ factors for spherical chambers are reproduced in Table X.

TABLE X. $A_{pn}(d)$ FACTORS FOR SPHERICAL IONIZATION CHAMBERS

Distance (cm)	Chamber radius (cm)									
	2.0	2.5	3.0	3.5	4.0	4.5	5.0	5.5	6.0	6.5
10.0	1.014	1.022	1.032	1.044	–	–	–	–	–	–
15.0	1.006	1.009	1.014	1.019	1.025	1.032	1.040	1.049	–	–
20.0	1.003	1.005	1.008	1.010	1.014	1.017	1.022	1.026	1.032	1.038
25.0	1.002	1.003	1.005	1.007	1.009	1.011	1.014	1.017	1.020	1.023
30.0	1.001	1.002	1.003	1.005	1.006	1.008	1.009	1.011	1.014	1.016
35.0	1.001	1.002	1.002	1.003	1.004	1.006	1.007	1.008	1.010	1.012
40.0	1.001	1.001	1.002	1.003	1.003	1.004	1.005	1.006	1.008	1.009
45.0	1.001	1.001	1.001	1.002	1.003	1.003	1.004	1.005	1.006	1.007
50.0	1.001	1.001	1.001	1.002	1.002	1.003	1.003	1.004	1.005	1.006
60.0	1.000	1.001	1.001	1.001	1.001	1.002	1.002	1.003	1.003	1.004
70.0	1.000	1.000	1.001	1.001	1.001	1.001	1.002	1.002	1.002	1.003
80.0	1.000	1.000	1.000	1.001	1.001	1.001	1.001	1.002	1.002	1.002
90.0	1.000	1.000	1.000	1.001	1.001	1.001	1.001	1.001	1.001	1.002
100.0	1.000	1.000	1.000	1.000	1.001	1.001	1.001	1.001	1.001	1.001

6.5.3. Correction for the attenuation of primary photons in air

For determination of the reference air kerma rate from the measured air kerma at the distance d , it is necessary to correct for the attenuation of the primary photons between the source and the ionization chamber. Table XI gives the k_{air} correction factors at different distances between the source and the ionization chamber [22, 25, 35].

TABLE XI. CORRECTION FACTORS FOR AIR ATTENUATION OF THE PRIMARY PHOTONS FROM ^{192}Ir , ^{137}Cs AND ^{60}Co BRACHYTHERAPY SOURCES

Distance (cm)	^{192}Ir	^{137}Cs	^{60}Co
10	1.001	1.000	1.000
20	1.002	1.000	1.000
30	1.004	1.001	1.000
40	1.005	1.001	1.000
50	1.006	1.001	1.000
60	1.007	1.001	1.000
70	1.009	1.002	1.000
80	1.010	1.002	1.000
90	1.011	1.002	1.000
100	1.012	1.002	1.000

6.5.4. Correction for transit effects, leakage current and recombination losses

While the source moves into the measurement position, and then away after the measurement, the detector measures a signal, referred to as the transit signal. This transit signal acts the same as the shutter effect of a ^{60}Co teletherapy unit. The magnitude strongly depends on the source-to-detector distance, and is significant at the distances used in calibration. Several techniques can be used to eliminate the transit component of the signal:

- Using an externally-triggered electrometer to collect charge during an interval after the source has stopped moving [23].
- Subtracting two readings taken for differing intervals to eliminate the transit charge common to both.
- Using a current reading after the source has stopped moving (if the signal is large enough).

The importance of electrical leakage currents in the individual dosimetry system should be evaluated since the signal levels measured during calibration are typically 50 to 100 times less than usually encountered in teletherapy measurements. This can be significant for most thimble or Farmer type ionization chambers. Larger volume spherical ionization chambers do not have this effect to a great extent. Generally if the leakage is greater than 0.1% of the signal, it should be taken into account.

A correction is also needed for the recombination losses and for the ambient temperature and pressure [36].

6.6. Uncertainty of free in-air calibration

The estimated relative uncertainty in a calibration of a HDR ^{192}Ir source using the seven-distance technique is typically 1.5% ($k=1$) [19]. This includes the 0.5% (^{137}Cs external beam) and 1% (X ray) contributions of the PSDL air-kerma calibrations of the chamber used for the free in-air measurements.

6.6.1. Traceability of free in-air calibration

The traceability of the calibration is to the PSDL that performs the calibrations of the chamber used for the free in-air measurements. Due to the lack of primary standards, it must be emphasized that for ^{192}Ir HDR calibrations, the traceability is not as robust as that e.g. in the ^{137}Cs or ^{60}Co brachytherapy source calibrations.

7. CALIBRATIONS USING WELL TYPE IONIZATION CHAMBERS

7.1. General guidance

7.1.1. *Well type chambers, electrometers and reference sources*

The well type chamber for brachytherapy source calibrations should be of the type designed specially for radiotherapy applications, capable of measuring the reference air kerma rate of both LDR and HDR sources. It is recommended that only well type chambers which are open to the atmosphere be used. If the chamber is sealed and the pressure of the gas is at a higher level than the ambient atmospheric pressure, it may develop a problem of slow leakage of the gas. In this case, a change in the calibration factor would result. Chambers open to the atmosphere need correction for temperature and pressure since the calibration factor is based upon a density of air corresponding to standard ambient conditions, usually 20°C and 101.3 kPa.

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

- The chambers measure only in units of activity.
- The chambers have settings for given radionuclides but not for 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 a significant 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.

It should be noted that the well type chamber and the electrometer can have independent calibration factors. If this is the case, the calibration factors must be multiplied together to form the total calibration factor of the well type chamber and electrometer system. Unless the calibration factor for the whole system of the SSDL has been provided by the calibration laboratory (e.g., the IAEA), the calibration factor of the electrometer must be determined separately by the SSDL, e.g. by comparison with other electrometers using a calibrated constant current source.

The equipment used at the IAEA Dosimetry Laboratory is described in Appendix A.

7.1.2. *Calibration point inside the well type chamber*

The calibration point of a well type chamber is defined as the point at which the centre of the source is positioned during the calibration procedure; this point may differ from one source to another depending upon the source length. Some chambers have a fixed, non-removable, spacer in the well and the source is then conveniently placed on the top of the spacer. Other models, on the other hand, have a mechanism to move and fix the source holder to different heights and the source is then placed at the bottom of the movable holder during the

calibration procedures. The location of the calibration point must be stated on the chamber's calibration certificate. Possible spacers and the outer dimensions of the source used to calibrate the chamber must also be stated in the calibration certificate. Spacers should be designated such that there is no possibility for confusion.

For chambers that are identical to the IAEA standard well type chamber, the calibration point is with the source at the position of maximum response. With the source positioned at this point, the uncertainty in the reference air kerma rate determination, due to positional uncertainty, is minimized. This position is dependent on the source type and must be determined prior to the calibration. Measurements are performed at different positions of the source along the axis of the chamber by inserting spacers of known length at the bottom.

7.1.3. Measurement techniques

All measurements should be done in a minimum scatter environment, with the chamber at least 1 m from any wall or floor. The chamber should be left to come to equilibrium with its surroundings before beginning calibration. The minimal time necessary for this is 30 minutes. Care should be taken that the temperature measured is that of the chamber volume and not the room temperature. A minimum of 4 significant digits should always be obtained for charge accumulated or current measurements. Thus charge should be accumulated for a set time depending on the activity of the source. A minimum of 5 measurements for each source insertion that are neither monotonically increasing nor decreasing should be obtained, and at least two source insertions should be made. For HDR sources these measurements should be within 0.3% of the average reading and the average of two sets of readings should be within 0.5%.

For beta-particle sources, measurements should be made at various orientations of the source about its cylindrical axis and the results averaged. Multiple insertions of the sources should also be made for loose seed trains delivered in a catheter. Since there is a polarity effect for beta particles, the calibration factor is only valid for the polarity stated in the calibration certificate.

7.1.4. Measurement corrections

Recombination correction, k_{recom} , may be determined using the two-voltage technique. If the ratio of the voltage used in this technique is exactly 2 (e.g. if 150V and 300V are used as is often the case with well type chambers) then the recombination correction can be determined from [37]

$$1/k_{\text{recom}} = 4/3 - [Q_1/(3 \cdot Q_2)]$$

where Q_1 is the charge collected at the higher voltage (i.e. at 300V) and Q_2 at the lower voltage (i.e. 150V).

Good quality chambers generally exhibit negligible recombination effect for brachytherapy sources.

Air density corrections (temperature and pressure) are calculated according to

$$k_{Tp} = \frac{(273.15 + T)}{(273.15 + T_0)} \cdot \frac{101.3}{p}$$

where T is the temperature in Celsius and p is the pressure in kPa and T₀ is the reference temperature at calibration (usually 20°C).

If the electrometer has been calibrated separately, the calibration factor of the electrometer, N_{elec}, must be applied

7.1.5. Calculation of well type chamber calibration factor

If the electrometer has been calibrated separately, the reference air kerma rate calibration factor of the well type chamber, N_{K_R}, is determined from,

$$N_{K_R} = \frac{K_R}{(M_u k_{Tp} k_{recom} N_{elec})} \quad (16)$$

where

K_R is the reference air kerma rate of the source and M_u is the scale unit reading, and k_{Tp}, k_{recom} and N_{elec} are corrections for the temperature and pressure, recombination losses and the electrometer calibration factor, respectively.

If the chamber and electrometer are calibrated as a system, then N_{elec} is considered to be unity.

7.1.6. Quality control of well type chamber measurements

As with any other type of chamber calibration, measurements with long-lived check sources should be made before and after any calibration of the well type chamber, for checking the stability of the chamber. Furthermore, the response of the chamber should be checked at regular intervals using the same sources. As recommended in Section 5.3, a ¹³⁷Cs source is needed to cover the energy region for high-energy photons. Another low-energy long-lived source, e.g. an ²⁴¹Am, is needed for checking the constancy for low-energy photon calibrations. It is recognized, however, that a long-lived low energy source may not be available. In this case, at least a ¹³⁷Cs or ⁶⁰Co check source shall be used as a quality assurance check on the chamber stability. Further, for chambers using the abbreviated quality assurance procedure, and if the chamber is used for ¹⁹²Ir, ¹²⁵I or ¹⁰³Pd source calibrations, the re-calibration interval should be shortened from 5 years to 2 years. For beta-particle brachytherapy sources, constancy checks may be performed with either the photon sources, or a long-lived beta-particle source. The check sources should be inserted into the chamber with an appropriate spacer and/or holder in a reproducible way. Readings using the constancy sources, corrected for temperature, pressure and decay of the source, should remain constant to within ±0.5%.

7.2. Calibration of SSDL reference sources

If the well type chamber is of a different type than that used by the IAEA Dosimetry Laboratory, the chamber's response curve will not necessarily be the same as that in Figure A2 (see Appendix A) and the SSDL should determine its characteristics if necessary. In addition, the ion recombination correction for the chamber must be determined.

The SSDL should calibrate its ^{137}Cs reference source using the calibrated well type chamber. The source is calibrated by inserting it along with the appropriate spacer into the well type chamber insert so that the centre of the active portion of the source is located at the point of calibration. The reading, corrected for temperature and pressure and multiplied by the well type chamber system calibration factor given by the IAEA Dosimetry Laboratory, will give the reference air kerma rate for the SSDL reference source.

7.3. Calibration of hospital's well type chamber

When the hospital's well type chamber system is calibrated at the SSDL, it is done using the SSDL reference source. At first, a response curve for the hospital's chamber must be determined. Then the source is inserted into the hospital's chamber using the appropriate spacer and insert. The correction for ion recombination for the hospital's chamber must be determined and accounted for if necessary. If the hospital's well type chamber is open to the atmosphere, the reading must also be corrected for the temperature and pressure. A calibration factor is then determined for the well type chamber system in terms of reference air kerma rate per unit current. For lower dose rates, the current is normally measured by accumulating charge in a given time.

If the calibration of the hospital's well type chamber system by the SSDL is performed in a hospital, the SSDL well type chamber system and/or reference sources must be transported. For safety reasons, the transport of the SSDL's source is not generally recommended. Instead, the hospital's ^{137}Cs source should be calibrated with the SSDL's well type chamber system and then used to calibrate the hospital's well type chamber system using the procedures outlined in the previous paragraph. In the hospital, all precautions and calibration conditions mentioned above shall be observed.

7.4. Calibration of hospital's non-standard ^{137}Cs sources

The calibration of any non-standard ^{137}Cs source (sources other than those given in Table A1 in Appendix A) of the hospital may also be done using the calibrated well type chamber system. Work at the IAEA Dosimetry Laboratory has shown that the difference in the well type chamber calibration factors for the two recommended reference sources is less than 1.0%. If a non-standard ^{137}Cs source is placed in the calibrated well type chamber, it would be expected that there might be at most a 2% to 3% additional uncertainty in the calibration of the source.

7.5. Guidance for some special cases

7.5.1. Calibration for ^{192}Ir LDR wires

LDR sources of ^{192}Ir are supplied in different forms, e.g. wires, hairpins, single pins, etc. The wire may be up to 500 mm long and is delivered in the form of a coil. It is not recommended to calibrate the whole coil by free in-air measurements due to the complicated geometry and possible self-absorption of photons in the coil. A suggested procedure for ^{192}Ir LDR wires is as follows:

- A piece with a length of 10 mm of the ^{192}Ir wire is calibrated free in-air using the methods described in Section 6.

- The calibrated wire is used to calibrate a well type chamber. This calibration is done with the source center at the position of maximum response of the well type chamber.

With this method, the well type chamber is calibrated in terms of reference air kerma rate for the specific length of 10 mm of the ^{192}Ir wire. Thus, a calibration factor, $N_{K_R,10\text{mm}}$, can be determined.

Alternatively, a well type chamber system with a ^{192}Ir LDR seed calibration can be used to calibrate a short segment (e.g. 10 mm) of ^{192}Ir wire. This method is expected to yield results within $\pm 3\%$ of the free in-air calibration technique.

Ideally, the ratio $M_u/(L \cdot K_{R,\text{wire}})$, where $K_{R,\text{wire}}$ is the reference air kerma rate per unit source length, should be independent of the wire length L . As can be seen from Figure A2, this ratio will vary with the source length and for calibration of wires of different lengths, it is necessary to apply a correction factor, k_L , which will depend upon the source length L . The reference air kerma rate of the wire with a length L is then:

$$K_R = N_{K_R,10\text{mm}} \cdot M_u \cdot k_L \quad (17)$$

where $N_{K_R,10\text{mm}}$ is the reference air kerma rate calibration factor for a 10 mm length ^{192}Ir wire, M_u the corrected charge and k_L is the correction factor that takes into account the differences in the length of the source that is calibrated and the 10 mm wire that was used to calibrate the well type chamber.

TABLE XII. CORRECTION FACTORS, k_L , FOR DIFFERENT LENGTHS OF LDR ^{192}Ir WIRES

Wire length (mm)	HDR 1000	SDS
10	1.000	1.000
30	1.005	1.012
50	1.012	1.017
70	1.029	1.038
90	1.050	1.070

The factor k_L may be determined with a 10 mm piece of wire that is used to measure the chamber response for different positions of the wire along the central axis of the chamber [25]. These corrections can then be obtained from the integration of the response curve over the actual source length.

In Table XII correction factors are given for the HDR 1000 (Standard Imaging) chamber and the Nucletron SDS (PTW) well type chambers.

The values in Table XII are consistent with values found by Drugge [25]. To use these values, the centre of the wire must be positioned at the calibration point of the well type chamber. Positioning of hairpin and single pin sources in calibration procedures must replicate the procedure used at the SSDL, since some well type chambers have a small diameter cavity and are sensitive to radial positional changes.

7.6. Calibration of source trains

For calibration of such long sources the response curve (variation of chamber output with position within the chamber) must be determined. It should be noted that the response curve measured with a photon source will be different for one measured with a beta particle source. Ideally, the response curve of the well type chamber should be flat over the region where the sources are located in the chamber during the measurement. This can partially be achieved in some cases by using larger (i.e. longer) chambers. As an example, LDR ^{192}Ir sources used in intravascular brachytherapy can be as long as 92 mm, and ^{192}Ir wires can be even longer. As shown above in Table XII, corrections may have to be applied to the readings of long sources in shorter chambers. Calibrations of source trains should be performed with inserts which allow the centre of the source train to be positioned at the calibration point of the well type chamber, and which allow for different source lengths.

7.7. Traceability of ^{137}Cs source calibrations

The traceability of brachytherapy ^{137}Cs source calibrations, when acquired by using the well type chambers and reference sources as recommended above, is from the user's source to a PSDL through the SSDL and the IAEA Dosimetry Laboratory.

8. CALIBRATION OF BRACHYTHERAPY SOURCES BY USING OTHER DETECTORS

8.1. General

The methods recommended in Sections 6 and 7, free in-air calibrations and the calibrations by using well type ionization chambers, do not apply to the calibrations of beta ray planar and concave sources used in ophthalmic brachytherapy. Further, the use of another calibrated detector may become the preferred method for the calibration of beta ray seed and wire sources (instead of the extrapolation chamber.)

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 that approach the required properties are radiochromic film, thin plastic scintillators, thin thermoluminescence dosimeters (TLDs), diode detectors, diamond detectors, thin alanine detectors, photo-stimulated luminescence (PSL) systems and radiochromic gel dosimeters. There are a 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 B. A number of other characteristics, where the suitability of the detector is dependent on the sources to be measured, are summarized in Appendixes C to E. The characteristics of a number of detectors are also discussed in detail by ICRU [4].

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 B), as in the case of some radiochromic films, 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 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.

Another approach to measure the source strength 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, $D(0.07 \text{ mm})$, 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/cm² measured at 30 cm in air, $D(0.07 \text{ mm}, 300 \text{ mm})$, is related to reference absorbed dose rate at 2 mm in water, $D_w(2 \text{ mm})$, from the same source via the conversion factor Λ_β which is defined as:

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

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.

8.1.1. Calibration of 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 (see Section 8.1.3) 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 XIII (reproduced from [4]). To the first approximation, the relative depth dose values obtained, down to about 5 mm depth, are expected to conform to within about 10 % of the reference curve values.

The reference data in Table XIII 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:

$$\frac{D(z, r_0)}{D(z_0, r_0)} = \exp(a_s + b_s z + c_s z^2 + d_s z^3 + e_s z^4 + f_s z^5) \quad (18)$$

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

TABLE XIII. RELATIVE AXIAL DEPTH-DOSE DISTRIBUTION IN WATER FOR A PLANAR ^{90}Sr SOURCE AND FOR A PLANAR AND CONCAVE ^{106}Ru SOURCES [4]

Depth (mm)	$^{90}\text{Sr}/^{90}\text{Y}$ planar	$^{106}\text{Ru}/^{106}\text{Rh}$ planar	$^{106}\text{Ru}/^{106}\text{Rh}$ 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 XIV. COEFFICIENTS OF THE FITTED RELATIVE DEPTH-DOSE FUNCTIONS OF BETA RAY SOURCES [4]

Coefficient	$^{90}\text{Sr}/^{90}\text{Y}$ planar	$^{106}\text{Ru}/^{106}\text{Rh}$ planar	$^{106}\text{Ru}/^{106}\text{Rh}$ 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

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.

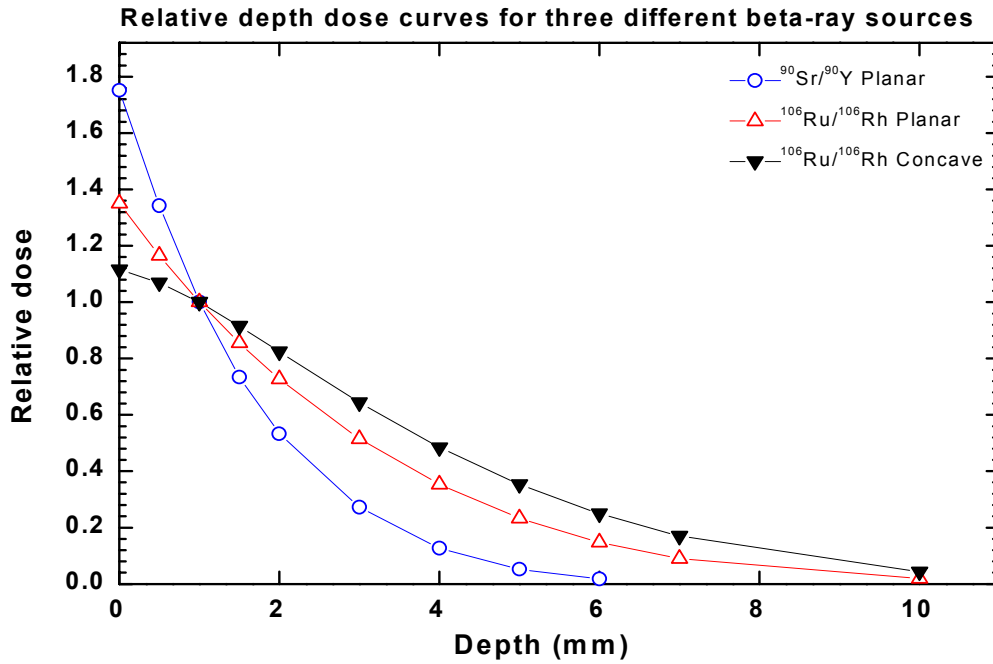


FIG. 5. Relative central depth dose curves for three beta ray sources normalized at 1 mm [4].

8.1.2. 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 F. 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.

8.1.3. 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 $D_{\text{avg}}(t,z)$ across a dosimeter of thickness t with its surface at depth z is given by [4]

$$D_{\text{avg}}(t,z) = \frac{\int_z^{z+t} D(z,r=0) dz}{t} \quad (19)$$

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 D, 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.

8.1.4. 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

$$k_G = R - (R^2 - d^2/4)^{1/2} \quad (20)$$

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

8.1.5. 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

$$D_m(r_m\rho_m) = (\eta_{m,w})^3 (\rho_m/\rho_w)^2 D_w(\eta_{m,w} r_w\rho_w) \quad (21)$$

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 XV.

TABLE XV. SCALING FACTOR $\eta_{m,w}$ FOR WATER-EQUIVALENT PLASTICS RECOMMENDED IN THIS GUIDE [4]

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 TM ”)	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.

8.1.6. Calibration of the detector

The preferred method of calibration of the detector is the calibration 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 B). The effective point of the detector must be placed at the depth in phantom where absorbed dose to water is specified.

8.2. Calibration of intravascular beta 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.

8.3. Uncertainties

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 of uncertainty. An example uncertainty analysis is given in Table XVI. 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 XVI. TYPICAL 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	

8.4. Traceability

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 in high-energy photon or electron beams, the traceability is that of the calibration of the ionization chamber.

9. QUALITY CONTROL

9.1. Safety aspects in the use of brachytherapy sources

The dose delivered to a patient undergoing brachytherapy treatment is directly proportional to the source strength. At the delivery of brachytherapy sources, these are accompanied with a certificate stating the source strength as determined by the manufacturer. Based on QC protocols, modern practice strongly recommends not to use this value as an input to dose calculation without independent verification by a qualified medical physicist.

A number of accidents have been reported in LDR and HDR brachytherapy treatments [38], resulting in an incorrect dose to the patient. The type of accident and their frequency is summarized in Table XVII.

Errors in the specification of the source activity, dose calculation or in the quantities and units resulted in doses that were up to 170% of the prescribed dose. Some of the accidents were caused by human mistakes, e.g. incorrect source was used for treatments because the colour coding of the source had faded. This is given under “Other” in the table, which includes also accidents caused by badly implanted sources, removal of the sources by the patient or otherwise dislodged sources. The most severe accident reported was due to equipment failure, where a lethal dose was delivered to the patient.

Of the total 32 cases reported, 7 could be attributed to the use of sources with incorrectly determined or stated activity [38]. In 6 of these, no independent check of the source strength was done. In 2 other cases the accident was caused by a mistake due to the incorrect use of quantities and units.

TABLE XVII. TYPE AND FREQUENCY FOR ACCIDENTS REPORTED IN BRACHYTHERAPY TREATMENTS

Accident caused by	Number of cases
Dose calculation error	6
Error in quantities and units	2
Incorrect source strength	7
Equipment failure	4
Other	13
Total	32

The recommended quantity by the ICRU for the specification of brachytherapy photon sources is reference air kerma rate [2, 3, 4]. However, other quantities are still in wide use, often dictated by those used in treatment planning systems. In such cases the use of conversion factors is necessary. Since conversion factors can vary substantially, due to the basic data or type of attenuation included, it is strongly suggested that only one quantity be used for dosimetry, i.e. the reference air kerma rate. With the use of a single quantity the amount of confusion would be reduced.

If a conversion from one unit to another must be done, a consistent set of conversion factors should be used. The subject of consistency is complicated and great care should be taken when using conversion factors. This can be exemplified by the following; the calibration performed by the manufacturer is traceable to a standards laboratory, but the source strength on the certificate is given using some other quantity. If there is a need to convert the quantity on the certificate, it must first be converted back to the traceable quantity using the same conversion factor as that used by the manufacturer. After this, a conversion to the desired quantity can be done. If this procedure is not followed, but the manufacturer converts the source strength given on the certificate using another factor than that used, the traceability of the source is lost.

The matter is further complicated if the dose planning system requires the source strength to be specified in some quantity, but for dose calculation purposes makes a conversion to another quantity. In this case, the documentation of the dose planning system must be consulted in order to determine the value of the conversion factor. With regard to these examples, it is easy to understand why severe accidents have occurred in this field.

When the source strength is entered into the dose planning system, the dose calculated with the system should be quality controlled. This can be done by calculating the dose at a well specified point using exact co-ordinates. The dose should then be compared with manual calculations, using a well established method. In the manual calculation of the dose, the source strength should be specified in terms of the traceable quantity, irrespective of the quantity that was entered into the dose planning system. The calculation should be done at a short distance, between 1 cm and 2 cm, because at these distances different calculation methods are, at least for ^{192}Ir , ^{137}Cs and ^{60}Co , in good agreement with one another, generally within 1%–2%. At larger distances the methods might differ, often due to different models for scatter and absorption correction, the effect of which is small at short distances.

The facilities in which brachytherapy sources are calibrated, i.e., calibration facilities and radiotherapy departments, shall meet the safety requirements established by International Basic Safety Standards for Protection against Ionizing Radiation and for the Safety of Radiation Sources (the BSS) [39].

The BSS include requirements for authorization of practices by the Regulatory Authority of the country, which apply to the practices referred to in this publication. The BSS include general requirements and detailed requirements on occupational, medical and public exposure, as well as potential exposure and emergency preparedness and response.

The BSS requirements on medical exposure establish that:

“Registrant and licensees shall ensure that: (a) the calibration of sources used for medical exposure be traceable to a Standards Dosimetry Laboratory; ... (c) sealed sources used for

brachytherapy be calibrated in terms of activity⁷, reference air kerma in air or absorbed dose rate in a specified distance, for a specified reference date;... ”

The standards are complemented by Safety Guides, which contain recommendations on how the requirements can be met. The relevant guides are: Occupational Radiation Protection [40], Assessment of Occupational Exposure Due to External Sources of Radiation [41] and Radiological Protection for Medical Exposure to Ionizing Radiation [42].

It is not the purpose of this report to focus on problems associated with the clinical use of brachytherapy sources. These have been addressed in detail in an IAEA report [43]. Therefore, the QC in the present report is limited to the calibration of brachytherapy sources; to the QC of the equipment used in the calibration and to the safety aspects related to the calibration procedures.

9.2. Well type chamber characteristics

Well type chambers provide a reliable method for calibrating brachytherapy sources before clinical use. There are at least two types of well type chambers that are used in many hospitals. High-pressure gas (usually argon) filled chambers, which were designed originally for assaying low activity radionuclides, and well type chambers that are open to the atmosphere. Loss of pressure due to gas leakage affects the sensitivity of the former type of chamber. Unlike chambers that are open to the atmosphere, such chambers do not require corrections for changes in the ambient temperature and pressure.

The insertion of a very high activity ¹⁹²Ir HDR source can cause a temperature increase inside the chamber [44]. Some chambers are designed with a Styrofoam insert to reduce this effect. The response of the chamber should be checked at regular intervals using a source of long half-life. A ¹³⁷Cs source is suitable for this purpose although other sources might be available. The source should be inserted into the chamber with appropriate spacer and/or holder in a reproducible way. Readings from use of the constancy source corrected for temperature, pressure and decay of the source should remain within $\pm 0.5\%$. The sensitivity of the chamber should be measured as a function of the depth of insertion of the source from the bottom of the chamber. The characteristic shape of this positional dependency depends upon the chamber design. A typical sensitivity plot is shown in Fig. 6.

Well type chambers respond to scattered radiation and should be used away from walls that might scatter radiation back to the chamber. Experimental determination of this effect might be required.

9.2.1. Source storage and handling of LDR sources

Suitable source storage containers are commercially available but can also be locally made. Whatever container is used, the dose equivalent rate at accessible distances from the surface of the container should not exceed 20 Sv/h.

⁷ As described in the present report and recommended by the ICRU, calibration of brachytherapy sources should be made in terms of reference air kerma rate (photon sources) or in terms of absorbed dose to water at a specified distance. Actually, calibration of photon emitting sources in terms of absorbed dose to water could also be considered, but this is unfortunately not technically possible today.

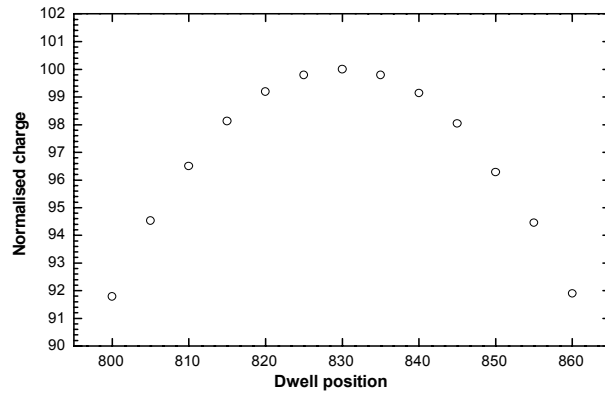


FIG. 6. Normalized charge versus source dwell position in a well type ionization chamber.

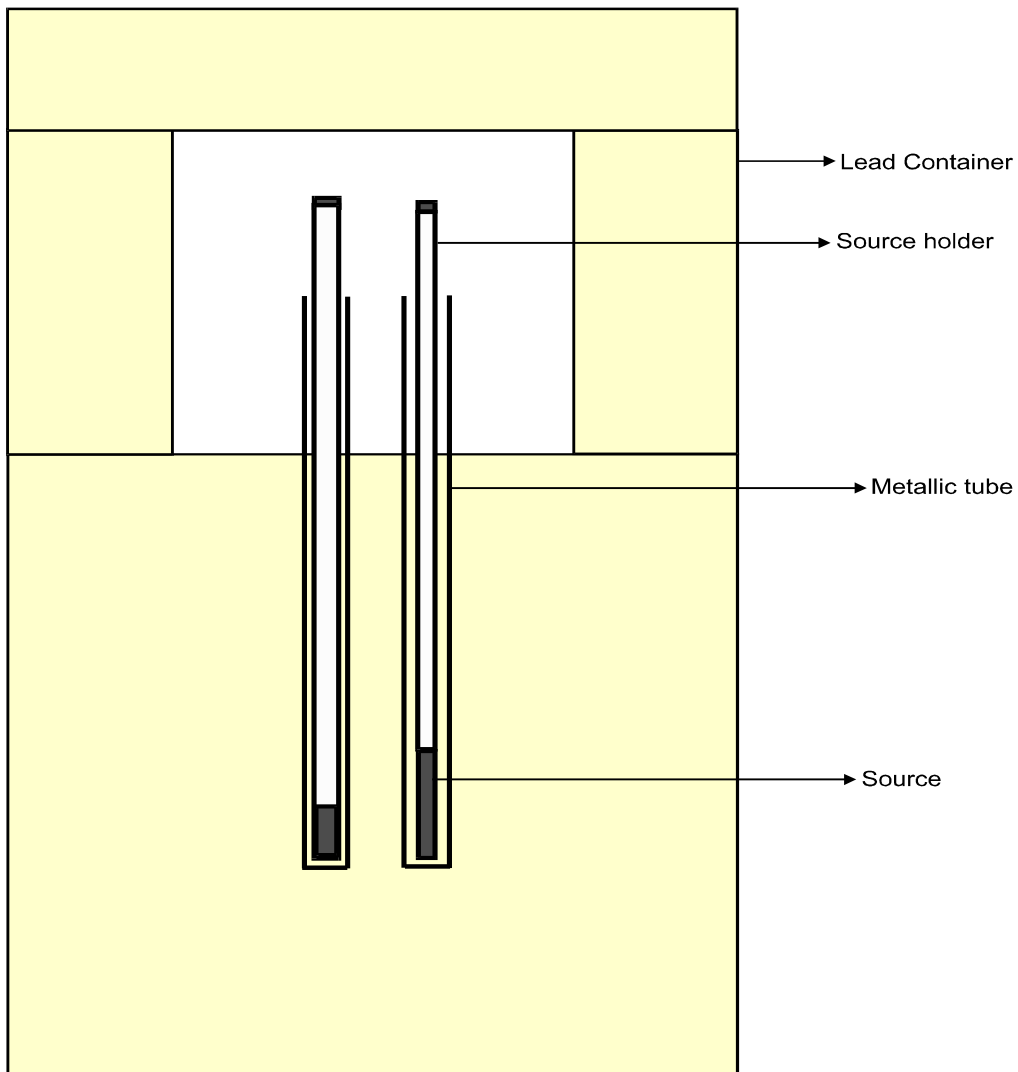


FIG. 7. Lead storage container used at IAEA Dosimetry Laboratory. The container has two metallic tubes in the centre, where the source holders are inserted.

A lead-shielded work bench and handling tools can be used for the safe handling of the sources. In the case of ^{137}Cs LDR reference sources, the sources should be loaded into Perspex tubes for ease of handling and to minimise radiation exposure. A cylindrical lead storage container, used at the IAEA Dosimetry Laboratory to store the ^{137}Cs reference sources is illustrated in Figure 7. The container has two metallic tubes near the centre to place the source holders.

9.3. Stability checks of the well type chamber

9.3.1. ^{137}Cs reference source check

At least one of the ^{137}Cs reference sources should be used to check the constancy of the well type chamber calibration. Such constancy checks should be made at least 4 times per year. The source should be inserted in the chamber with the appropriate spacer under reproducible conditions as mentioned in prior sections. The reading from the reference source corrected for temperature and pressure and for the decay of the source should remain within $\pm 0.5\%$ of the average of the 4 previous readings.

9.3.2. Other constancy checks

Other constancy checks may be performed if equipment is available. For example, a low activity ^{241}Am source of the type used for constancy tests of large volume ionization chambers may be inserted in the well type chamber and constancy established. It must be mentioned that the current obtained is several magnitudes of order smaller than when using a ^{137}Cs LDR source. The electrometer must therefore be able to measure currents of the order of pico-Amperes. The advantage of using ^{241}Am for checks is that its average photon energy is rather low (approximately 60 keV), being in the same range as the photon energy of ^{125}I .

9.4. Radiation safety

9.4.1. Leakage testing of the ^{137}Cs reference sources

The leakage of the reference sources should be tested by wet wipes every time a new reference source is received and in connection with each replacement of the Perspex insert tubes. With the help of the source handling tongs to minimise the radiation exposure of the operator, the source is wiped with a swab or tissue moistened with methanol or water, and the activity removed is measured. The wipe can be measured by a contamination monitor or gamma-spectrometric equipment, sensitive enough to detect the acceptable limit of 0.18 kBq. The leakage test should be done by an experienced radiation safety physicist.

9.5. Other precautions

Since the continuous exposure of the Perspex insert tubes to radiation makes them fragile and prone to breakage, it is recommended that they are replaced every six months and in no case, less often than once a year.

The sources shall be marked so that they can be easily identified. An up-to-date inventory of the sources must be kept and their storage marked with appropriate signs of radiation hazard. A general purpose survey meter must be available for monitoring of radiation levels near the sources and their containers.

9.6. Maintaining the traceability

As a regular monitoring of the traceability of the brachytherapy calibrations at the SSDLs, a recalibration of the SSDL well type chamber is recommended at regular intervals, as described in Section 7.1.6 or if the results of the constancy tests suggest a change in the sensitivity of the well type chamber.

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APPENDIX A
BRACHYTHERAPY STANDARDS AT THE IAEA DOSIMETRY LABORATORY

The well type chamber acquired by the IAEA for the calibration service to the SSDs is of type HDR-1000 Plus, designed at the University of Wisconsin and manufactured by *Standard Imaging Inc.* The diameter of the chamber is 102 mm, its height 156 mm and has an active volume of 245 cm³. Special inserts are provided for holding the sources, which are cylinders of diameter 35 mm and height 121 mm, with different inner diameters to suit different diameter sources. The outer aluminium wall of the chamber is 20 mm thick. The chamber has a vent hole to maintain the internal air at ambient atmospheric conditions. The electrometer used with the well type chamber is CDX-2000B, a digital portable instrument from *Standard Imaging Inc.*

To achieve the traceability of calibrations with the minimum uncertainty, it is essential that the reference sources and the source holders used for the calibrations of the well type ionization chambers, be as similar as possible in the different steps of the traceability chain. The ¹³⁷Cs sources used by the IAEA are shown in Table A1. For ease of handling, the sources are loaded into Perspex tube holders. The sources are then fixed in the holders using Perspex rods as illustrated in Figure A1.

TABLE A1. BRACHYTHERAPY REFERENCE SOURCES AT THE IAEA DOSIMETRY LABORATORY

Radio-nuclide	Type	Code	Nominal activity (MBq)	Encapsulation (mm of SS)	External dimensions Length	Diameter
¹³⁷ Cs	Tube	CDCSJ5	2313	0.5	20.0	2.65
¹³⁷ Cs	Cylinder	CDC1100	3700	0.5	8.0	3.20

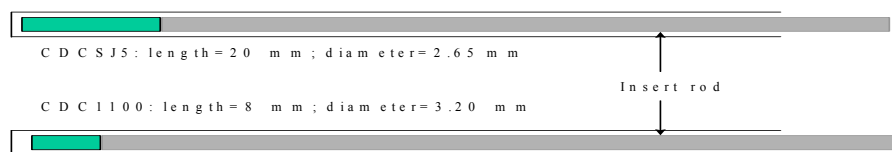


FIG. A1. The sources are inserted in Perspex tubes of length 150 mm and held in a fixed position using insert rods.

The ¹³⁷Cs sources of the IAEA have been calibrated in terms of reference air kerma rate at the NIST, USA. The reference air kerma rate of the sources on 1st May 1996, were 339 μGy·h⁻¹ for the CDC1100 type source and 190.5 μGy·h⁻¹ for the CDCSJ5 type source, with an estimated uncertainty of less than 2% at the 95% confidence level.

For most calibrations of the hospital well type chambers, the clinical sources available at the clinic can be used by the SSDs for the calibrations.

As an example of calibration point determination, the relative variation of the chamber response for the IAEA chamber, normalized to the maximum value, is shown in Figure A2. In all these measurements, the leakage current contributed less than 0.05% to the collected charge. It can be seen that the maximum response of the IAEA chamber is obtained for the CDCSJ5 type source (total length: 20 mm) when a 39 mm spacer is inserted at the bottom of the well, whereas a 45 mm spacer is needed for the CDC1100 type source (total length: 8 mm). This means that the maximum response is obtained when the centre of the source is at about 50 mm from the bottom of the well cavity. The response decreases by about 0.5% for a shift of about 9 mm on either side of the position of the maximum response.

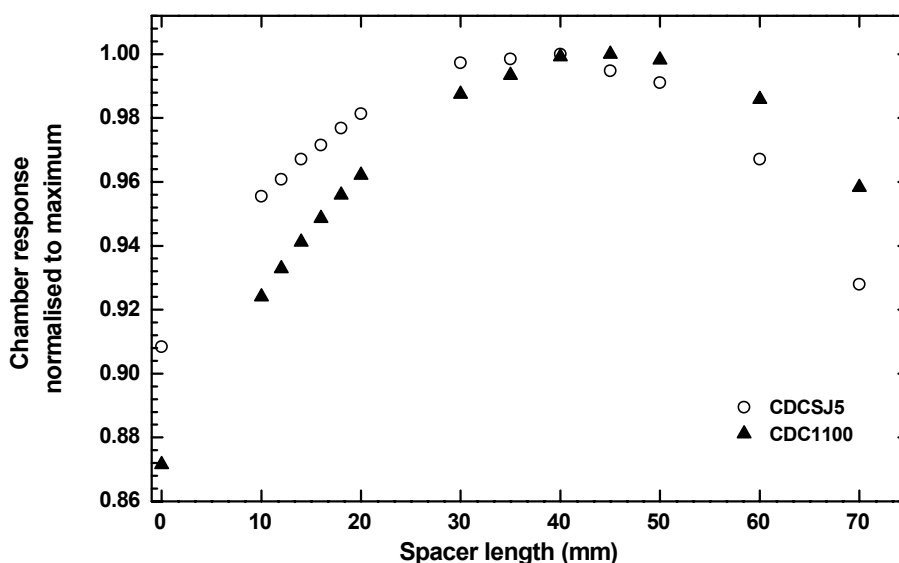


FIG.A2. Variation of the response of the well type ionization chamber with the length of the spacer.

Traceability from the IAEA to the SSDL

This section describes the procedures at the IAEA Dosimetry Laboratory and the SSDLs to establish the traceability of calibrations of ^{137}Cs LDR brachytherapy sources, using well type ionization chambers. As discussed earlier, the traceability for ^{137}Cs sources is the basis of maintaining the traceability also for other brachytherapy sources which can be calibrated by the same well type chamber.

The recommended method is based on the acquisition by the SSDLs of sources and a well type chamber similar to those at the IAEA Dosimetry Laboratory. The suggested method is that the SSDLs calibrate their well type chamber at the IAEA Dosimetry Laboratory using the IAEA reference sources. The various steps involved in establishing a calibration chain for the LDR ^{137}Cs brachytherapy sources from the PSDL to the hospital users through the IAEA Dosimetry Laboratory are as follows:

- The IAEA has two ^{137}Cs brachytherapy sources, calibrated at a PSDL in terms of reference air kerma rate. The sources, together with a well type chamber and an electrometer constitute the IAEA brachytherapy dosimetry standard.

- The SSDLs acquire an uncalibrated ^{137}Cs brachytherapy source and a well type chamber similar to those at the IAEA. The well type chamber/electrometer system and the source will constitute the SSDL brachytherapy dosimetry standard.
- The SSDL's well type chamber/electrometer system is calibrated at the IAEA Dosimetry Laboratory using the IAEA brachytherapy dosimetry standard.
- The SSDL measures the reference air kerma rate of its source using the calibrated system under the same conditions used for the calibration at the IAEA.

The SSDL calibrates user's sources and well type chambers using its standard.

Uncertainties

The overall relative uncertainty in the calibration of the IAEA reference ^{137}Cs sources at the NIST has been quoted as 2% at the 95% confidence level, i.e. approximately 1% for one standard deviation. The addition of the uncertainty of the measurements at the IAEA Dosimetry Laboratory yields a combined relative standard deviation of about 1.2% ($k=1$). Details on uncertainty estimation are given in Table XIII.

TABLE A2. ESTIMATED UNCERTAINTIES (%) FOR THE CALIBRATION OF THE SSDL WELL TYPE IONIZATION CHAMBER FOR ^{137}Cs AT THE IAEA DOSIMETRY LABORATORY

Uncertainty component	Type A	Type B
1. Measurements at the IAEA Dosimetry Laboratory:		
– source positioning in the well type chamber	0.04	
2. Charge measurement:		
– stability of the system (electrometer + chamber)	0.30	
3. Correction for influence quantities	0.2	
Recombination correction		0.1
Half-life of Cs-137		0.12
Impurity of the source		0.57
Square sum of components 1 to 3	0.13	0.35
Combined uncertainty, type A+B (1 standard deviation)	0.69	
4. Calibration of IAEA reference sources at NIST (type A+B)	1.00	
Total combined uncertainty, type A+B (1 standard deviation)	1.21	

Note: — The uncertainties for source position and stability are determined from a series of measurements made at the IAEA Dosimetry Laboratory.

- The uncertainty for the correction for influence quantities is taken from Ref. [36].
- The uncertainty in the half-life is given by the Nuclear Data Section of the IAEA.
- The uncertainty due to the impurity is taken as maximum probable presence of ^{134}Cs quoted by the supplier.

The SSDL should prepare a table of uncertainties for their well type chamber calibrations similar to Table XIII. If the source used by the SSDL is not the same type as that used by the IAEA laboratory, an additional uncertainty of up to 1% will be present in the table of uncertainties.

APPENDIX B. SOURCE INDEPENDENT CHARACTERISTICS OF DIFFERENT DETECTORS

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 environmental 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 C. DETECTOR SYSTEMS FOR CALIBRATION OF LOW ENERGY PHOTON SOURCES

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
Radiochromic film	Good	Good	Poor	Poor	N/A	Poor	Fair
TLD (LiF)	Poor	Fair	Fair	Fair	N/A	Poor	Poor
Plastic scintillator	Poor	Good	Fair	Fair	N/A	Poor	Fair
Diode	Fair/ Poor	Poor	Fair	Fair	N/A	Poor	Poor
Alanine	Poor	Good	Poor	Fair	N/A	Fair	Fair
PSL	Good	Poor	Good	Fair	N/A	Poor	N/A
Diamond	Poor	Fair	N/A	N/A	N/A	Poor	N/A
Parallel-plate ion chamber	N/A	N/A	N/A	N/A	N/A	Good	N/A
Polymer gels	Fair	Good	N/A	N/A	N/A	Good	Fair

APPENDIX D. DETECTOR SYSTEMS FOR CALIBRATION OF BETA RAY OPHTHALMIC APPLICATORS

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	Poor	Poor	N/A	Good	Fair
TLD (LiF)	Poor	Fair	Fair	Fair	N/A	Good	Poor
Plastic scintillator	Good	Good	Fair	Fair	N/A	Good	Fair
Diode	Fair/	Poor	Good	Fair	N/A	Fair	Poor
	Poor						
Alanine	Poor	Good	Poor	Fair	N/A	Fair	Fair
PSL	Good	Poor	Good	Fair	N/A	Fair	N/A
Diamond	Poor	Good	Good	Fair	Poor	Fair	Fair
Parallel-plate ion chamber	Poor	Fair/	Poor	Good	N/A	Fair	Poor
	Poor						
Polymer gels	Fair	Good	N/A	N/A	N/A	Fair	N/A

APPENDIX E. DETECTOR SYSTEMS FOR CALIBRATION OF BETA RAY SEED AND LINE SOURCES

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	Poor	Poor	N/A	Good	Fair
TLD (LiF)	Poor	Fair	Fair	Fair	N/A	Good	Poor
Plastic scintillator	Poor	Good	Fair	Fair	N/A	Fair	Fair
Diode	Fair/	Poor	Fair	Fair	N/A	Poor	Poor
	Poor						
Alanine					N/A	N/A	N/A
PSL	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
Parallel-plate ion chamber	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

APPENDIX F. GENERAL DATA ON DIFFERENT DETECTOR SYSTEMS

CHARACTERISTICS OF FEW COMMERCIALY AVAILABLE DETECTOR SYSTEMS

Detector	Effective thickness (mm)	Covering thickness (mm)	Offset depth mm (mg/cm ²)	Measurement diameter (mm)	Physical diameter (mm)
Radiochromic Film 16-18 µm emulsion layer on 0.1 mm PTP* backing	0.0017	0	0 (0.12)		Selectable
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 G. TRACEABILITY CHAIN AND CALIBRATION REQUIREMENTS FOR BRACHYTHERAPY SOURCES

TRACEABILITY OF CALIBRATIONS AND CALIBRATION CHECKS FOR BRACHYTHERAPY SOURCES

Step	Photon sources, long-lived nuclides <i>All clinical sources to be calibrated</i>	Photon sources, short-lived nuclides <i>All or random sample, min. 10 % of clinical sources to be calibrated</i>	Beta-ray sources <i>All clinical sources to be calibrated</i>
	^{137}Cs , (^{60}Co)	^{192}Ir	^{90}Sr - ^{90}Y , ^{106}Ru - ^{106}Rh , ^{32}P
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)
Working standard at PSDL	Large volume ionization chamber, free in-air measurements +reference source	Well type ionization chamber	Extrapol. chamber or calibrated detector +calibrated planar 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).	Calibrated planar source +Calibrated detector
Hospital user	Well type ionization chamber +reference source	Well type ionization chamber	Calibrated detector +planar reference source
			Calibrated detector +planar reference source
			Well type ionization chamber +ref. source
			Well type ion. Chamber +reference source
			Well type ion. Chamber +ref. source

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

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