

Clinical dosimetry — Beta radiation sources for brachytherapy

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National foreword

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Foreword

ISO (the International Organization for Standardization) is a worldwide federation of national standards bodies (ISO member bodies). The work of preparing International Standards is normally carried out through ISO technical committees. Each member body interested in a subject for which a technical committee has been established has the right to be represented on that committee. International organizations, governmental and non-governmental, in liaison with ISO, also take part in the work. ISO collaborates closely with the International Electrotechnical Commission (IEC) on all matters of electrotechnical standardization.

International Standards are drafted in accordance with the rules given in the ISO/IEC Directives, Part 2.

The main task of technical committees is to prepare International Standards. Draft International Standards adopted by the technical committees are circulated to the member bodies for voting. Publication as an International Standard requires approval by at least 75 % of the member bodies casting a vote.

Attention is drawn to the possibility that some of the elements of this document may be the subject of patent rights. ISO shall not be held responsible for identifying any or all such patent rights.

ISO 21439 was prepared by Technical Committee ISO/TC 85, *Nuclear energy*, Subcommittee SC 2, *Radiation protection*.

Introduction

Clinical dosimetry covers the methods by which values of the relevant physical quantity, absorbed dose to water, can be measured at a given point by the use of calibrated instruments in a clinical setting. The application of beta radiation sources for brachytherapy requires new and skilled methods for adequate clinical dosimetry necessitated by the short range of the beta radiation. This causes large dose-rate gradients around beta radiation sources, and hence it is necessary that the detector volumes for absorbed-dose measurements be extremely small. This leads to the requirement for highly specialized detectors and calibration techniques, and it is necessary to scrutinize closely every calibration obtained in one beta radiation field and determine if it is applicable in another field.

It is necessary that an appropriate quality system be implemented and maintained in the hospital for clinical beta radiation source dosimetry. It is the responsibility of the medical physicist to carry out testing and calibration activities for any source in such a way as to meet the requirements for adequate dosimetry. This International Standard gives guidance on how to satisfy these needs.

Clinical dosimetry — Beta radiation sources for brachytherapy

1 Scope

This International Standard specifies methods for the determination of absorbed-dose distributions in water or tissue that are required prior to initiating procedures for the application of beta radiation in ophthalmic tumour and intravascular brachytherapy [1], [2], [3]. Recommendations are given for beta radiation source calibration, dosimetry measurements, dose calculation, dosimetric quality assurance, as well as for beta radiation brachytherapy treatment planning. Guidance is also given for estimating the uncertainty of the absorbed dose to water. This International Standard is applicable to “sealed” radioactive sources, such as plane and concave surface sources, source trains of single seeds, line sources, and shell and volume sources, for which only the beta radiation emitted is of therapeutic relevance.

The standardization of procedures in clinical dosimetry described in this International Standard serves as a basis for the reliable application of beta radiation brachytherapy. The specific dosimetric methods described in this International Standard apply to sources for the curative treatment of ophthalmic disease, for intravascular brachytherapy treatment, for overcoming the problem of restenosis and for other clinical applications using beta radiation.

This International Standard is geared towards organizations wishing to establish reference methods in dosimetry aiming at clinical demands for an appropriately small uncertainty of the delivered dose. This International Standard does not exclude the possibility that there can be other methods leading to the same or smaller measurement uncertainties.

2 Normative references

The following referenced documents are indispensable for the application of this document. For dated references, only the edition cited applies. For undated references, the latest edition of the cited document (including any amendments) applies.

ISO/IEC Guide 98-3, *Uncertainty in measurement — Part 3: Guide to the expression of uncertainty in measurement (GUM:1995)*

ISO/IEC Guide 99, *International vocabulary of metrology — Basic and general concepts and associated terms (VIM)*

ISO 6980-2, *Nuclear energy — Reference beta particle radiation — Part 2: Calibration fundamentals related to basic quantities characterizing the radiation field*

ICRU Report 51, *Quantities and Units in Radiation Protection Dosimetry*

3 Terms and definitions

For the purposes of this document, the terms and definitions given in ICRU Report 51, ISO Guide 99 and ISO 6980-2, and the following apply.

3.1

absorbed dose

D

quotient of $d\bar{\epsilon}$ by dm , where $d\bar{\epsilon}$ is the mean energy imparted by ionizing radiation to matter of mass dm , as given by Equation (1):

$$D = \frac{d\bar{\epsilon}}{dm} \quad (1)$$

NOTE The absorbed dose is designated in units of joules per kilogram, with the special name of gray (Gy).

3.2

absorbed dose to water

D_w

quotient of $d\bar{\epsilon}$ by dm , where $d\bar{\epsilon}$ is the mean energy imparted to water by ionizing radiation to a medium of mass dm , as given by Equation (2):

$$D_w = \frac{d\bar{\epsilon}}{dm} \quad (2)$$

NOTE The absorbed dose to water is designated in units of joules per kilogram, with the special name of gray (Gy).

3.3

acceptance test

contractual test carried out by the user on receipt of a new instrument or source(s) in order to verify compliance with contractual specifications

NOTE 1 An acceptance test of an instrument is carried out after new equipment has been installed, or major modifications have been made to existing equipment.

NOTE 2 An acceptance test of a source is carried out on each source before being put into service for the first time. If a consignment contains more than one source, it is carried out on all sources of a particular type.

3.4

active source length

ASL

length of the source over which the absorbed dose rate at a defined distance from the source axis is within a specified ratio of the maximum absorbed dose rate at this distance

3.5

afterloading

automatically or manually controlled transfer of one or more sealed radioactive sources between a storage container and pre-positioned source applicators for brachytherapy

3.6 average beta energy

E_{ave}

quotient of beta energy averaged over the distribution, Φ_E , of the beta particle fluence with respect to energy as given by Equation (3):

$$E_{ave} = \frac{\int_0^{E_{max}} E \Phi_E(E) dE}{\int_0^{E_{max}} \Phi_E dE} \quad (3)$$

where $\Phi_E = d\Phi/dE$

3.7 brachytherapy

intracavitary, interstitial, superficial (including ophthalmic), or intraluminal (e.g. intravascular) radiotherapy in the immediate vicinity of one or more sealed or unsealed radioactive sources

3.8 calibration

set of operations that establish, under specific conditions, the relationship between values of a quantity and the corresponding values traceable to primary standards

NOTE 1 For an instrument, a calibration establishes, under specific conditions, the relationship between values of a quantity indicated by a measuring instrument or measuring system and the corresponding values realized from the standards.

NOTE 2 For a source, a calibration establishes, under specific conditions, the value of a quantity produced by the source.

3.9 clinical target volume CTV

gross tumour or target volume (GTV) with the addition of a margin that accounts for cells that are clinically suspected but have unproven involvement

NOTE In malignant disease, e.g. ophthalmic tumours, these oncological safety margins account for subclinical disease. In restenosis treatment, the CTV includes the full interventional length (IL) of the vessel with all vessel wall layers and with the addition of proximal and distal safety margins to include all tissue possibly injured during the interventional process.

3.10 detector test source

radiation source used for the determination of the long-term stability of a radiation detector

3.11 dosimeter

⟨beta radiation therapy⟩ equipment that uses detectors for the measurement of absorbed dose, or absorbed dose rate, in beta radiation fields as used in radiation therapy

NOTE A radiotherapy dosimeter contains the following components: one or more detector assemblies, a measuring assembly (including possibly a separate display device), one or more detector test sources (optional) and one or more phantoms (optional).

3.12 dwell time

time a radioactive source or source train remains at a selected treatment position

3.13
effective point of measurement

P_{eff}

point at which the absorbed dose rate in an undisturbed medium is determined from the detector signal

3.14
extrapolation chamber

ionization chamber capable of having a collection volume that is continuously variable to a vanishingly small value by changing the separation of the electrodes, which allows the user to extrapolate the measured ionization density to zero collecting volume

NOTE The extrapolation chamber serves as a primary standard, under proper conditions of use (see Annex C).

3.15
fluence

Φ

quotient of dN by dA_s , where dN is the number of particles incident on a sphere of cross-sectional area dA_s , as given by Equation (4):

$$\Phi = dN/dA_s \quad (4)$$

3.16
gross tumour or target volume
GTV

macroscopic extent and location of target tissue that can be observed or visualized using applicable imaging modalities

NOTE In malignant disease, e.g. ophthalmic tumours, target tissue means the demonstrable tumour growth. In restenosis treatment, the GTV includes the full vessel extent injured during the interventional process.

3.17
influence quantity

quantity that can have a bearing on the result of a measurement without being the subject of the measurement

3.18
interventional length
IL

length of the vessel injured during the interventional process

3.19
lesion length
LL

stenotic or occluded length of the vessel segment as determined by the interventionalist

3.20
maximum beta energy

E_{max}

highest value of the energy of beta radiation emitted by a particular radionuclide that can emit one or several continuous spectra of beta radiation each with a characteristic maximum energy

3.21
measurement standard

instrument that defines, represents physically, maintains or reproduces the unit of measurement of a quantity (or multiple or sub-multiple of that unit) in order to transfer it to other instruments by comparison

3.22

planning target volume

PTV

clinical target volume (CTV) plus safety margins to account for physiological movements and changes, as well as for various set-up uncertainties

3.23

point of test

point at which the conventional true value is determined and at which the reference point of the dosimeter is placed for calibration and test purposes

3.24

primary standard

measurement standard (of the highest metrological quality) that defines, represents physically, maintains or reproduces the unit of measurement of a quantity (or a multiple or sub-multiple of that unit) in order to transfer it to other instruments by comparison

NOTE 1 The primary standard is operated by a national laboratory under reference conditions and its accuracy has been verified by comparison with the comparable standards of institutions participating in the International Measurement System.

NOTE 2 A primary standard realizes the quantity being measured without reference to any other standard of the same type.

3.25

ionizing radiation

emission and propagation of energy through space or through a material medium in the form of electromagnetic waves or particles that have the potential to ionize an atom or molecule through atomic interactions

3.26

radiation detector

equipment, generally a sub-assembly, or substance that, in the presence of radiation, provides by either direct or indirect means a signal or other indication suitable for use in measuring one or more quantities of the incident radiation

3.27

reference absorbed dose to water

D_0

absorbed dose to water at the reference point

3.28

reference conditions

set of influence quantities for which the calibration is valid without any correction

NOTE The reference conditions for the quantity being measured may be chosen consistent with the properties of the instrument being calibrated. The quantity being measured is not an influence quantity.

3.29

reference isodose length

RIL

vessel length at the reference distance enclosed by a certain defined percentage isodose of the reference dose at P_{Ref}

NOTE 1 The reference distance is measured from the source axis to a line parallel to the source axis on which P_{Ref} is located.

NOTE 2 For example, ESTRO recommends the 90 % isodose.

3.30
reference lumen diameter
RLD

diameter of the vessel lumen after angioplasty as determined by angiography in a representative plane within the planning target volume

3.31
reference orientation of a detector

orientation of the dosimetry detector with respect to the direction of the incident radiation stated by the manufacturer

3.32
radionuclide purity

proportion of the total activity present in the form of the stated radionuclide

NOTE The radionuclide purity is generally expressed as a percentage.

3.33
reference point

P_{ref}
(for source calibration) point in a source radiation field at which the reference absorbed dose rate is specified, and which is also used for normalization of relative measurements

3.34
reference point of a detector

point of a detector that is placed at the point of test for calibrating or testing purposes

NOTE The distance of measurement refers to the distance between the reference points of the radiation source and of the detector.

3.35
routine calibration

calibration appropriate to a routine application of a source or an instrument

NOTE A routine calibration may be of a confirmatory nature when it is performed either to check the calibration carried out by the manufacturer together with an instrument, or to check whether the calibration is sufficiently stable during the continued, long-term use of a source or an instrument. When considering the most practical way to perform a routine calibration, results obtained in a type test can turn out to be helpful, for example in selecting the phantom.

3.36
secondary standard

standard whose value is assigned by comparison with a primary standard of the same quantity

3.37
source applicator

(brachytherapy) device to position one or more radiation sources at the intended treatment positions

NOTE The radiation source may be a fixed part of the applicator, and the applicator may, furthermore, include protective shielding and/or a source guide.

3.38
source train

sequence of sealed radioactive sources, possibly separated by non-radioactive spacers, that is specified by a single value and calibrated as a whole

3.39
special calibration

calibration of a source or an instrument for a special case similar to that performed in connection with a type test

NOTE A special calibration is performed, for example, if the source or instrument is used under special circumstances or if the routine or type testing provides insufficient information.

3.40

standard test conditions

conditions under which all influence quantities and instrument parameters have their standard test values

NOTE Ideally, calibrations should be carried out under reference conditions. As this is not always achievable (e.g. for ambient air pressure) or convenient (e.g. for ambient temperature), a (small) interval around the reference values may be used. In principle, corrections should be made for the deviations of the calibration factor (if dimensionless) or calibration coefficient (if the instrument indication has different units from the calibration quantity) from its value under reference conditions caused by these deviations. In practice, the uncertainty aimed at serves as a criterion to determine whether it is necessary to take an influence quantity into account by an explicit correction or whether its effect may be incorporated into the uncertainty. During type tests, all values of influence quantities that are not the subject of the test are fixed within the interval of the standard test conditions.

3.41

test

⟨of an instrument⟩ measurement intended to confirm that an instrument is functioning correctly and/or the quantitative determination of the variations of the indication of the instrument over a range of radiation, electric and environmental conditions

NOTE Four distinct categories of instrument testing, of which calibration is a part, are generally recognized: type test, acceptance test, special calibration, routine calibration.

3.42

test

⟨of a source⟩ measurement intended to confirm that a source is functioning correctly and/or that the encapsulation is intact, and/or the quantitative determination of the variations of the field of the source over a range of radiation, electric and environmental conditions

NOTE Four distinct categories of source testing, of which calibration is part, are generally recognized: type test, acceptance test, special calibration, routine calibration.

3.43

traceability

property of the result of a measurement or the value of a standard whereby it can be related to stated references, usually national standards or International Standards, through an unbroken chain of comparisons each having a stated uncertainty

3.44

transfer standard

standard used as an intermediary to compare standards and establish traceability

3.45

treatment parameter

factor that describes one aspect of the irradiation of a patient during radiotherapy, such as radiation energy, absorbed dose, treatment time

3.46

treatment time

time between initiation and termination of irradiation, excluding any time in the ready state after interruption

3.47

type test

⟨of an instrument⟩ test intended to determine the characteristics of a particular type or model of a production instrument

NOTE 1 This type test involves extensive testing over a wide range of quantities that can have a bearing on the result of a measurement without being the objective of the measurement: the “influence quantities”. For ionizing radiation detectors, such influence quantities are, for instance, energy, angle of incidence, dose or dose rate and radiation type, usually under a variety of environmental conditions.

NOTE 2 A type test is normally performed on a prototype or on an instrument taken at random from a production batch and intended to be typical of the type. A type test will normally be carried out by National or Secondary Standard Laboratories, which may make the information available to the instrument user.

3.48 type test

⟨of a source⟩ test intended to determine the characteristics of a particular type or model of a production source

NOTE 1 This type test involves extensive testing for a number of conditions that can have a bearing on the result of an irradiation.

NOTE 2 A type test is normally performed on a prototype or on a source taken at random from a production batch and intended to be typical of the type. A type test will normally be carried out by National or Secondary Standard Laboratories, which may make the information available to the source user.

3.49 water equivalence

property of a material that approximates the radiation attenuation and scattering properties of water for a specified range of radiation energies

3.50 water-equivalent material

material that absorbs and scatters a specified radiation quality to the same degree as water for a specified range of radiation energies

3.51 water phantom water-equivalent phantom

object made from water or a water-equivalent material having essentially the same radiation interaction properties as liquid water with respect to the dosimetric procedure under consideration

4 Beta radiation sources and source data

4.1 Ophthalmic and dural brachytherapy sources

Brachytherapy has been used since the beginning of the 20th century when applications using radium and radon seeds in skin applicators were performed. Besides beta radiation, these natural radioactive sources emit alpha and gamma radiation. A typical indication is the control of the formation of keloids. The development of artificial, less radiotoxic radioactive sources allowed an increase in the activity concentration and thus a reduction in treatment time, and an improvement in radiation protection as well. With the experience of ophthalmic brachytherapy using cobalt-60 (^{60}Co) applicators, beta radiation from strontium-90/yttrium-90 ($^{90}\text{Sr} + ^{90}\text{Y}$) planar and later curved applicators also started being used in the 1950s for the treatment of lesions of the eye, such as pterygia. Often referred to as “plaques”, in this International Standard, they are referred to as “applicators” or “radioactive ophthalmic brachytherapy sources”.

In the 1980s, radiotherapy for eye malignancies (e.g. uveal melanoma, retinoblastoma, hemangioma) [4], [5], [6], [7], [8], [9] was found to offer a therapeutic alternative to enucleation, being at least equally effective in controlling tumour growth and at the same time eye- and vision-sparing. For such ophthalmic treatments, a number of different beta emitters, and also photon sources, were used in the past, e.g., strontium-90/yttrium-90 ($^{90}\text{Sr} + ^{90}\text{Y}$), ruthenium-106/rhodium-106 ($^{106}\text{Ru} + ^{106}\text{Rh}$), cobalt-60 (^{60}Co), iridium-192 (^{192}Ir), gold-198 (^{198}Au) and radium-226 (^{226}Ra). Presently $^{106}\text{Ru} + ^{106}\text{Rh}$ ophthalmic brachytherapy sources are widely used, especially in Europe, and remain commercially available. Also $^{90}\text{Sr} + ^{90}\text{Y}$ ophthalmic brachytherapy sources are applied for a few cases [10], although they are currently not being manufactured.

For completeness, we mention that in the US, custom-made ophthalmic brachytherapy sources employing iodine-125 (^{125}I) or palladium-103 (^{103}Pd) are mainly used. Also, eye applicators which combine iodine-125 and ruthenium-106/rhodium-106 have recently been introduced into the clinical routine at Essen University Hospital [11]. Clinical dosimetry of this applicator and photon sources are both beyond the scope of this International Standard.

Very recently, thin yttrium-90 (^{90}Y) foil applicators have been used to treat spinal dura after tumour removal to control microscopic residual disease [12].

4.2 Intravascular brachytherapy sources

In intravascular brachytherapy (IVB), the vessel section injured by the interventional process of widening is treated with either beta or photon radiation [13], [14], [15]. In the coronary artery tree, the injured section lengths are usually on the order of 2 cm to 4 cm in arteries with diameters of 2 mm to 4 mm. It is also necessary to treat longer or more complex target volumes (long lesions up to 9 cm, multifocal lesions or bifurcations) in coronaries and in large peripheral vessels (tens of centimetres). This requires line sources with a very narrow diameter, less than 1 mm, able to fit through a brachytherapy catheter. Typical arrangements include encapsulated line sources mounted on the end of wires that can be used to insert and remove the sources to and from the treatment volume. Line sources may also be realized from linear arrays of "seeds," which can be delivered to the target site either manually or automatically. Radionuclides that have been used for these sources include ^{32}P , $^{90}\text{Sr} + ^{90}\text{Y}$, and ^{90}Y . The physical length of these sources varies (3 cm to 6 cm) to adequately cover the target volume. Stepping short wire sources (0,5 cm to 2 cm) are used to treat longer target volumes.

Other sources have been applied to the restenosis problem, however they were for the most part unsealed and their dosimetry is beyond the scope of this International Standard. For completeness, they include ^{188}Re and ^{186}Re radioactive liquid, ^{133}Xe gas-filled balloons, ^{32}P -coated balloons and radioactive stents [14]; no further details on these sources are given in this International Standard.

4.3 Characteristics of radionuclides

Table 1 shows a compilation of half-lives (with uncertainties), and maximum and average energies for the different beta radiation sources most commonly used for clinical applications [16].

Table 1 — Properties of radionuclides in the most commonly used clinical beta radiation sources

Beta emitter	Half life d	E_{max} MeV	E_{avg} MeV	Major photon radiation with percentage per decay
^{90}Sr	$10\,523 \pm 22$	0,546	0,195 8	none
^{90}Y	$2,667 \pm 0,008$	2,280 1	0,933 6	none
^{32}P	$14,263 \pm 0,003$	1,710 5	0,694 9	none
^{106}Ru	$373,59 \pm 0,15$	0,039 4	0,010	none
^{106}Rh	$(3,449 \pm 0,009) \text{ E-4}$	3,541 0	1,410	0,512 MeV (20 %) 0,622 MeV (10 %) 1,0 MeV (1,6 %) 1,13 MeV (0,4 %) 1,55 MeV (0,2 %)

4.4 Source specification

4.4.1 General

The manufacturer of brachytherapy sources shall provide the following information:

- reference data set (RDS) of the given source type, and
- calibration data (CD) of the specific source.

NOTE Appropriate documents containing the required data include the source certificate (SC) (as given in ISO 2919 ^[175] and IAEA No. TS-R-1 ^[17]) and the "Instructions for use".

The manufacturer is liable for his products according to legal requirements. Thus, the reference data set and the calibration data shall contain all data on which treatment planning at the required level is based. The measurement information required in the calibration data is important also for the clinical user who is required to make an independent verification of the source properties.

See Annex B for examples of data sheets.

4.4.2 Reference data set

4.4.2.1 General

The RDS shall contain at least

- the manufacturer's name, and address;
- the radionuclide, half life and maximum beta energy of the nuclide, and major photon radiation components and energies;
- a statement of radionuclide purity;
- source type identification, nominal dose rate and nominal contained activity;
- design of the source, including dimensions and composition (both radioactive core and encapsulation).

4.4.2.2 Reference data set for area sources

An RDS specifically for area (ophthalmic concave or planar) sources shall contain at least

- the active and physical diameters, window thickness and material, radius of curvature, dimensions of cutouts (e.g. see Annex B: product data sheet);
- typical values of the relative axial dose rate (depth dose distribution) with a resolution of at least 1 mm starting closer than 1 mm from the applicator surface to the bremsstrahlung background;
- typical two-dimensional distribution of the relative dose rate with a resolution of at least 10 % (in at least one plane including the source axis and, in the case of asymmetric sources, in a second orthogonal plane through the cutout), as well as in at least one plane perpendicular to the source axis preferably through the reference point.

4.4.2.3 Reference data set for line sources

An RDS specifically for line (intravascular seed arrays, wire or balloon) sources shall contain at least

- the seed number (if applicable), nominal inactive and active source length and diameter, number and position of markers on active and dummy sources, design and dimensions of the catheter and relevant tolerances;
- typical values of the relative radial dose rate as a function of the distance from the source axis with a resolution of at least 0,1 mm from 0,5 mm to the bremsstrahlung background, including values at about 50 %, 75 %, and 125 % of the range of the beta radiation;
- typical values of the relative longitudinal dose rate, along a line parallel to the source axis through P_{ref} ;
- support for three-dimensional dosimetry for intravascular sources (optional) (see 5.4).

4.4.3 Calibration data

4.4.3.1 General

Calibration data (CD) shall contain at least the reference dose rate at the calibration date/time with uncertainty (expanded with $k = 2$), with the method of determination and with a statement on the traceability to a primary standard.

4.4.3.2 Calibration data for area sources

Calibration data specifically for area (ophthalmic concave or planar) sources shall contain at least

- tabulated values of the absorbed dose rate to water (relative or absolute) as a function of distance from the applicator surface along the source axis, with a resolution of at least 1 mm, starting closer than 1 mm from the applicator surface and extending to the bremsstrahlung background, with uncertainty (expanded with $k = 2$) and method of determination;
- relative dose rate distribution at 1,0 mm distance from the surface normalized to the value at the axis with uncertainty (expanded with $k = 2$) and method of determination, and values of derived source non-uniformity (see 5.7.3) and source asymmetry (see 5.7.4);
- relative dose rate distribution in planes perpendicular to the source axis (to allow comparison with measurements in the clinic; see 11.1.6) (optional);
- source data to support three-dimensional dose distribution calculation (optional).

4.4.3.3 Calibration data for line sources

Calibration data specifically for line (intravascular seed arrays, wire or balloon) sources shall contain at least

- relative radial dose rate distribution (measured on a line vertical to the source axis through P_{ref} with resolutions of at least 0,5 mm);
- relative longitudinal dose rate distribution (measured at a line parallel to the source axis through P_{ref});
- statement on active source length;
- value of source non-uniformity (see 5.7.3);
- statement on equatorial anisotropy (see 5.7.5).

5 Dose calculation parameters and formalisms

5.1 General

In 5.2 to 5.7 are introduced the relevant radiation-field parameters and formalisms for expressing dose distributions about beta radiation sources. In this International Standard, discussion is limited to four source geometries: point, line, planar and concave.

5.2 Parameters of the radiation field

5.2.1 General

It is often convenient to parameterize the radiation field rather than to express the absorbed dose rate as a function of position around the source. An obvious first step in this parameterization is to use the absorbed dose rate at the reference point, and then use a relative value of the absorbed dose rate at other points. These relative values do not change with time (as does the absorbed dose rate for radionuclide sources) and presumably are the same from source to source for the same radionuclide and source construction.

5.2.2 Reference absorbed dose to water

The choice of the reference distance for the reference absorbed dose (rate) to water is based on the geometry of the sources used, the characteristic of the near-field dose distribution of the applied beta nuclides and the distance of clinical interest. Therefore, the reference point for planar beta radiation sources is located 1 mm from the source surface and for concave sources, 2 mm from the source surface, both on a line passing through the centre of the source area. The choice of 2 mm for the reference point distance for concave applicators, rather than the 1 mm recommended by the ICRU Report 72^[18], is justified by the difficult source geometry, which makes measurements at the closer distance a problem. For beta radiation point and line sources, the reference point is located at a distance of 2 mm measured from the source centre, and for line sources it is perpendicular to the source axis^{[18], [19], [20], [21], [22], [23]}.

The absorbed dose rate at the reference point, P_{ref} , defined as the reference absorbed dose rate to water, is denoted by $\dot{D}_0(\vec{r}_0)$, where $\vec{r}_0$ is the location of the reference point. The absorbed dose rate at any arbitrary location, $\vec{r}$, about the source is then given by Equation (5):

$$\dot{D}(\vec{r}) = \dot{D}_0(\vec{r}_0)R(\vec{r}) \quad (5)$$

where $R(\vec{r})$ is the relative absorbed dose rate function.

5.2.3 Dose distribution coordinate systems

The choice of coordinate system in which to express $\vec{r}$ is usually guided by the source geometry. For point-like sources, it is convenient to use spherical coordinates, $\vec{r}(r, \theta, \phi)$; however for planar, concave and line sources, a cylindrical coordinate system, $\vec{r}(\rho, z, \phi)$, is better adapted, especially if the source exhibits symmetry around the source axis. In this case, the absorbed dose rate varies only in the radial direction ("away" from the axis of a line source or the central axis of an area source) and in the axial direction ("along" a line source, or away from the surface of an area source). Equation (5) then reduces to the form given in Equation (6):

$$\dot{D}(\rho, z) = \dot{D}(\rho_0, z_0)R(\rho, z) \quad (6)$$

where ρ and z are the radial and axial coordinates, respectively, and (ρ_0, z_0) are the coordinates of the reference point. The function $R(\rho, z)$, when given in a tabular format, is commonly known as an "away and along" table. Figure 1 illustrates the three coordinate systems used. It is important to note the opposite meanings of away and along for the line source versus the area source geometries. The reference coordinates in the recommended systems for the various beta radiation source geometries are shown in Table 2.

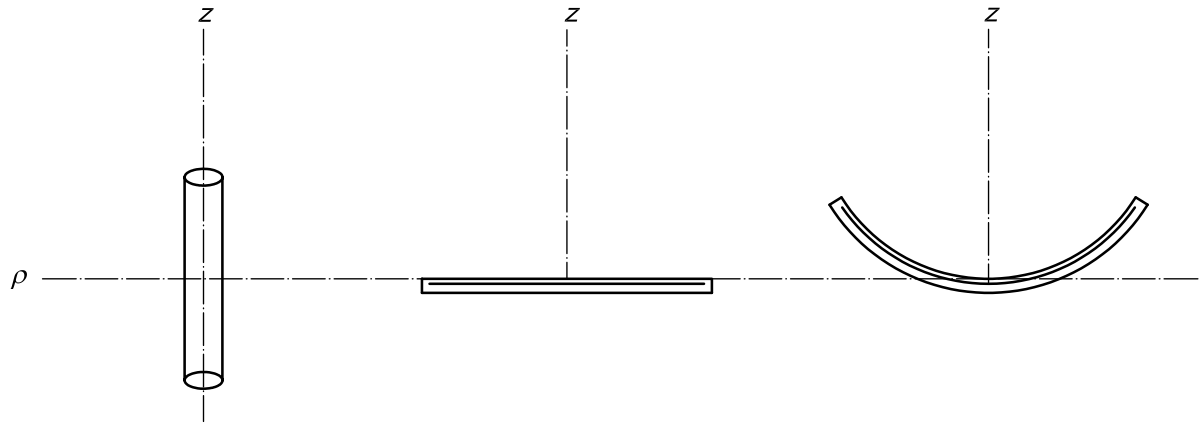


Figure 1 — Cylindrical coordinate systems for beta radiation source geometries

Table 2 — Reference points, P_{ref} , for beta radiation sources

Source geometry	ρ mm	z mm
Line source	2	0
Planar source	0	1
Concave source	0	2

The formalisms discussed assume cylindrically symmetric sources. For cylindrically non-symmetric sources (e.g. notched applicators), it is necessary that the formalisms be extended to include the dependence on the azimuthal angle, ϕ . For such sources, the Z axis should be specified by the manufacturer.

5.3 Radial dose profile

In spherical coordinates for a symmetric point like source, Equation (5) reduces to the form of Equation (7):

$$\dot{D}(r) = \dot{D}_0 R(r) \quad (7)$$

where $R(r)$ is known as the radial dose profile and is analogous to the central-axis depth dose function in radiotherapy with beam sources.

For line sources, it is possible to define a similar function, which corresponds to the $z = 0$ column of an “away and along” table. In this case, Equation (7) takes the form of Equation (8):

$$\dot{D}(\rho, z) = \dot{D}_0 R(\rho) H(\rho, z) \quad (8)$$

where $H(\rho, z)$ is the off axis dose function, normalized to unity for $z = z_0 = 0$. The derivation of Equations (9) and (10) should be obvious:

$$R(\rho) = \dot{D}(\rho, z_0) / \dot{D}_0 \quad (9)$$

and

$$H(\rho, z) = \dot{D}(\rho, z) / \dot{D}(\rho, z_0) \quad (10)$$

Because of the very high dose-rate gradients close to seed and line sources, a different approach is used for field parameterization of these sources (see 5.4).

For area sources, this is expressed in the form of Equation (11):

$$\dot{D}(\rho, z) = \dot{D}_0 R(z) H(\rho, z) \quad (11)$$

where

$H(\rho, z)$ is the off-axis dose function, normalized to unity for $\rho = \rho_0 = 0$;

$$R(z) = \dot{D}(\rho_0, z) / \dot{D}_0; \quad (12)$$

$$H(\rho, z) = \dot{D}(\rho, z) / \dot{D}(\rho_0, z). \quad (13)$$

5.4 Normalization of relative-dose data for seed sources

5.4.1 General

The protocol recommended by the AAPM TG 60 Report [22] for the calculation of the three dimensional dose distributions around beta radiation sources used in intravascular brachytherapy is based on the older TG-43 protocol [24], developed to apply only to photon-emitting sources like seeds used for interstitial brachytherapy. Therefore, the formalism has been expressed in spherical coordinates assuming symmetry about the source axis. The positions are represented in the remaining two coordinates, r and θ . Figure 2 shows a representation of this coordinate system. Using this formalism, the absorbed dose rate, $\dot{D}(r, \theta)$, is given by Equation (14):

$$\dot{D}(r, \theta) = \dot{D}_0(r_0, \theta_0) \cdot \frac{G(r, \theta)}{G(r_0, \theta_0)} \cdot g(r) \cdot F(r, \theta) \quad (14)$$

This approach is chosen for field parameterization rather than the “away and along” table approach [see Equation (8)] because use of the geometry function removes much of the high cell-to-cell variation in the “away and along” tables, making interpolations more accurate, as described in 5.4.2 to 5.4.4.

5.4.2 Geometry function, $G(r, \theta)$

The geometry function accounts for the variation of dose due to an idealized activity distribution of the source. It should be emphasized that it is only a convenient construct, meant to make the interpolations of the other parameters more accurate by removing the inverse square dependence of the dose rate distribution. The value of the geometry function is calculated under the line source approximation as given in Equation (15):

$$G(r, \theta) = \frac{\beta}{L r \sin(\theta)} \quad (15)$$

where

L is a length, usually taken as the active source length;

β is the angle subtended by the source length, L , at the point (r, θ) .

Referring to Figure 2, it can be seen that $\beta = \theta_2 - \theta_1$. The geometry function is meant to represent the absorbed dose distribution in the absence of scattering and absorption for a line source of length, L .

5.4.3 The radial dose function, $g(r)$

The radial dose function represents the variation of the absorbed dose rate along the perpendicular bisector of the source, and includes the effects of absorption and scatter in water. It can be calculated from the radial dose profile, $\dot{D}(r, \theta_0)$, using Equation (16):

$$g(r) = \frac{\dot{D}(r, \theta_0)}{\dot{D}(r_0, \theta_0)} \cdot \frac{G(r_0, \theta_0)}{G(r, \theta_0)} \quad (16)$$

The first quotient is simply the radial dose profile, $R(r)$, expressed in polar coordinates.

The radial dose function can be given in the form of a polynomial function, which is evaluated at the desired radial distance, r . This function is unity for r_0 .

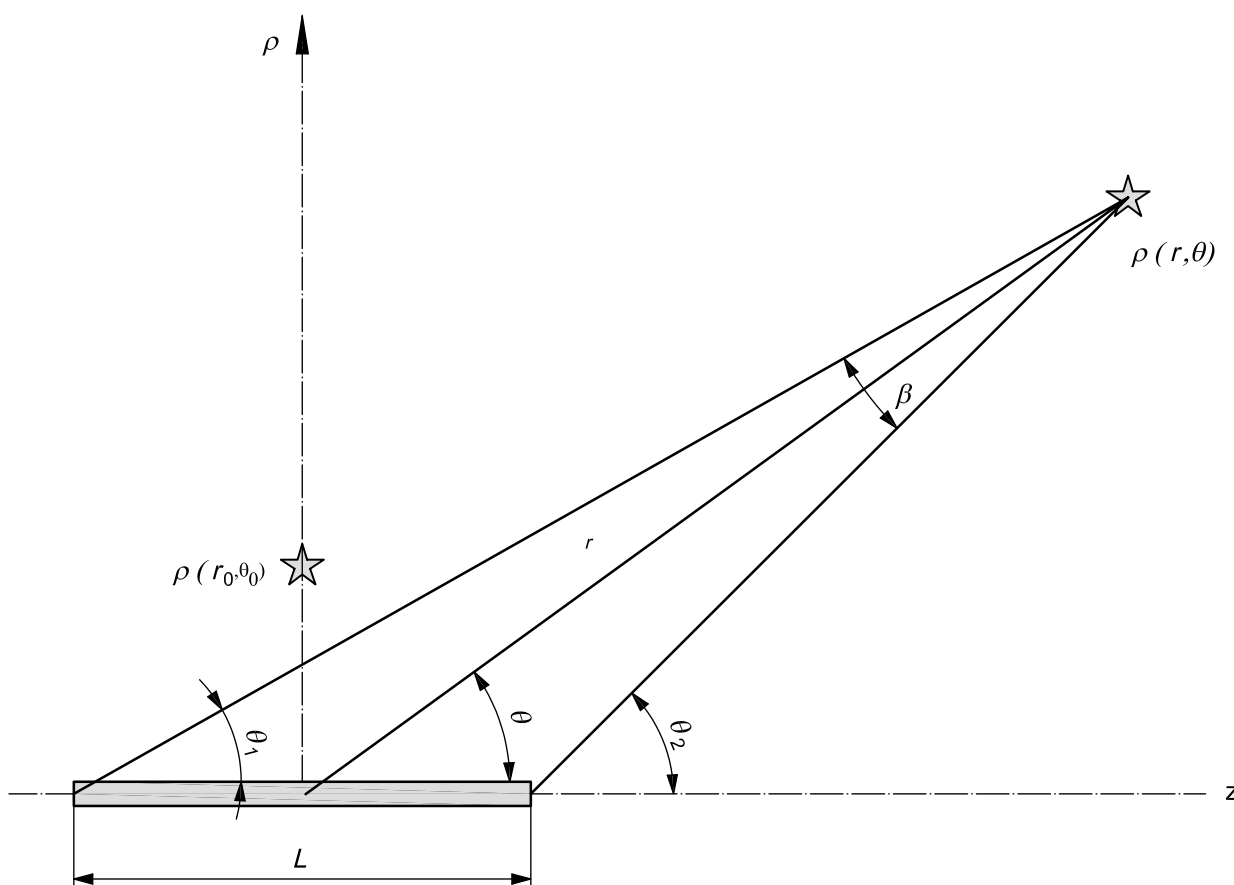


Figure 2 — TG-43 protocol coordinate system

5.4.4 The anisotropy function, $F(r, \theta)$

The anisotropy function represents the variation of the absorbed dose rate around the source in the θ direction and includes the effects of absorption and scatter in water. It can be calculated from the off-axis dose profile, appropriately converted to polar coordinates using Equation (17):

$$F(r, \theta) = \frac{\dot{D}(r, \theta)}{\dot{D}(r, \theta_0)} \cdot \frac{G(r, \theta_0)}{G(r, \theta)} \quad (17)$$

Again, the first quotient is simply the off-axis dose function, $H(r, \theta)$, normalized to unity at θ_0 . The anisotropy function is given in tabular format, where it is possible to interpolate the radial distance and the angle to obtain the value of the function at the desired coordinates.

5.5 Adaptation of the TG-43/60 formalism for a long beta radiation line source

5.5.1 General

The application of the TG-43/60 formalism breaks down in the case of a long beta line source or an entire source arrangement consisting of several seeds [26], [27]. As already introduced in 5.4, the AAPM TG 43/TG 60 formalism should be used for single-seed sources or small segments of a wire source. One solution to the breakdown problem is to use the results of calculations for small segments of line sources. One of the motivations for calculating only small segments is the possibility of modelling dose distributions from non-uniform sources, as well as to study the effect of changes in source length. Furthermore, using a 2,5 mm wire segment, there are no problems with a breakdown in the AAPM TG 43 formalism [25]. In addition, it is possible to make dose-profile calculations for a curved source, as can also be done with the seed-based sources, using the TG-43/60 formalism. This is a very powerful argument for not changing the formalism, but rather using data for a shorter wire segment to model the dose distribution for a longer source.

5.5.2 Introduction of cylindrical coordinates and new terminology for functions

Schaart, *et al.*, [26] have proposed an adaptation of the formalism that appears to work well for intravascular line sources but that preserves the essential advantages of the AAPM TG 43/TG 60 approach. The new formalism is conceptually similar to the AAPM TG 43/TG 60 formalism. However, the parameters of the new formalism are redefined to avoid the singularity problem described above. Furthermore, the new formalism is defined in cylindrical coordinates, which better match the source-target geometry. This has the advantage that an evenly spaced matrix of data points within the region of interest is easily obtained using equidistant intervals along the axes.

In the cylindrical coordinate formalism, the dose-rate distribution about a beta radiation line source in water is expressed as given in Equation (18):

$$\dot{D}(\rho, z) = \dot{D}_0 \cdot \frac{G(\rho, z)}{G(\rho_0, z_0)} \cdot g(\rho) \cdot F(\rho, z) \quad (18)$$

Here, $\dot{D}_0$ denotes the source strength that is defined as the absorbed dose rate in water at the reference point, i.e. at $\rho_0 = 2$ mm and $z_0 = 0$. Similar to the older TG-43 formalism, a geometry function is used to suppress the influence of the inverse square law on the other parameters. The general definition of the geometry function has been given in 5.4.2. Since the dose rate along a long, uniform beta radiation line source is uniform and independent of source length except near the ends of the source, it is appropriate to use the geometry function for an infinite line source, as originally proposed by Schaart, *et al.* [26]. In this approximation, the geometry function reduces to the form of Equation (19):

$$G(\rho, z) = G(\rho) = \frac{1}{\rho} \quad (19)$$

The transverse dose function, $g(\rho)$, which is normalized to unity at the reference point, is defined as given in Equation (20):

$$g(\rho) = \frac{G(\rho_0, z_0)}{G(\rho, z_0)} \cdot \frac{\dot{D}(\rho, z_0)}{\dot{D}(\rho_0, z_0)} \quad (20)$$

The transverse dose function plays a role similar to that of the radial dose function in that it characterizes the dose-rate distribution along the transverse bisector of the source. However, the transverse dose function is a function of the cylindrical radial distance, ρ , instead of the spherical distance, r . Therefore, if the above geometry function is used, $g(\rho)$ characterizes the entire dose distribution about a uniform beta line source, except in the dose fall-off regions near the ends of the source.

The non-uniformity function, $F(\rho, z)$, which is normalized to unity on the transverse axis, is defined as given in Equation (21):

$$F(\rho, z) = \frac{G(r, z_0)}{G(r, z)} \frac{\dot{D}(\rho, z)}{\dot{D}(\rho, z_0)} \quad (21)$$

If the geometry function in Equation (19) is used, the first term on the right-hand side vanishes so that the non-uniformity function completely describes the axial variation of the dose rate in terms of the ratio of the dose rate at a point and the dose rate on the transverse axis at the same radial distance. Thus, this function accounts not only for the dose fall-off near the source ends, but also for any dose-rate non-uniformity that can result from variations in, for example, the activity distribution or the encapsulation thickness along the length of the source.

5.6 Reference data sets

Reference values of relative dose-rate data are used both for simple manual calculation as well as for treatment-planning software. While they can be obtained by comparisons and averages of independently performed theoretical modeling calculations, for values obtained from measurements of real sources, it is necessary that high accuracy and reproducibility be achieved in order to avoid unacceptably high uncertainties in the resulting data. This issue should be assessed whenever reference-source data are presented.

At the present time, there are no published tabulations of reference-data sets for ophthalmic brachytherapy sources. However, some representative data for a planar $^{90}\text{Sr} + ^{90}\text{Y}$ source and a concave $^{106}\text{Ru} + ^{106}\text{Rh}$ source, taken from ICRU Report 72 [18], are given in Annex A.

For two beta radiation intravascular brachytherapy sources, reference data have been published [27]. These data are shown in Annex A for $^{90}\text{Sr} + ^{90}\text{Y}$ seed sources and for ^{32}P wire sources. In addition, calculated data are also given in Annex A for ^{32}P wire segments that can be used to build up an arbitrarily long source.

Also, as an aid to the calculation of $G(r, \theta)$, a sample table is given for a 2,5 mm source. Note that the quantity $r^2 G(r, \theta)$ is tabulated. This is an interesting quantity since it indicates any deviations from the inverse square law due to the line source approximation.

5.7 Parameters for source uniformity characterization

5.7.1 General

Real radiation sources are characterized by a less than perfectly uniform distribution of the radioactive material within the source volume. Additionally, the influences of encapsulation and its mechanical tolerances are more important for beta radiation than for photon sources. For this reason, parameters to characterize this non-uniform distribution are required to set limits on acceptable non-uniformities. Moreover, these limits can represent boundary criteria for the use of nominal dose-rate values.

5.7.2 Average reference absorbed dose rate to water for a line source

With the assumption of a uniform dose-rate distribution, the concept of reference absorbed dose rate, while defined at a point, can be extended to the useful length of a beta radiation source. And *vice versa*, the average absorbed dose rate of a beta radiation line source can be defined by the average of the absorbed dose rates at a distance of 2 mm from the source axis over the active length source minus a margin at each end of the source [28]. For ^{32}P sources, this margin shall be taken as 2,5 mm, while for $^{90}\text{Sr} + ^{90}\text{Y}$ sources, the margin shall be taken as 3,0 mm. To distinguish this averaged quantity from the reference absorbed dose rate, $\dot{D}_0$, the symbol $\dot{D}_{\text{avg}}$ is used.

5.7.3 Source non-uniformity

For area sources that have rotational symmetry, source non-uniformity at a given distance from the source surface is expressed as the percentage value of the maximum deviation of absorbed dose rate from the absorbed dose rate on the source axis ($\rho = 0$) within a radius given by $0,8 R_{50}$, where R_{50} is the radius of the 50 % isodose contour. Mathematically, the source non-uniformity, U_F , expressed as a percentage, is given by Equation (22) [28]:

$$U_F = \max \frac{(|\dot{D}_{\min} - \dot{D}_{\rho=0}|, |\dot{D}_{\max} - \dot{D}_{\rho=0}|)}{\dot{D}_{\rho=0}} \times 100 \quad (22)$$

Central to this definition is the concept of field centre, the algorithm for the determination of which is given in detail in ICRU Report 72 [18].

Uniformity measurements for area sources should be performed at a distance of 2 mm or less from the source surface in water-equivalent material. In practice, such a measurement parallel to, or on the surface of, a concave applicator is quite difficult to perform. Practical guidance in performing these measurements is given in 11.1.6.

For line sources, a similar concept is used, except that the deviation is determined and expressed relative to the average absorbed dose rate and the values compared restricted to those within the active source length less the margins given in 5.2.2. Thus, line source uniformity, U_L , expressed as a percentage, is given by Equation (23) [28]:

$$U_L = \max \frac{(|\dot{D}_{\min} - \dot{D}_{\text{ave}}|, |\dot{D}_{\max} - \dot{D}_{\text{ave}}|)}{\dot{D}_{\text{ave}}} \times 100 \quad (23)$$

5.7.4 Source asymmetry for rotationally symmetric area sources

Another useful parameter for characterizing source uniformity is the source asymmetry, U_{AS} , expressed as a percentage, as given by Equation (24) [28]:

$$U_{AS} = \max \left[\frac{\dot{D}_{\max}(\rho) - \dot{D}_{\min}(\rho)}{\dot{D}_{\text{ave}}(\rho)} \right] \times 100 \quad (24)$$

where $\dot{D}_{\text{ave}}(\rho)$ is the average of values around the circle defined by radius, ρ , at measurement depth, z . The distance from the source surface should be the same as used for the determination of non-uniformity.

5.7.5 Equatorial anisotropy for line sources

Analogous to source asymmetry, this parameter describes the variation in uniformity as measured at a fixed value of ρ , in a plane perpendicular to the source axis passing through the source centre ($z = 0$). U_{EA} , expressed as a percentage, is given by Equation (25):

$$U_{EA} = \max \left[\frac{\dot{D}_{\max}(\rho, 0) - \dot{D}_{\min}(\rho, 0)}{\dot{D}_{\text{ave}}(\rho, 0)} \right] \times 100 \quad (25)$$

6 Calibration and traceability

6.1 Measurand

The measurand absorbed dose to water, which is the clinically relevant quantity and is now being used in external beam dosimetry worldwide, shall also be employed to calibrate beta radiation brachytherapy sources. It is the most relevant quantity biologically and should be used in radiotherapy in general. Primary and secondary standards for realizing and disseminating the unit “gray” shall be designed for this measurand.

6.2 Traceability

To ensure that dose measurements performed by different users with different dosimeters are comparable, it is necessary that the instruments used be properly calibrated with the stated uncertainty, and that the measurements at the user's sites be traceable to a primary standard. This can be done either by a direct calibration of the user's instruments against the primary standard or in a two- or multiple-step process by transferring the unit “gray” to secondary and transfer standards. It is necessary to recognize that each additional transfer step increases the uncertainty of the measurement at the clinical site.

6.3 Reference point

For ophthalmic beta radiation sources, it is necessary that calibrations be performed at the reference point, P_{ref} , of $z = 1$ mm for planar sources ^[18] and $z = 2$ mm for concave sources, as measured on the central axis of the source (see 5.2.2 and 5.2.3).

For intravascular brachytherapy beta radiation sources, it is necessary that calibrations be performed at the reference point, P_{ref} , of 2 mm measured from the source centre perpendicular to the source axis ^{[18], [19], [20], [21], [22]}.

6.4 Primary standards

For beta radiation fields, the extrapolation chamber is used as the primary standard. A detailed description of the procedure for absorbed dose to water by means of extrapolation chambers is given in Annex C.

6.5 Secondary standards

For beta radiation, a secondary standard can be either a dosimetric system of proper accuracy calibrated by a national laboratory or a suitably calibrated radioactive beta radiation source whose radiation field is characterized by measurements performed at the national laboratory. The secondary standard can be used for the calibration of transfer standards or the user's dosimetry system. An example of a beta radiation secondary standard is described in detail in Clause C.7.

6.6 Transfer standards

Transfer standards or devices are used to transfer the unit of the measurement quantity from the well defined reference conditions of a primary or secondary standard to the conditions of a dose measurement in practice, where the actual values of the influence quantities can significantly differ from the reference conditions during calibration of the transfer standard. Thus, it is necessary to know the performance of the standard with respect to the relevant influence quantities, so that variations in the response of the instrument due to variations of influence quantities can be accounted for by correction factors. This requires a detailed investigation of such a transfer standard, which, for example, may be performed by a type test. Examples of dosimetry systems used as transfer devices are described in Clause D.1.

6.7 Calibration of therapeutic beta radiation sources

6.7.1 General

A measurement geometry as close to the clinical situation as possible shall be selected for the reference absorbed dose rate calibration. Thus, the measurement of the reference absorbed dose rate with a calibrated detector should be carried out in a water phantom whenever possible.

It is recommended that the calibration of therapeutic beta radiation sources be based on a dose-measuring system traceable to a national or international standard. An example of such a system is a $^{90}\text{Sr} + ^{90}\text{Y}$ secondary standard and a certified scintillation dose ratemeter.

6.7.2 Calibration of ophthalmic beta radiation sources

The calibration of an ophthalmic beta radiation source should be performed with a system calibrated by a national metrology institute or traceable to a primary standard, or with a measuring system that is calibrated in terms of absorbed dose to water relative to such a certified system by means of a $^{90}\text{Sr} + ^{90}\text{Y}$ secondary standard. It is anticipated that $^{106}\text{Ru} + ^{106}\text{Rh}$ may also be available as a secondary standard in the future. As an alternative, one may use other well characterized radiation fields, such as a ^{60}Co therapy beam or therapy electron beam spectra, whose dose rates at the depth of measurement are determined with instrumentation traceable to a primary standard [29]. Care should be taken to ensure that the system responds the same to the calibration beams as to the beta radiation source. It is necessary to stress that such procedures can result in an increased uncertainty because of the uncertainty in the energy and beam-quality dependence of detector response.

From the practical point of view, it is convenient to perform a calibration procedure by determining a complete depth dose distribution of the beta radiation source. Thus, the comparison of the measured curve with a standard depth dose distribution allows one to identify systematic deviations immediately.

A major requirement for any detector used for calibrations is that the dimensions of the sensitive volume are sufficiently small to allow one to neglect the variations in dose deposition within the sensitive volume (see 7.2.1). Although the dose gradient near a beta radiation applicator source follows an almost exponential fall off in the radial direction away from the source surface, usually the dose distribution is nearly constant in the two other dimensions. Thus the detector need only to be thin in one dimension, and the integration effect of the detector in the other dimensions can be neglected or taken into account by a simple unfolding procedure.

6.7.3 Calibration of intravascular beta radiation sources

For intravascular brachytherapy beta radiation sources, calibration should be done by using a recommended [19] system calibrated by a national metrology institute or traceable to a primary standard, or with a measuring system that is calibrated in terms of absorbed dose to water relative to such a certified system. For reference distances for dose specification, see Table 2. For intravascular brachytherapy sources, the reference distance is 2 mm from the source centre ($\rho = 2 \text{ mm}$, $z = 0 \text{ mm}$). Calibrations are performed in terms of absorbed dose to water. Reference conditions for calibration are the position of the source in the middle of a water phantom with dimensions of at least $15 \text{ cm} \times 15 \text{ cm} \times 15 \text{ cm}$ and a specific position of measurement at a radial distance, P_{ref} , measured from the middle of the source axis to the centre of the detector.

7 Dose measurements in-phantom and measurement corrections

7.1 Measurements in water or a water-equivalent phantom

It is useful to define three different types of measurements, corresponding to the three components of the complete three-dimensional representation of the radiation field:

- a) absolute dose rate at the reference location;

- b) relative dose rate along the perpendicular bisector;
- c) relative dose rate at points off the perpendicular bisector.

When considering line sources, it should be noted that all the formalisms so far assume that the radiation field is symmetric around the source axis. To verify this assumption, it is prudent to assess this assumption of uniformity by a fourth class of measurements: relative dose rate around the source axis at the reference distance (or some suitable substitute).

Absorbed dose to tissue is prescribed, but measurements can be made only in materials that approximate the scattering and absorption properties of tissue. Tissue equivalence describes the degree to which the material matches the tissue; it is usually quantified by a comparison of mass-energy absorption coefficients (for photons) and stopping powers (for beta radiation) over the energy range of interest. Since there is considerable variation in tissue composition and density in the human body, water is commonly used as the reference medium for performing the measurements. The use directly in water poses problems for some detectors, and it is necessary that they be enclosed in some sort of waterproof covering, which might or might not be water-equivalent for the radiation type and energy range being measured. Also, accurate source-detector positioning is more difficult in a water phantom. For this reason, recourse is often made to solid phantoms made from water-equivalent plastic. Such phantoms offer very reproducible positioning, within the limits achievable by precision machining. Accuracy of positioning is governed by how well the distances between source and detector can be measured. Some commonly used plastics are discussed in the next section.

Because of the difficult characteristics of near-field dosimetry, it is well worth the effort for the researcher to take great care in the design and setup of source-detector combinations. For precise positioning, solid phantoms are preferred, although the use of solid water analogues comes at the price of increased uncertainty because of the differences in radiation absorption and scattering between the solid medium and water. Precise machining is important, particularly for beta radiation measurements because of the possibility of beta radiation streaming in any air gaps, which results in loss of absorbed dose at the measurement point. Charge build-up can also be an issue for the higher-dose rate beta radiation sources, so the use of a conducting material can be a consideration. When catheters are used instead of bare sources, the position of the source within the catheter should be taken into account. In cases when this positioning might not be reproducible, as in the case of loose seeds injected hydraulically, multiple irradiations should be performed and the results averaged. When measurements of intravascular brachytherapy sources are performed in a water phantom, it is convenient to use after-loading techniques. Care should be taken so that the catheter into which the source is loaded does not change position when a source is inserted. It is helpful to use a cathetometer for performing precise measurements between the source and the measurement plane. For measurements of concave ophthalmic brachytherapy sources, again, water is the preferred medium, but recourse can also be made to solid phantoms of water-equivalent material of spherical shape to conform to the applicator. These phantoms can be cut to accept dosimeters (see, for example, Reference [29]). Appropriate water-equivalent materials for phantom construction are discussed in Clause D.2.

7.2 Detectors for beta radiation

7.2.1 Near-field dosimetry

Dosimetric measurements in close proximity to beta radiation brachytherapy sources are very difficult to perform. These extremely inhomogeneous dose distributions with a steep slope over several orders of magnitude of dose within a few millimeters require the measurement of absorbed dose within very small volumes. Thus, in order not to disturb the physical situation being measured, it is necessary that the dosimetry detector be as tiny and as water-equivalent as possible^[31]. Although for central axis depth dose measurements for planar and concave beta radiation applicators where the lateral-dose gradient is not large, very thin and flat detectors can be used. Dose measurements close (less than 5 mm) to an ophthalmic brachytherapy source and intravascular sources require the use of detectors providing spatial resolutions on the order of 0,1 mm to 1,0 mm for all dimensions. Such small dimensions require great sensitivity (for active devices) or can involve long irradiation times (for passive devices). In addition, it is necessary that the detector probe-packaging material, especially the front cover, be very thin to achieve precise depth dose or off-axis dose measurements, and should be water-equivalent as well. Further requirements are accuracy in absolute dose measurement (capable of being calibrated), precision and linearity of response over several orders of

magnitude of dose (rate). The spatial variation of the dose distributions inside the detector volume is usually not negligible. The effective point of measurement of the absorbed dose (or dose rate) depends on the distance to the source and the detector orientation as well. It is necessary that possible energy dependence or anisotropy of response be taken into account. Thus, it is necessary that the effective reference point of dose measurement in the vicinity of a brachytherapy source be determined and considered for each detector, for each type of radiation field, for each position inside the field and probably for each orientation of the detector probe (see 7.4).

7.2.2 Possible detectors for calibration of beta radiation sources

The requirements of 7.2.1 allow only the smallest volume plastic scintillator systems, radiochromic films, and appropriately small silicon diodes of routine use for source calibration. Other detectors, such as small-volume ionization (including extrapolation) chambers, MOSFET detectors, liquid ion chambers, alanine films, gel dosimeters, small-volume diamonds and, for some applications, ultra-thin thermoluminescent dosimeters (TLD's) may also be used, but are not recommended for routine measurements (see Annex D).

7.3 Conversion of absorbed dose in solid phantoms to absorbed dose to water

Absorbed dose measurements of therapeutic beta radiation sources are often performed in solid phantoms made from low-atomic-number materials such as A-150 plastic, WT1, polystyrene or polymethylmethacrylate (PMMA) (see 7.1). The depth dose distribution in such a solid phantom is generally different from that in water, therefore requiring a correction. Monte Carlo calculations, in which the phantom medium is replaced by water, can be used for such corrections. Another possibility is to correct through scaling.

Traditionally, beta radiation measurements made in low-atomic-number media other than water were scaled by expressing depths and distances in terms of areal density, that is, mass per unit area (see, for example, Reference [30]). A more accurate scaling method has been proposed by Cross [32] in the late 1960s. Cross proposed that the depth-dose distributions of plane and point-like beta radiation sources in different low- Z media are very similar and can be related to one another via a scaling factor based on distance. Since then, this so-called scaling method has been used for a variety of applications and the results in low- Z media appear to be accurate to within 2 % to 3 % for the types of sources just mentioned [33]. Recently, the applicability of the scaling method has been extended to line sources with zero or finite diameter by Schaart [34].

The similarity between beta radiation depth-dose distributions in different media can be understood as follows. At beta radiation energies, the mass stopping power, S/ξ , of low- Z media is dominated by the mass electronic (or collisional) stopping power S_{el}/ξ . According to the well-known Bethe formula, S_{el}/ξ is approximately proportional to the electron density, $N_A Z/M$, where N_A is Avogadro's number, Z is the atomic number and M the molar mass of the medium [33], [35]. The electron density tends to decrease with increasing Z ; so if the transport of electrons were determined only by S_{el}/ρ , one would expect them to penetrate more deeply (in terms of areal density) into higher- Z materials. This is, for example, reflected in the increase of the continuous slowing down range, R_{CSDA} , (in units of mass per area) with increasing Z . It can also be seen from Bethe's equation that the variation of, S_{el}/ρ , with Z is nearly independent of its variation with energy. As a result, the ratio of the stopping powers experienced by an electron in two different materials is nearly the same at all energies. One can, therefore, expect that the dose distributions in different media are related by a scaling factor based on distance equal to this ratio.

It is necessary that the influence of electron scattering be taken into account as well, however. As follows from the expression by Mott, the single-scattering cross-section per nucleus is approximately proportional to Z^2 [33]. Since the scattering events are statistically independent, one may argue that the mean-square angle of deflection per unit areal density due to nuclear scattering, also called the mass scattering power, is proportional to $N_A Z^2/M$ [33], [35]. This phenomenon tends to increase the mean path length (and therefore the energy loss) per unit areal density with increasing Z .

The net result appears to be a decrease of the average depth of penetration (in units of mass per area) with increasing Z [36], [37]. More importantly, Mott's equation shows that the relative angular distribution of the scattering probability is nearly the same for all low- Z elements. The same is therefore true for materials composed of multiple low- Z elements. It then follows that the scattering probability per unit energy loss of such materials is approximately proportional to an effective atomic number defined by Equation (26) [32]:

$$\bar{Z} = \frac{\sum_i N_i Z_i^2}{\sum_i N_i Z_i} \quad (26)$$

where

N_i is the relative number of nuclei of charge Z_i .

Furthermore, it follows from Mott's and Bethe's equations that this variation with $\bar{Z}$ is nearly independent of the variation with electron energy. Thus, by approximation, the ratio of the scattering probabilities per unit energy loss of different low- Z materials is independent of angle and remains constant with electron energy.

Cross, therefore, proposed that the dose distributions in different media can be related by a distance scaling factor, equal to the product of the stopping power ratio and some function of $\bar{Z}$ [32]. This function is obtained empirically by deriving scaling factors relative to water from experimental or calculated dose distributions and dividing these by the appropriate mass stopping power relative to water, S/S_w . Since stopping power ratios vary somewhat with energy, the ratio of the ranges of 500 keV electrons, R_w/R , is usually taken as the average mass stopping power relative to water. It appears that the results can then be fitted by Equation (27) [33]:

$$\eta = (0,777 + 0,037\,56\,\bar{Z} - 0,000\,66\,\bar{Z}^2) \frac{S}{S_w} \quad (27)$$

Scaling factors obtained from Equation (27) are given in Table D.2 for several different media [33]. The estimated standard uncertainty in these values is 1,5 %. These scaling factors can be used to scale the dose distributions between different, homogenous low- Z media, provided that the source geometry shows a sufficient degree of symmetry. For an (effectively) infinite, planar beta radiation source, the dose distribution in water as a function of the distance, x (expressed in units of length), to the source can be calculated from the dose distribution, $D(x)$, in a different low- Z medium as given in Equation (28) [32], [33]:

$$D(x) = \eta \cdot D_w\left(\eta \frac{\xi}{\xi_w} x\right) \quad (28)$$

where

η is the scaling factor, or relative attenuation, of the medium relative to water;

ξ is the density of the medium;

ξ_w is the density of the water.

It is noted that the same relationship holds for broad, parallel beta-particle beams incident on a slab of material. In this case, the variable, x , is the depth of penetration, expressed in units of length, into the material (see, e.g., ISO 6980-2 for a detailed example).

For a point source, the scaling relationship is given in Equation (29) [32], [33]:

$$D(r) = \eta^3 \cdot \frac{\xi^2}{\xi_w^2} \cdot D_w\left(\eta \frac{\xi}{\xi_w} r\right) \quad (29)$$

were r is the distance to the point source, expressed in units of length. For the more general case of a source with spherical symmetry and finite diameter, it has been shown that Equation (39) holds [34]:

$$r^2 \cdot D(r) = \eta \cdot [r_s + \eta \frac{\xi}{\xi_w} (r - r_s)]^2 \cdot D_w(r_s + \eta \frac{\xi}{\xi_w} [r - r_s]) \quad (30)$$

where r_s is the outer radius of the source, expressed in units of length.

For a line source having a diameter of zero and a length larger than twice the range of the beta particles, the depth-dose distributions in different media are related as given in Equation (31) [34]:

$$D(\rho) = \eta^2 \cdot \frac{\xi}{\xi_w} \cdot D_w(\eta \frac{\xi}{\xi_w} \rho) \quad (31)$$

where ρ is the radial distance to the longitudinal source axis, expressed in units of length. For the more general case of a cylindrically symmetric source with a finite diameter and a length exceeding twice the beta-particle range, Equation (32) holds [34]:

$$\rho \cdot D(\rho) = \eta \cdot [\rho_s + \eta \frac{\xi}{\xi_w} (\rho - \rho_s)] \cdot D_w(\rho_s + \eta \frac{\xi}{\xi_w} [\rho - \rho_s]) \quad (32)$$

where ρ_s is the outer radius of the source, expressed in units of length.

7.4 Effective point of measurement in the detector

A detector with a finite volume yields a reading proportional to the absorbed dose averaged over the volume of the detector.

For water-equivalent detectors, the average absorbed dose, D_{avg} , over a detector with a volume, V_{det} , is given by Equation (29):

$$D_{\text{avg}} = \frac{1}{V_{\text{det}}} \int_{V_{\text{det}}} D_w(\vec{r}) dV \quad (33)$$

where

dV is an infinitesimal volume at point $\vec{r}$ within the detector volume;

$D_w(\vec{r})$ is the absorbed dose at that point.

The effective point of measurement, P_{eff} , is the point in water with the same absorbed dose as the average absorbed dose over the detector, i.e. $D_w(P_{\text{eff}}) = D_{\text{avg}}$.

An equivalent way of defining the effective point of measurement is to calculate the average electron fluence, $\Phi_{\text{Avg,Det}}$, within the detector volume and define P_{eff} as the point in the undisturbed medium with the same electron fluence as $\Phi_{\text{Avg,Det}}$. In this case, the electron fluence includes the beta particles emitted from the source and electrons liberated by bremsstrahlung.

The effective point of measurement for a particular source/detector geometry can be assessed with reference dosimetry data. Generally, if the detector is reasonably thin (less than 1 mm in the dimension of the highest dose gradient), differences between the effective point of measurement and the centre of the detector sensitive volume are minimal. Recent investigations [90] at the Technische Bundesanstalt (PTB) with plastic scintillator detectors showed that, due to the shift of the effective point of measurement regarding the reference point of the detector, it is necessary to expect a maximum systematic deviation of 1 % or 2,5 % in the case of dosimetry of eye applicators or line sources, respectively [38]. The results are based on the