Dose calculation for photon-emitting brachytherapy sources with average energy higher than 50 keV: Report of the AAPM and ESTRO

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Purpose: Recommendations of the American Association of Physicists in Medicine (AAPM) and the European Society for Radiotherapy and Oncology (ESTRO) on dose calculations for high-energy (average energy higher than 50 keV) photon-emitting brachytherapy sources are presented, including the physical characteristics of specific ^{192}Ir, ^{137}Cs, and ^{60}Co source models.
Methods: This report has been prepared by the High Energy Brachytherapy Source Dosimetry (HEBD) Working Group. This report includes considerations in the application of the TG-43U1 formalism to high-energy photon-emitting sources with particular attention to phantom size effects, interpolation accuracy dependence on dose calculation grid size, and dosimetry parameter dependence on source active length.
Results: Consensus datasets for commercially available high-energy photon sources are provided, along with recommended methods for evaluating these datasets. Recommendations on dosimetry characterization methods, mainly using experimental procedures and Monte Carlo, are established and discussed. Also included are methodological recommendations on detector choice, detector energy response characterization and phantom materials, and measurement specification methodology. Uncertainty analyses are discussed and recommendations for high-energy sources without consensus datasets are given.
Conclusions: Recommended consensus datasets for high-energy sources have been derived for sources that were commercially available as of January 2010. Data are presented according to the AAPM TG-43U1 formalism, with modified interpolation and extrapolation techniques of the AAPM TG-43U1S1 report for the 2D anisotropy function and radial dose function.
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Key words: brachytherapy, TG-43 formalism, high-energy brachytherapy sources, Monte Carlo, experimental dosimetry, quality assurance

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I. INTRODUCTION

In 1995, the American Association of Physicists in Medicine (AAPM) Task Group No. 43 published a clinical protocol on dosimetry for interstitial brachytherapy sources,[1] colloquially known as the "TG-43 formalism," and provided reference dosimetry datasets for several designs of ^{192}Ir, ^{125}I, and ^{103}Pd sources commercially available at the time. This report was instrumental in enhancing dose calculation accuracy and uniformity of clinical dosimetry practices for low-energy photon-emitting sources following general acceptance and implementation of the TG-43 dose calculation formalism by the brachytherapy vendor, treatment planning systems (TPS), and user communities. Development of the TG-43 methods in the area of low-energy brachytherapy source dosimetry, defined as sources emitting photons of average energy less than or equal to 50 keV, was carried out by the AAPM Low Energy Brachytherapy Source Dosimetry (LEBD) Working Group. In response to the vastly increasing use of low-energy interstitial brachytherapy sources, especially for permanent prostate implants, and the increasing number and variable design of commercially available low-energy sources, LEBD continued to develop the TG-43 formalism and to prepare reference-quality AAPM consensus dosimetry datasets from published dosimetry papers. Most of the recent LEBD recommendations and advances in dosimetric characterization, recommended dose calculation methodologies, and data evaluation for low-energy interstitial brachytherapy are summarized in two key reports: the 2004 update of the TG-43 report (TG-43U1)[2] and its 2007 supplement (TG-43U1S1).[3,4] In the field of high-energy brachytherapy dosimetry, the TG-186 report will provide guidance for early adopters of model-based dose calculation algorithms. The model-based dose calculation algorithms (MBDCA) Working Group will develop a limited number of well-defined test case plans and perform MBDCA dose calculations and comparisons. However, there will remain for the foreseeable future a need for reference dosimetry data obtained in liquid water phantoms to evaluate the uniform clinical implementation and robustness of these advanced dose calculation algorithms.

Many publications propose various dose-estimation methods and dosimetric parameters for specific high-energy brachytherapy sources (defined as photon-emitting sources with average photon energies exceeding 50 keV) including ^{192}Ir, ^{137}Cs, ^{60}Co, and ^{198}Au sources. Many new source designs, especially high-dose rate (HDR) and pulsed-dose rate (PDR) sources, have been introduced for use in remote-afterloading machines, while traditional low-dose rate (LDR) sources such as ^{192}Ir seeds in ribbons, ^{192}Ir wires, and ^{137}Cs tubes and spheres remain a mainstay for a number of

brachytherapy applications. New brachytherapy radionuclides, such as ^{169}Yb (Refs. 5 and 6) and ^{170}Tm,[7–9] are being actively investigated for application in HDR brachytherapy and should be discussed in the forthcoming TG-167 report. Also, new ^{60}Co sources have been designed to be used with HDR afterloaders.[10–12] HDR remote afterloading units are generally replacing traditional LDR ^{192}Ir and ^{137}Cs sources for intracavitary and interstitial brachytherapy applications. This trend will continue as other new high-energy brachytherapy sources are developed. It is paramount that the computational and experimental tools used in investigations to evaluate single-source dose distributions, consensus dataset formation processes, and calibration processes are able to support the level of dosimetric accuracy and precision required to safely and efficiently deliver brachytherapy to patients.[13,14] To ensure that these criteria are met, reference dosimetry datasets obtained from these investigations must be independently verified for accuracy and be readily available in a format accepted by commonly used planning systems. The AAPM has made recommendations on dose calculation formalisms and the choice of dosimetry datasets for brachytherapy sources in its TG-43,[1] TG-56,[13] and TG-59 (Ref. 15) reports. Currently, the number of source models in clinical use is very large, and medical physicists have few resources to turn to for selecting the best dosimetry parameters for a given source model. The availability in tabular form of critically evaluated and complete consensus dosimetry datasets for all commonly used sources, for use with the updated TG-43 formalism, would be of substantial benefit to clinical end users.

The AAPM has reviewed and published reference-quality dosimetry datasets for low-energy brachytherapy sources in the LEBD reports (TG-43, TG-43U1, and TG-43U1S1). No similar effort has been attempted by AAPM or the European Society for Radiotherapy and Oncology (ESTRO) for high-energy sources, nor have societal recommendations been made concerning appropriate methods for the acquisition and formation of such datasets. To fill this void, the AAPM Brachtherapy Subcommittee (BTSC) formed the High Energy Brachytherapy Source Dosimetry (HEBD) Working Group to focus on photon-emitting brachytherapy sources with average energy higher than 50 keV. This group has the following charges:

1. To compile a list of high-energy brachytherapy sources commonly used in North America and Europe, for which the dosimetry datasets and guidelines recommended by HEBD will apply.
2. To develop dosimetric prerequisites for routine clinical use of high-energy brachytherapy sources similar in scope to the low-energy brachytherapy dosimetry prerequisites.[16]
3. To develop an extension of the TG-43 dose calculation formalism that is applicable to elongated sources, i.e., with maximum linear dimensions that are large or comparable to typical calculation distances.
4. To provide consensus datasets for the sources defined in charge 1 above, using the currently acceptable dose calculation formalisms.
5. To perform a review of existing clinical source strength calibration requirements and recommendations for high-energy (LDR/HDR/PDR) sources.
6. To provide a Brachytherapy Source Registry (BSR) for web-based access to high-energy brachytherapy source dosimetry data that satisfy the prerequisites defined in charge 2.

The objective of this report is to fulfill charges 1 and 4. Charge 2 was addressed in the first publication of the group[17] developing a set of dosimetric prerequisites for routine clinical use of brachytherapy sources with average energy higher than 50 keV. These broad recommendations form the basis of the more detailed recommendations provided by this report. Charge 3 has been adopted as the principal charge by the joint AAPM/ESTRO Task Group No. 143 on Dosimetric Evaluation of Elongated Photon-Emitting Brachytherapy Sources. Charge 5 on high-energy source calibrations is in progress for inclusion in a complementary report. Charge 6, to expand the BSR in an analogous manner as done for low-energy sources, is an on-going collaborative project involving the Radiological Physics Center (RPC), the AAPM BTSC and BSR Working Group, and the ESTRO BRAchytherapy PHYsics Quality assurance System (BRAPHYQS) subcommittee, analogous to the BTSC. Specifically, the current report addresses the following:

(a) Review the construction and available published dosimetry data for high-energy ^{192}Ir, ^{137}Cs, and ^{60}Co sources that (i) continue in clinical use in North America or Europe and (ii) satisfy the AAPM's dosimetric prerequisites[17] (charge 1).
(b) Perform a critical review of the existing TG-43U1 formalism[2] as used heretofore mainly for low-energy brachytherapy sources. Extension of the TG-43 dose calculation formalism was not performed as considered in charge 3.
(c) Critically review published dosimetric data for each of the prerequisite-compliant source models listed in (a) and develop a complete consensus dataset to support clinical planning for each source model (charge 4).
(d) Develop guidelines for investigators on the use of computational and experimental dosimetry for determination of high-energy brachytherapy source dosimetry parameters.

The full report containing detailed descriptions on the sources included in this report, along with quantitative consensus datasets, is available on the AAPM website.

The recommendations included herein reflect the guidance of the AAPM and the ESTRO for brachytherapy users and may also be used as guidance to vendors in developing good manufacturing practices for sources used in routine clinical treatments.

Certain materials and commercial products are identified in this report in order to facilitate discussion and methodology description. Such identification does not imply recommendation nor endorsement by any of the professional organizations or the authors, nor does it imply that the materials or products identified are necessarily the best available for these purposes.

II. PHYSICAL CHARACTERISTICS OF HIGH-ENERGY PHOTON-EMITTING BRACHYTHERAPY SOURCES

The photon-emitting brachytherapy sources included in this report have average energies exceeding 50 keV. Only sources intended for conventional clinical interstitial and intracavitary use were included; sources intended for intravascular brachytherapy are covered by AAPM Task Groups TG-60 (Ref. 18) and TG-149.[19] Similarly, electronic brachytherapy sources will be addresed by the AAPM Task Groups TG-167 and TG-182. The limit of 50 keV was established by the AAPM to separate high-energy sources from those addressed by the LEBD.[16]

This report addresses brachytherapy source models that were commercially available as of January 2010. For sources that were commercially available, the goal was to generate consensus datasets in a format acceptable to commercial treatment planning systems. For sources that are in current clinical use but no longer manufactured, the scientific literature was reviewed and acceptable published datasets were identified. In a few cases, datasets were included for sources that are no longer in clinical use to assist in the retrospective calculation of dose distributions.

The radionuclides considered in this report and described in this section are ^{192}Ir, ^{137}Cs, and ^{60}Co. Their most important physical properties are presented in Table I; see the National Nuclear Data Center (NNDC)[20] for a more complete description. Baltas *et al.*[21] also provides a clear description of these radionuclides. Detailed information on recommended photon spectra is provided in Sec. V.D.1.

^{198}Au (half-life 2.7 days) brachytherapy sources have been used extensively in the past for treatment of various tumors including gynecological, breast, prostate, head and neck, and other soft tissue cancers. These sources were generally of low activity (typically mCi) and were in the form of seeds or "grains." ^{198}Au emits a wide spectrum of x-rays and gamma-rays with an average energy of approximately 400 keV. The use of this radionuclide has decreased in recent years, perhaps because of the availability of competing radionuclides. These include ^{125}I (half-life 59.4 days) and ^{103}Pd (half-life 17.0 days), both of which have longer half-lives, making shipment and scheduling of treatments more convenient, and lower photon energies, leading to more acceptable radiation safety characteristics than ^{198}Au. Vicini *et al.*[22] conducted a survey of 178 publications reporting on prostate brachytherapy between 1985 and 1998. They found that ^{198}Au had not been used for monotherapy according to these studies and had been used in combined modality therapy only in 11% of cases. Correspondingly, they found that ^{125}I and ^{103}Pd were used far more frequently. Yaes[23] showed that, regardless of treatment site, the heterogeneity of the dose distributions from ^{198}Au could be greater than those from ^{125}I and ^{103}Pd. Similarly, Marsiglia *et al.*[24] reported that ^{198}Au implants more often showed significant cold spots, and generally inferior dosimetric coverage, than did implants with other radionuclides. These reports, together with others reporting on comparisons with other radionuclides, have resulted in relatively infrequent use of ^{198}Au. As a result, this report will not address ^{198}Au brachytherapy sources.

II.A. ^{192}Ir

The ^{192}Ir half-life of 73.81 days allows it to be easily used for temporary implants. Its high specific activity makes it practical to deliver sources of activities of as much as hundreds of GBq. ^{192}Ir decays to several excited states of ^{192}Pt via β^- (95%) and ^{192}Os via electron capture (EC) (5%), emitting on average 2.3 gamma rays per disintegration with a range of energies between 0.061 and 1.378 MeV and a mean energy of 0.355 MeV. The β^- rays emitted have a maximum energy of 0.675 MeV and an average energy of 0.1807 MeV. ^{192}Ir is produced from enriched ^{191}Ir targets (37% natural abundance) in a reactor by the (n, γ) reaction, creating HDR ^{192}Ir sources (typically 1 mm diameter by 3.5 mm length cylinders) with activities exceeding 4.4 TBq. HDR ^{192}Ir sources are encapsulated in a thin titanium or stainless steel capsule and laser welded to the end of a flexible wire. Electrons from β^-decay are absorbed by the core and the capsule.[25–28]

TABLE I. Physical properties of radionuclides considered in this report. Data have been taken from the NNDC (Ref. 20). Mean photon energy values are calculated with a cut-off of $\delta = 10$ keV. Data on Auger and IC electrons are not included.

	^{192}Ir	^{137}Cs	^{60}Co
Half-life	73.81 days	30.07 yr	5.27 yr
Type of disintegration	β^- (95.1%), EC (4.9%)	β^- (100%)	β^- (100%)
Maximum x-ray energy (keV)	78.6	37.5	8.3
Gamma energy-range (keV)	110.4–1378.2	661.6	1173.2–1332.5
Mean x-ray and gamma energy (keV)	350.0	613.0	1252.9
Maximum β^- ray energies (keV)	81.7 (0.103%) 258.7 (5.6%) 538.8 (41.43%) 675.1 (48.0%)	514.0 (94.4%) 1175.6 (5.6%)	318.2 (99.88%) 1491.4 (0.12%)
Mean β^- ray energy (keV)	180.7	188.4	96.5
Air-kerma rate constant, $\Gamma_{\delta=10\text{ keV}}$ (μGy m^2 h^{-1} MBq^{-1})	0.1091	0.0771	0.3059
Specific activity (GBq mg^{-1})	341.0	3.202	41.91

II.B. ^{137}Cs

The ^{137}Cs half-life of 30.07 yr enables use over a long period of time. Its low specific activity makes it practical for LDR implants. ^{137}Cs decays purely via β^-, mainly (94.4%) to the second excited state of ^{137}Ba, where the de-excitation to the ground state (90%) with emission of a gamma ray of 0.662 MeV (absolute intensity 85.1%) is in competition with internal conversion (IC) (10%). The β^- rays emitted have a maximum energy of 0.514 MeV. A second β^- decay branch (5.6% probability) to the ^{137}Ba ground state occurs, with maximum β^- ray energy of 1.176 MeV. ^{137}Cs is extracted from ^{235}U fission products, with the ^{137}Cs trapped in an inert matrix material such as gold, ceramic, or borosilicate glass. The sources are doubly encapsulated with a total of 0.5 mm thick stainless steel. Electrons from β^- decay are absorbed by the core and the capsule.[28] Cylindrical source models commercially available are manufactured with 3 mm diameter and external lengths up to 21 mm. Spherical sources are made for use in remote-afterloading intracavitary brachytherapy catheters.

II.C. ^{60}Co

The ^{60}Co half-life of 5.27 yr and its high specific activity make it practical for HDR brachytherapy implants. Newly designed HDR sources have been introduced in the market. ^{60}Co undergoes β^- decay to the excited states of ^{60}Ni (94.4%). De-excitation to the ground state occurs mainly via emission of γ-rays of 1.173 and 1.332 MeV, each with an absolute intensity of nearly 100%. The main β^- rays emitted (99.88%) have a maximum energy of 0.318 MeV and an average energy of 0.096 MeV. ^{60}Co is produced through neutron capture by ^{59}Co, but its long half-life requires long irradiation times for sufficient source strength. HDR ^{60}Co sources have dimensions similar to those of ^{192}Ir (Sec. II.A). The low-energy electrons emitted by ^{60}Co are easily absorbed by the cobalt source material or encapsulation layers, resulting in a "pure" photon source.[10,28]

III. CONSIDERATIONS APPLYING THE TG-43U1 FORMALISM TO HIGH-ENERGY PHOTON-EMITTING BRACHYTHERAPY SOURCES

The TG-43 formalism[1] was initially developed for use in interstitial brachytherapy including low-energy ^{125}I and ^{103}Pd seeds and high-energy ^{192}Ir seeds in ribbons. In 2004, the AAPM TG-43U1 report[2] updated the formalism and provided data for several new models of low-energy seeds. The application of this formalism has subsequently been extended significantly by the brachytherapy physics community, making it the international benchmark for nearly all brachytherapy sources in brachytherapy dosimetry publications and brachytherapy TPS. The TG-43 formalism applied to low-energy sources has the following advantages:

(1) Dosimetric modeling of seeds using the point-source approximation is facilitated by averaging dose anisotropy over all solid angles. This method of calculation is used primarily for permanent prostate brachytherapy where seed orientation is not discernable in clinical practice for nonstranded applications and due to the large number of seed orientations.
(2) Accurate interpolation of the dose distribution is readily achieved because the geometric dependence of dose fall-off as a function of radial distance r and polar angle θ is accounted for. This allows the use of a limited dataset while providing for robust dose calculation.
(3) An analytic, uniform approach to brachytherapy dose calculation is readily available, thereby promoting consistent clinical practice worldwide.

The TG-43 formalism[1,2] assumes a water medium with superposition of single source dose distributions, no inter-source attenuation (ISA) effects, and full scatter conditions (infinite or unbounded water medium) at dose calculation points-of-interest (POIs). Partial scatter conditions can potentially be accommodated through the use of appropriate correction factors.[29–32] This approximation of realistic clinical conditions is pertinent for both low-energy and high-energy brachytherapy applications and is discussed in detail by Rivard *et al.*[33,34]

Variable tissue composition has a larger influence on low-energy brachytherapy source dosimetry than for high-energy sources due to the photoelectric effect and its high cross section at low energies. However, the effect of scatter conditions is more important for high-energy brachytherapy dosimetry. For low-energy brachytherapy, mostly conducted as prostate implants, the surrounding tissue is adequate to provide full scatter conditions. In contrast, high-energy brachytherapy implants vary from those deeply positioned (e.g., gynecological) to surface applications (e.g., skin), with scatter significantly influencing dose calculations at clinically relevant POIs. It is not clear whether a simple modification of the current TG-43 formalism can account for partial radiation scatter conditions utilizing the current TG-43 based TPS. Alternatively, new dose calculation algorithms that correct for partial radiation scatter conditions are emerging.

As for low-energy brachytherapy sources, especially those used in multisource LDR implants, ISA effects are also present for high-energy LDR sources such as ^{192}Ir and ^{137}Cs. However, the clinical trend in the high-energy source domain is that HDR and PDR are more prevalent than the LDR procedures.

One important limitation of current TPS dose calculation tools is the near-universal neglect of applicator shielding. For example, doses to the rectal and bladder walls are generally not accurately calculated for gynecological implants, and subsequently the reported doses associated with toxicities are incorrect. Correction methods[35,36] were developed based on attenuation values that were experimentally obtained, giving reasonable values in specific clinical applications such as shielded cylinders.[37] Shielding is also present on some vaginal applicators to protect the healthy vagina at variable applicator angles. Fortunately, the use of magnetic resonance imaging (MRI) is increasing relative to computed tomography (CT) for cervical brachytherapy. With the use

of MRI-compatible applicators, imaging artifacts due to high-Z shields are mitigated. New algorithms that account for these effects are now appearing in commercial TPS as reviewed by Rivard *et al.*[33,34] and is the subject of the active AAPM Task Group 186.

The TG-43 formalism was originally applied to sources with active lengths ranging from 2 to 4 mm, while typical HDR/PDR sources have active lengths ranging from 0.5 to 5 mm, and some high-energy LDR sources such as ^{137}Cs tubes have active lengths >15 mm. Other LDR sources have a variable active length and/or curved active components like ^{192}Ir wires. An approach to dose calculation for these sources that falls within the framework of the TG-43 formalism is presently being developed by the AAPM Task Group 143.

In Sec. III.C of this report, the dependence of dosimetry parameters for high-energy sources on source active length is discussed, as is the effect of phantom size used in dose calculations and/or measurements. The latter discussion includes a methodology to convert datasets from bounded to unbounded (full scatter) conditions to compare data from different publications. The procedure used in this report for developing consensus datasets is based on this conversion methodology in some cases. Adaptation of extrapolation–interpolation techniques presented in the AAPM TG-43U1 and TG-43U1S1 reports was performed for high-energy sources. Finally, aspects specific to high-energy sources such as the electronic equilibrium region close to the source and the need for higher spatial resolution of the dose distribution close to source are addressed.

III.A. Phantom size effects

A limitation of the TG-43 formalism when applied to high energy sources is the assumption of fixed scatter conditions at calculation points, without consideration of the tissue boundaries. The TG-43 dose calculation formalism assumes an infinite scattering medium and can result in overestimation of absorbed dose at a low-density interface. In many clinical settings, the actual scatter conditions may significantly deviate from these reference conditions, leading to significant dose overestimates, e.g., when the source is near the surface of the patient. This is often the case for breast implants. For example, some breast protocols [e.g., Radiation Therapy Oncology Group (RTOG) protocol 0413] require that the dose homogeneity index include the skin dose calculations. Errors/limitations in calculating dose at shallow depths affect the dose calculation.

Serago *et al.*[38] showed a dose reduction at points close to low-density interfaces of up to 8% for HDR ^{192}Ir brachytherapy as typical for breast implants performed as a boost. Mangold *et al.*[39] showed deviations of up to 14% with measurements close to the tissue–air interface, whereas Wallner and colleagues[40] found the TPS to overestimate dose by no more than 5% at points close to the skin and lung for partial breast irradiation. However, Raffi *et al.* found TPS dose overestimations of up to 15%.[41]

Lymperopoulou *et al.*[42] reported that the skin dose overestimation can increase from 15% to 25% when ^{169}Yb is used in place of ^{192}Ir. Pantelis *et al.*[43] showed for breast implants at 2–5 cm depths with Monte Carlo (MC) radiation transport methods that the TPS overestimates by 5%–10% the isodose contours lower than 60% of the prescribed dose. Other extreme clinical situations are superficial implants involving shallow clinical target volume (CTV) irradiations, or intraoperative brachytherapy for which specialized applicators have been designed. In the latter situation, Raina *et al.*[44] showed differences of up to 13% between the dose calculated for actual and full scatter conditions in the surface tissue layer. In practice, this difference can be minimized by adding bolus, but this may not be clinically beneficial.

TPS calculations are based on interpolation over stored two-dimensional (2D) water dose rate tables which assume cylindrically symmetric sources and applicators, a uniform water-equivalent medium, and negligible ISA effects. Usually, these dose rate tables consist of TG-43 parameter values or away-along dose rate tables. In principle, it seems logical that the tables include larger distances to avoid extrapolation. Although these larger distance values are not often clinically significant, accurate data are useful for dose calculations to radiosensitive anatomical structures outside the CTV, especially when the patient has undergone external beam radiotherapy. For low-energy brachytherapy dosimetry, the TG-43U1 report[2] recommended that the radial dose function $g(r)$ extends to 7 cm for ^{125}I and to 5 cm for ^{103}Pd, which correspond to values of approximately 0.5% and 0.3% of the dose rate at 1 cm, respectively. Also in the TG-43U1 report, recommendations for good practice for MC dosimetry included determination of the dose distribution for $r \leq 10$ cm, with at least 5 cm of backscatter material for ^{125}I and ^{103}Pd. As will be justified below for high-energy sources, the recommended range for $g(r)$ is $r \leq 10$ cm.

Another issue is whether the TG-43 dosimetry parameters and the dose rate tables used by the TPS should be obtained with full scatter conditions for the complete range of distances. This issue is related to the appropriate phantom size to be used in MC calculations (henceforth labeled with "MC" subscript) or experimental purposes (henceforth labeled with "EXP" subscript), in order to establish the reference dose rate distributions used as input and benchmark data for TPS clinical dosimetry. For high-energy sources, an effectively unbounded spherical phantom radius R of 40 cm is recommended to promote uniformity of dose calculations for $r < 20$ cm, since it is not possible to cover all applications that move from superficial to deeper implants by selecting a smaller phantom size. Another issue to consider is the promise of new TPS algorithms to solve traditional calculation limitations such as tissue heterogeneities, patient and applicator scatter of radiation, intersource effects, and shielding corrections. These new algorithms will be discussed in Sec. V.

Phantom size is well known to be an important consideration in brachytherapy dosimetry. Ellet[45] studied boundary effects for photon source energies ranging from 0.03 to 2.75 MeV by comparing the dose in water spheres of radius $R = 10$, 20, 30, and 40 cm with the dose in an unbounded medium. Doses were observed to be within 5% of the values in an unbounded medium at distances of more than one

mean free path from the interface (citing a mean free path of 2.19 cm for an energy of 0.03 MeV, 9.10 cm for 0.364 MeV, 11.7 cm for 0.662 MeV, and 17.3 cm for 1.46 MeV). Williamson[46] compared MC calculations for ^{192}Ir assuming an unbounded water phantom and a $R = 15$ cm spherical phantom with measured data from the Interstitial Collaborative Working Group for a cubic phantom of approximate size $(20 \times 20 \times 20)$ cm^3. Agreement within 5% was observed up to 5 cm from the source, but differences of 5%–10% were noted for $r > 5$ cm. Williamson and Li[47] found a difference of 12% at $r = 12$ cm from a microSelectron PDR ^{192}Ir source between the dose calculated in an unbounded water phantom and that obtained with a spherical phantom ($R = 15$ cm). Venselaar *et al.*[48] measured the influence of phantom size on dose by changing the water level in a cubic water tank for ^{192}Ir, ^{137}Cs, and ^{60}Co sources. Significant dose differences were observed between experiments with different phantom sizes. Karaiskos *et al.*[49] performed MC and thermoluminescent dosimetry (TLD) studies of the microSelectron HDR ^{192}Ir source using spherical water phantoms with $R = 10$–50 cm. They ascertained that phantom dimensions significantly affect $g(r)$ near-phantom boundaries where deviations of up to 25% were observed. They did not observe significant differences in the anisotropy function $F(r, \theta)$ for the different values of R. Other investigators have found a dose dependence on R due to the different scatter conditions.[50–56]

Perez-Calatayud *et al.*[29] presented a study where $_{MC}g(r)$ was obtained for water phantoms with $5\text{ cm} \leq R \leq 30$ cm (^{125}I and ^{103}Pd) and $10\text{ cm} \leq R \leq 50$ cm (^{192}Ir and ^{137}Cs). They showed that dose differences with respect to full scatter conditions for ^{192}Ir and ^{137}Cs sources, in the case of the most popular phantom size cited in the literature ($R = 15$ cm), reached 7% (^{192}Ir) and 4.5% (^{137}Cs) at $r = 10$ cm, but were only 1.5% (^{192}Ir) and 1% (^{137}Cs) at $r = 5$ cm. For $R = 40$ cm and ^{192}Ir or ^{137}Cs, the dose rate was equivalent to an unbounded phantom for $r \leq 20$ cm, since this size ensured full scatter conditions. For ^{125}I and ^{103}Pd, $R = 15$ cm was necessary to ensure full scatter conditions within 1% for $r \leq 10$ cm.[29] These results agree with the subsequent study by Melhus and Rivard,[30] who in addition showed that for ^{169}Yb, a radius of $R \geq 40$ cm is required to obtain data in full scatter conditions for $r \leq 20$ cm. Perez-Calatayud *et al.*[29] developed a simple expression relating values of $g(r)$ for various phantom sizes based on fits to the dose distributions for ^{192}Ir and ^{137}Cs. This expression is useful to compare published dose rate distributions for different phantom sizes and to correct $g(r)$ values for bounded media of radius $10\text{ cm} \leq R \leq 40$ cm to unbounded phantom values. Differences between corrected dose rate distributions and the corresponding MC results for a given phantom size were less than 1% for $r < R - 2$ cm if $R < 17$ cm and for $r < 15$ cm if $R \geq 17$ cm. At larger distances r, the fitted dose rate distribution values did not lie within the 1% tolerance. These relations were based on the previous result that for $R = 40$ cm the dose rate was equivalent to an unbounded phantom for $r \leq 20$ cm. Some dosimetry investigators have used a 40-cm-high cylindrical phantom with a 20-cm radius in their MC studies. It has been shown that this phantom is equivalent to a spherical phantom with a 21-cm radius.[29] The expression developed by Perez-Calatayud *et al.*[29] is not applicable to the outer 2 cm of this phantom.

To date, most published MC high-energy brachytherapy dosimetry studies have been performed in a water sphere with $R = 15$ cm,[10,46,55,57–61] a cylindrical phantom of size $40\text{ cm} \times 40\text{ cm}$,[62–69] or a sphere with $R = 40$ cm.[70–72] Granero *et al.*[31] developed correction factors expressed as fourth-degree polynomials to transform $g(r)$ data for ^{192}Ir and ^{137}Cs obtained using commonly published phantom sizes into approximate $g(r)$ values for unbounded phantom conditions, with agreement within 1%.[29–31] These correction factors are given in Table II.

In this joint AAPM/ESTRO report, $g(r)$ values from published studies obtained under bounded conditions have been transformed to full scatter conditions with the correction factors in Table II. So, with these relationships, TPS users can transform data from the literature obtained in a bounded medium to input data in full scatter conditions for $r \leq 15$ cm.

When different datasets obtained with different phantom sizes are compared, the boundary scatter defect must be taken into account. At $r = 1$ cm, full scatter exists within 0.5% for all studies, hence the dose rate constant Λ is directly comparable in all cases. As noted in the literature,[49] $F(r, \theta)$ has been shown to be nearly independent of phantom

TABLE II. Polynomial coefficients of the correction factors (CF) used to quantitatively compare bounded to unbounded radial dose functions for common phantom shapes and sizes. CF was fitted as $CF = C_0 + C_1\, r + C_2\, r^2 + C_3\, r^3 + C_4\, r^4$. These coefficients have been obtained by Granero (Refs. 29 and 162) in a re-evaluation of their study which takes into account that with the coefficients in the original publication, $g(r = 1\text{ cm})$ was not exactly one.

	Sphere $CF = \frac{g(R_{sph} = 40\text{ cm}, r)}{g(R_{sph} = 15\text{ cm}, r)}$ $1\text{ cm} \leq r \leq 15\text{ cm}$		Cylinder $CF = \frac{g(R_{sph} = 40\text{ cm}, r)}{g(R_{cyl} = 20\text{ cm}, r)}$ $1\text{ cm} \leq r \leq 20\text{ cm}$		Cube $CF = \frac{g(R_{sph} = 40\text{ cm}, r)}{g(R_{cube} = 15\text{ cm}, r)}$ $1\text{ cm} \leq r \leq 15\text{ cm}$	
CF parameter	^{192}Ir	^{137}Cs	^{192}Ir	^{137}Cs	^{192}Ir	^{137}Cs
C_0 (dimensionless)	1.002	1.001	1.001	1.001	1.002	1.001
C_1 (cm^{-1})	-3.52×10^{-3}	-2.28×10^{-3}	-1.23×10^{-3}	-1.09×10^{-3}	-3.27×10^{-3}	-1.85×10^{-3}
C_2 (cm^{-2})	2.06×10^{-3}	1.24×10^{-3}	3.00×10^{-4}	4.02×10^{-4}	1.31×10^{-3}	8.89×10^{-4}
C_3 (cm^{-3})	-2.39×10^{-4}	-1.35×10^{-4}	-2.40×10^{-5}	-3.93×10^{-5}	-2.46×10^{-4}	-9.45×10^{-5}
C_4 (cm^{-4})	1.38×10^{-5}	7.78×10^{-6}	1.90×10^{-6}	2.08×10^{-6}	8.50×10^{-6}	5.23×10^{-6}

TABLE III. Interpolation and extrapolation recommendations for high-energy (*low-energy*) (Ref. 3) brachytherapy sources for the line-source approximation.

Parameter	$r < r_{min}$ Extrapolation	$r_{min} < r \leq r_{max}$ Interpolation	$r > r_{max}$ Extrapolation
$g_L(r)$	Nearest neighbor or zeroth-order extrapolation *(Ditto)*	Linear (*log-linear*) using datapoints immediately adjacent to the radius of interest	Linear using data of last two tabulated radii (*single exponential function based on fitting $g_L(r)$ datapoints for the furthest three r values*)
$F(r,\theta)$	Nearest neighbor or zeroth-order extrapolation *(Ditto)*	Bilinear (*bilinear*) interpolation method for $F(r,\theta)$ *(Ditto)*	Nearest neighbor or zeroth-order *r*-extrapolation *(Ditto)*

size. Consequently, research has focused on $g(r)$. Anagnostopoulos *et al.*[54] proposed a calculation algorithm based on the scatter-to-primary ratio to relate $g(r)$ for one spherical phantom size to $g(r)$ for other R values. Russell *et al.*[73] proposed another dose calculation algorithm based on primary and scatter dose separation involving parameterization functions which could also be used to correct the scatter defect. Melchert *et al.*[74] developed a novel approach inspired by field theory to calculating the dose decrease in a finite phantom for ^{192}Ir point source(s).

III.B. Dose calculation grid size and interpolation accuracy

Traditionally, brachytherapy TPS utilized analytical methods such as the Sievert integral[75] to generate dose rate tables for conventional LDR brachytherapy sources such as ^{137}Cs tubes and ^{192}Ir wires. These systems then utilized the same method for data interpolation to calculate dose for clinical implants. However, current TPS used for HDR, PDR, and LDR brachytherapy allow direct introduction of tabulated dosimetry parameters from the literature. Some of this information is included in the TPS default dosimetric data supplied by the TPS manufacturer. In some systems, values of the dosimetry parameters are manipulated from one format to another in order to match the dose calculation algorithm used by the system. Examples include changing from rectangular to polar coordinates, using different mathematical functions to fit and smooth tabulated data, and extrapolating data outside of the available data range. Therefore, it is desirable that TG-43 consensus data be presented with adequate range and spatial resolution in order to facilitate input and verification of the accuracy of the TPS dose calculation algorithm.

A review of the published data on dosimetry parameters for various high-energy brachytherapy sources indicates that different authors have used a variety of spatial and angular increments and ranges in their dosimetric procedures. Therefore, a clear methodology for interpolation or extrapolation of the published data may be required to determine dose rate distributions at spatial locations not explicitly included in the published data. The AAPM TG-43U1 report[2] provided guidelines for interpolation and extrapolation of one-dimensional (1D) and 2D dosimetry parameters. The 2007 supplement (i.e., TG-43U1S1)[3,4] included further clarification and modifications of the interpolation and extrapolation techniques in order to make these procedures more accurate. Unlike for low-energy sources, the 1D approximation for high-energy brachytherapy source dosimetry is not recommended, based on the smaller number of sources generally used, known source orientation(s), and the method used for source localization. In this section, the parameter range and spatial resolution, as well as interpolation and extrapolation recommendations are provided. The AAPM TG-43U1S1 recommendations for interpolation and extrapolation of 2D dosimetry are summarized in Table III.

With respect to the angular resolution for $F(r,\theta)$, 10° steps were generally recommended by the TG-43U1 report, although 1° steps near the source long axes may be needed to have 2% interpolation accuracy over the range of angles. For radial resolution, TG-43U1 recommended $F(r,\theta)$ be tabulated at 0.5, 1, 2, 3, and 5 cm for ^{103}Pd and also at 7 cm for ^{125}I. For $g_L(r)$, the recommended range was the same as the $F(r,\theta)$ radial range, but no specifics were provided concerning radial resolution. However, both the TG-43U1 and TG-43U1S1 reports required that the $g_L(r)$ radial resolution permit log-linear interpolation and fitting with ±2% accuracy. The radial coordinate mesh recommended by HEBD is similar to that recommended by LEBD for low-energy sources. However, a maximum range of 10 cm is indicated since the dose rate here is about 1% of the value at r_0 due to the more uniform $g(r)$ behavior for high-energy sources. The minimum r value for the high-energy consensus datasets will also differ based on consideration of radiological interactions. Some differences between low-energy and high-energy source dosimetry include the following:

(1) From a clinical perspective, there is more concern with dose accuracy along the longitudinal axis region of the source for high-energy sources as there is a larger proportion of treatments in which the dose along this axis is included in the prescription (e.g., dome applicators for hysterectomyzed patients, endometrial applicators) than for low-energy brachytherapy. In contrast, permanent prostate implants use many seeds, and the longitudinal axis region is less relevant because of volume averaging and the contribution of many seeds with variable axis orientation.[76–78]
(2) For high-energy sources, MC-based dosimetry is the predominant method in part due to its robustness at these energies. When measurement conditions are subject to challenges (associated with detector energy response, detector radiation sensitivity, positioning uncertainty, detector volume averaging, influence of radiation scatter

conditions on results, etc.), the role of experimental dosimetry for high-energy brachytherapy may be more limited than MC-based dosimetry. Experiment may primarily serve to validate MC and to obtain Λ for averaging with MC-derived values since MC is primarily used to determine $F(r,\theta)$ and $g_L(r)$ for high-energy sources. Consequently, range and spatial resolution limitations are not of concern for MC methods and high-energy brachytherapy source dosimetry. However, caution must be taken at close distances if electron transport and electron emissions are not considered.

A study by Pujades-Claumarchirant *et al.*[79] has been performed for high-energy sources to check methods of interpolation/extrapolation that allow accurate reproduction of $g_L(r)$ and $F(r,\theta)$ from tabulated values, including the minimum number of entries for $g_L(r)$ and $F(r,\theta)$ that allow accurate reproduction of dose distributions. Four sources were studied: ^{192}Ir, ^{137}Cs, ^{60}Co, and a hypothetical ^{169}Yb source. The r mesh was that typically used in the literature: 0.25, 0.5, 0.75, 1, and 1.5 cm, and for 2–10 cm in 1 cm steps, adding the point $r_{gmax}=0.33$ cm for ^{60}Co and $r_{gmax}=0.35$ cm for ^{137}Cs near the maximum value $g(r_{gmax})$. For $F(r,\theta)$, the entries for polar angles close to the source long axis were evaluated at four different step sizes: 1°, 2°, 5°, and 10°. For $g_L(r)$, linear interpolations agreed within 0.5% compared with MC results. The same agreement was observed for $F(r,\theta)$ bilinear interpolations using 1° and 2° step sizes.

Based on the Pujades-Claumarchirant *et al.* study,[79] minimum polar angle resolutions of 2° (0° to 10° interval), 5° (10° to 30° interval), and 10° (30° to 90° interval) with the addition of corresponding supplementary angles as applicable if dosimetric asymmetry about the transverse plane is >2% are recommended. Further, use of bilinear and linear interpolation for $F(r,\theta)$ and $g_L(r)$, respectively, is recommended since log-linear interpolation is not a significant improvement over linear $g(r)$ interpolation for high-energy sources.[79]

$F(r,\theta)$ and $g_L(r)$ extrapolation for $r>10$ cm could be performed by linear extrapolation from the last two tabulated values. However, because of the inverse square law, the dose rate is very low and not clinically relevant. If dosimetric accuracy is required for $r>10$ cm, for example, to calculate organ-at-risk dose, users must refer to the original MC publication.

In contrast with low-energy brachytherapy dosimetry, extrapolation for high-energy sources for $r\leq r_{min}$ is complicated. Electronic equilibrium is reached within a distance of 0.1 mm from the capsule for a low-energy source due to the short electron range. Thus, it can be assumed that collisional kerma is equal to absorbed dose everywhere. For high-energy brachytherapy dosimetry, the region of electronic disequilibrium near the source and the contribution from emitted electrons can be important issues and are not considered in most MC publications.

In a recent study of Ballester *et al.*,[28] MC calculations scoring dose and taking into account electronic emission are compared with MC calculations scoring collisional kerma at short distances for spherical sources with active and capsule materials mimicking those of actual sources. Electronic equilibrium is reached to within 1% for ^{192}Ir, ^{137}Cs, ^{60}Co, and ^{169}Yb at distances greater than 2, 3.5, 7, and 1 mm from the source center, respectively. Electron emissions are important (i.e., >0.5% of the total dose) within 3.3 mm of ^{60}Co and 1.7 mm of ^{192}Ir source centers but are negligible over all distances for ^{137}Cs and ^{169}Yb. Ballester *et al.*[28] concluded that electronic equilibrium conditions obtained for spherical sources could be generalized to actual sources, while electron contributions to total dose depend strongly on source dimensions, material composition, and electron spectra. Consequently, no extrapolation method can accurately predict near-source dose rate distributions because they depend on both the extent of electronic disequilibrium and the electron dose at distances closer than the minimum tabulated results.

However, tabular data containing voids close to and inside the source should not be presented, and adoption of the TG-43U1S1 extrapolation method for $r<r_{min}$ using the nearest neighbor data for $g_L(r)$ is recommended until such time as future studies generate data for this region. For $F(r,\theta)$, HEBD decided to take advantage of partial data and proposed the following approach as a compromise to maintain consistency with the TG-43U1S1 report: fill in missing data for partially complete $F(r,\theta)$ tables using linear extrapolation in polar angle for fixed r based on the last two tabulated values and use zeroth order (nearest neighbor) extrapolation for $r<r_{min}$ as recommended in the AAPM TG-43U1S1 report. It is emphasized that extrapolated values are only included for the purpose of providing complete data tables as required by some TPS. Dose data outside the source obtained from these extrapolated values could be subject to large errors due to beta (electron) contribution, kerma versus dose differences, and linear extrapolation limitations. Data inside the source are only provided for TPS requirements and they do not have any physical meaning. These extrapolated values should be used with caution in clinical dosimetry because potentially large errors exist; this scenario is different from the low-energy case of TG-43U1S1 where differences between MC calculated and extrapolated doses are generally minimal.

The formalisms of the 1995 (Ref. 1) and 2004 (Ref. 2) TG-43 reports were based on dosimetric characteristics of seed models for brachytherapy sources containing ^{125}I, ^{103}Pd, and ^{192}Ir having nearly spherical dose distributions given their relatively large ratios of radial distance to active source length. Therefore, it is appropriate to use the polar coordinate system to describe dosimetric parameters around these sources. However, several investigators have shown that this approach fails when the active length is greater than the distance to the POI.[80–83] Alternatively, the advantage of using the cylindrical coordinate system (Y, Z) based TG-43 formalism has been demonstrated for dose calculations around elongated brachytherapy sources by Patel *et al.*[84] and Awan *et al.*[85] Detailed comparisons between the polar and cylindrical coordinate based formalisms are given by Awan *et al.*[85] and the forthcoming AAPM TG-143 report. In these comparisons, it has been demonstrated that the basic dosimetry parameters in the two coordinate systems are very similar. The main difference is in the $F(r,\theta)$ definition

$$F_{\mathrm{pol}}(r,\theta) = F_{\mathrm{cyl}}(Y,Z)\frac{g(Y)}{g(r)}. \quad (1)$$

However, the cylindrical coordinate system based formalism provides a more accurate tool for interpolation and extrapolation of dosimetry parameters for a given source, since the spatial sampling better approximates the cylindrical radiation dose distribution. For the high-energy sources considered in this report, the active length up to 1.5 cm, the TG-43 approach using polar coordinates also applies well if adequate mesh resolution is utilized, and then it is recommended here. Dosimetric considerations (source calibration, TG-43 parameter derivation, TPS implementation, etc.) for sources with larger active lengths and curved lengths are being evaluated by AAPM TG-143.

III.C. Dosimetry parameter dependence on active length

The dosimetric properties of a brachytherapy source depend upon the geometry and material composition of the source core and its encapsulation. For high-energy photon emitters such as ^{192}Ir, the material composition dependence is much less pronounced than that for low-energy emitters such as ^{125}I.[1,2] This leads to a greater similarity of TG-43 dosimetry parameters for high-energy sources containing the same radionuclide and having comparable dimensions than for low-energy sources. For example, a study by Williamson and Li comparing the original Nucletron microSelectron Classic HDR ^{192}Ir source with the PDR source and the old VariSource HDR ^{192}Ir source revealed that they have nearly identical Λ values, and their $g_L(r)$ data agreed within $\sim$1% for $r > 0.5$ cm.[47] Selected reports from the literature describing such similarities for ^{192}Ir, ^{137}Cs, and ^{60}Co brachytherapy sources are summarized below.

Wang and Sloboda compared the transverse plane dose distributions for four ^{192}Ir brachytherapy sources (Best Medical model 81-01, Nucletron microSelectron HDR and PDR ^{192}Ir sources, Varian VariSource HDR) and five hypothetical ^{192}Ir cylindrical source designs using the EGS4 MC code.[86] The transverse-plane dose rate and air-kerma strength s_K per unit contained activity were calculated in a spherical water phantom of $R = 15$ cm and a dry air sphere of 5 m diameter, respectively, to study the influence of the active length L and R on these quantities. For $r \geq 4L$, the transverse-plane dose rate and s_K depended on R but not on L and were proportional to the corresponding quantities for an unencapsulated point source to within 1%. When the transverse-plane dose rate was normalized to s_K, differences in the dose rate profiles between the various sources disappeared for $r \geq 4L$. For $r < 4L$, the transverse-plane dose rate and s_K were dependent on both R and L, and the geometry function $G(r,\theta)$ was the principal determinant of the shape of the normalized dose rate profile. Photon absorption and scattering in the source had a considerably smaller influence and partly compensated one another, whereas differences in the photon energy fluence exiting the source were not of sufficient magnitude to influence absorption and scattering fractions for the dose rate in water. Upon calculating Λ and $g_L(r)$ for the four real sources using $G_L(r,\theta)$ (except for the microSelectron PDR source for which the particle streaming function $S_L(r,\theta)$ was used),[87] observed differences in Λ were explained on the basis of differences in $G_L(r,\theta)$ and source core diameter d. For $r \geq 1$ cm, $g_L(r)$ were similarly identical within 1%, and small differences for $r < 1$ cm were caused by varying degrees of photon absorption and scattering in the sources.

Karaiskos and colleagues obtained TG-43 dosimetry parameters for ^{192}Ir wire of active lengths $0.5\,\mathrm{cm} \leq L \leq 12\,\mathrm{cm}$ and internal diameters $d = 0.1$, 0.3, and 0.4 cm using an in-house MC code and a modified Sievert-integral method.[88] They employed $G_L(r,\theta)$, as they had previously shown it to introduce differences of <1% compared with the particle streaming function with $S_L(r,\theta)$ for $r > L/2$ and similarly small differences for clinically relevant wire lengths of 4–6 cm for $r \leq L/2$.[89] With the line source approximation, a scaling relation holds between geometry functions for sources of different active lengths L and L'

$$\frac{G_L(r,\theta)}{G_{L'}(r',\theta)} = \frac{\beta/Lr\sin(\theta)}{\beta/L'r'\sin(\theta)} = \frac{L'r'}{Lr} = \left(\frac{L'}{L}\right)^2, \quad (2)$$

where β is the angle subtended by the active length with respect to the calculation point $P(r,\theta)$ and $r'/r = L'/L$ due to similar triangles. Karaiskos *et al.* subsequently determined that Λ for wires of equal length was only weakly dependent on d, differences being <3%. Based on this observation, they showed that Λ for any ^{192}Ir source of active length L can be related to that of a reference source of active length L_{REF} using

$$\frac{\Lambda_L}{G_L(r_0,\theta_0)} = \frac{\Lambda_{L_{\mathrm{REF}}}}{G_{L_{\mathrm{REF}}}(r_0,\theta_0)}. \quad (3)$$

This relation was found to hold to within 2% for $L_{\mathrm{REF}} \leq 5$ cm and to <3% when realistic HDR ^{192}Ir brachytherapy sources were considered. Thus, this relation can be used to calculate Λ for ^{192}Ir wires of arbitrary length and may also be useful to check the consistency of EXP- or MC-derived Λ values for other source models. The investigators also determined that $g_L(r)$ for $0.2\,\mathrm{cm} \leq r \leq 10\,\mathrm{cm}$ was independent of L to <2% and of d to <3%. They concluded that MC calculated values of $g_L(r)$ for L set to 5 cm were adequate for most any length. Sievert-integral[75] calculated $F(r,\theta)$ values decreased as d increased but by no more than 3% over all radial distances examined. MC-calculated $F(r,\theta)$ values were nearly unity for polar angles $30^\circ \leq \theta \leq 90^\circ$ for all r and L. However, a strong dependence on both r and L was observed for $\theta < 30^\circ$. This was due in part to the fact that the main dose contributor to a point close to the source is the source segment closest to that point, whereas for points further away from the source, the entire source length contributes and $F(r,\theta)$ decreases for polar angles close to the long axis due to oblique filtration within the source structure. van der Laarse *et al.*[82] further developed these ideas to create a new method, named the two length segmented method (TLS), which models brachytherapy dose parameters for ^{192}Ir wires of any length and shape using dose parameters for straight wire segments 0.5 and 1.0 cm in length. The resultant dose

rate distributions around straight and U-shaped wires agreed better with MC calculations than those obtained with the point segmented source method, line segmented source method, or Karaiskos *et al.*[88] dose calculation models.

Papagiannis *et al.*[90] performed a dosimetry comparison of five HDR ^{192}Ir sources (old and new Nucletron microSelectron, old and new Varian VariSource, and the "Buchler" source), the LDR ^{192}Ir seed (Best Medical model 81-01), and the LDR ^{192}Ir wire source (Eckert & Ziegler BEBIG GmbH) in a $R = 15$ cm liquid water sphere using their own MC code. They tested the validity of Eq. (3) relating Λ using the reference geometry function $G_L(r_0, \theta_0)$, with a point ^{192}Ir source serving as the reference, and found that the ensuing expression

$$\Lambda = 1.12 \times G_L(r_0, \theta_0) \ [\text{cGy h}^{-1} \text{U}^{-1}] \quad (4)$$

yielded Λ with differences $<2\%$ at reference radial distances of 1 and 2 cm for any of the ^{192}Ir source designs. The value 1.12 cGy·cm^2·h^{-1}·U^{-1} corresponds to Λ for an ^{192}Ir point source.[91] These investigators also found that $g(r)$ for all sources except the Buchler source were in close agreement for distances $0.1 \text{ cm} \leq r \leq 5$ cm and lay within 2% of $g(r)$ for a point ^{192}Ir source. The Buchler source presented a slight increase at radial distances $r < 0.5$ cm, possibly arising from hardening of the emerging photon spectrum due to its larger source core diameter. All sources were observed to exhibit non-negligible anisotropy, with $F(r, \theta)$ values being strongly dependent on source geometry. $F(r = 0.2 \text{ cm}, \theta)$ did not significantly differ from unity over all polar angles for all sources since the main contributor to the dose rate at P(r, θ) is the source segment closest to that point.

Using their established MC code, Karaiskos *et al.*[55] compared the dosimetry of the old and new Nucletron microSelectron PDR ^{192}Ir source designs in a $R = 15$ cm liquid water sphere. They found the Λ to be identical to each other and to that for a point source to within statistical uncertainties of $\sim$0.5% and explained the result in terms of Eq. (3) on the basis of the short L of 0.6 and 1.0 mm for the sources. Using $S_L(r, \theta)$,[87,89] the $g_L(r)$ values were found to be identical within 1% to those obtained using the linear source approximation $G_L(r, \theta)$ over the distance interval $0.1 \text{ cm} \leq r \leq 14$ cm. When the point source geometry function r^{-2} was used, differences $>1\%$ were observed only for $r < 0.3$ mm. The $F(r, \theta)$ for both source designs was found to be significant only at polar angles close to the longitudinal source axis ($\theta < 30^\circ$ and $\theta > 150^\circ$) and to be greatest within these angular intervals at intermediate radial distances for reasons discussed previously.[90] The new design presented increased $F(r, \theta)$ up to 10% at polar angles near $\theta = 0^\circ$ (distal end of the source) as a result of its longer active core.

Casal *et al.*[63] and Perez-Calatayud *et al.*[67] calculated the dose rate distributions around three different LDR ^{137}Cs sources (Amersham models CDCS-M, CDC-1, and CDC-3) in a 40 cm high, 40 cm diameter water cylinder using the GEANT3 MC code.[63,67] TG-43 dosimetry parameters were obtained using $G_L(r, \theta)$. For the model CDCS-M source, they found $\Lambda/G_L(r_0, \theta_0)$ constancy, 1.05 cGy cm^2/(h U), within 0.9% for the corresponding ratio of the model CDC-J source, which had the same encapsulation but a 1.5 mm shorter active length. The latter ratio was determined from MC data published by Williamson.[57] For the CDC-1 and CDC-3 sources, the values of $\Lambda/G_L(r_0, \theta_0)$ differed by only 0.1%. For all three sources, $g_L(r)$ was no more than 1% different from the normalized Meisberger polynomial for $0.5 \text{ cm} \leq r \leq 10$ cm.[92] The $F(r, \theta)$ results corresponded to the varying self-attenuation associated with the different source designs.

Papagiannis *et al.*[10] compared the dosimetry of three HDR ^{60}Co sources containing two active pellets in contact or spaced 9 or 11 mm apart, used in the Ralstron remote afterloader. MC calculations for a $R = 15$ cm water sphere were done with the group's own simulation code and included electron transport for $r < 0.5$ cm. The dose rate distribution around the source having the pellets in contact closely resembled that for an unencapsulated ^{60}Co point source. As r increased sufficiently for the point-source approximation to apply, the dose rate distributions for the other two designs also conformed to that of a point source, presenting only minor spatial dose anisotropy close to the source long axis. The main influence on Λ once again proved to be the spatial distribution of activity, represented by $G_L(r, \theta)$, for reasons similar to those cited for commercial ^{192}Ir source designs. Consequently, the relation

$$\Lambda = 1.094 \times G_L(r_0, \theta_0) \ [\text{cGy h}^{-1} \text{U}^{-1}], \quad (5)$$

where $\Lambda = 1.094$ cGy·cm^2·h^{-1}·U^{-1} for a ^{60}Co point source[91] was used to obtain Λ values for realistic ^{60}Co sources within $\pm 2\%$. Using the $G_L(r, \theta)$, $g_L(r)$ also agreed within 2% for $0.5 \text{ cm} \leq r \leq 15$ cm. However, using $G_P(r)$, differences of up to 28% were noted. $F(r, \theta)$ for all three source designs calculated using $G_L(r, \theta)$ indicated that dose anisotropy was negligible for $r \leq 1$ cm and was only evident for $r > 1$ cm at points close to the source drive wire ($\theta \sim 180^\circ$).

In summary, the dosimetry for $r < 2$ cm is primarily determined by the contained activity distribution for high-energy photon-emitting brachytherapy sources. The influence of photon attenuation and scattering in the source core and capsule is comparatively smaller in magnitude and is further diminished when $\dot{D}(r, \theta)/S_K$ is calculated. As a consequence, Λ for commercially available ^{192}Ir, ^{137}Cs, and ^{60}Co brachytherapy sources containing the same radionuclide are equal (within a few percent) to the product of Λ for an unencapsulated point source and $G_L(r, \theta)$. Corresponding $g_L(r)$ values for sources containing the same radionuclide that have been extracted from dose distribution data using $G_L(r, \theta)$ also agree to within a few percent over the radial interval $0.3 \text{ cm} \leq r \leq 10$ cm. Self-attenuation in the active core and surrounding encapsulation characterizing each source design influences $F(r, \theta)$.

IV. CONSENSUS DATASET METHODOLOGY

The source models reviewed in this report satisfy the AAPM/ESTRO recommendations published by the HEBD in Li *et al.*[17] The consensus methodology for these high-energy sources is similar to that recommended for low-energy sources by LEBD[2] but has been adapted for high-energy sources. According to these HEBD recommendations,[17] there are two source categories:

(1) For conventional encapsulated sources similar in design to existing or previously existing ones, a single dosimetric study published in a peer-reviewed journal is sufficient. MC or experimental dosimetry (or both) methods may be used.
(2) For all other high-energy sources, at least two dosimetric studies published in peer-reviewed journals by researchers independent of the vendor, one theoretical (i.e., MC-based) and one experimental, are required.

In the present report, all ^{192}Ir and ^{137}Cs sources are categorized as "conventional encapsulated sources". While not commercially available at the time of publication of the current recommendations, HDR ^{60}Co sources are also included in this first category. The remaining radionuclides, ^{169}Yb and ^{170}Tm, fall into the second category.

Similarly to the AAPM TG-43U1 report, appropriate publications reporting single source dosimetry were evaluated. For each source model, a single TG-43U1 consensus dataset ($_{CON}L$, $_{CON}\Lambda$, $_{CON}g_L(r)$, $_{CON}F(r,\theta)$ including data up to $r = 10$ cm) was derived from multiple published datasets as detailed below. If items essential to critical evaluation were omitted from a publication, the authors were contacted for information or clarification.

The methodology followed to derive a consensus dataset was as follows:

(a) The peer-reviewed literature was examined to identify candidate datasets for each source model that were derived either from measurements or MC studies and that followed the guidelines of the TG-43U1 (Ref. 2) and HEBD report.[17] The quality of each dataset was then examined, taking into consideration salient factors such as data consistency, MC code benchmarking, etc.
(b) The value of $_{CON}\Lambda$ was obtained from MC data for the following reasons: MC results uncertainties were always less than the measured uncertainties. Frequently, only MC results were available without measured results, and the variations of $_{MC}\Lambda$ were typically less than the MC uncertainties for high-energy sources. The $_{EXP}\Lambda$ values have been in good agreement with MC. For example, Daskalov *et al.*[93] showed that $_{EXP}\Lambda$ for the mHDR-v2 source agreed with $_{MC}\Lambda$ to within 2%. The value from Meisberger *et al.*[92] agreed to within 0.3%.
(c) In most cases, $_{CON}g_L(r)$ and $_{CON}F(r,\theta)$ were taken from a single MC study. When available, experimental studies were used to validate $_{MC}g_L(r)$ and $_{MC}F(r,\theta)$. Data selection was based on highest spatial resolution (r and θ), largest radial range, and highest degree of smoothness. Even though some selected published data used the point-source approximation or the particle streaming function,[87,89] that data were transformed for use with the linear geometry function.
(d) Values of $_{CON}g_L(r)$ were determined for full scatter conditions as described in Sec. III.A and for values of $r \leq 10$ cm.
(e) As described in Sec. III.B, a candidate publication's $g_L(r)$ and $F(r,\theta)$ data were examined to determine whether the values at short distances took into account a possible lack of electronic equilibrium (if collisional kerma was simulated instead of absorbed dose) and included any non-negligible beta component. This issue should be addressed in the publication, because of the dependence of $g_L(r)$ at short distances on capsule material and thickness. If it was not, data at affected small r were removed. Future publications need to explicitly consider these electronic dose effects.
(f) If the liquid water phantom used in a selected MC calculation did not generate $g_L(r)$ under full scatter conditions for $r \leq 10$ cm, the data were corrected to unbounded conditions as justified in Sec. III.A according to the polynomial corrections in Table II. These modified values are indicated using [brackets] in the consensus dataset tables.
(g) If some consensus dataset values were selected for inclusion from a nonideal candidate dataset in order to cover a larger range of distances and angles, these data are *italicized* as was done in the TG-43U1 report.
(h) For sources included in this report, AAPM/ESTRO recommends the 2D brachytherapy dosimetry formalism and 2D tables: $F(r,\theta)$, $G_L(r,\theta)$, and $g_L(r)$. Source orientation is considered in all currently available TPS for nonpermanent implants. From the clinical point of view, source orientation is more relevant along and near the source long axis for high-energy dosimetry. There are a significant number of treatments in which the long-axis dose close to the first source position is included in the target prescription (i.e., gynecological applications). In contrast, it is less relevant for low-energy permanent implants with many seeds, where source orientation averaging is adequate.
(i) Data interpolation of $g_L(r)$ and $F(r,\theta)$ is needed for dataset comparison and within consensus tables. In the TG-43U1 report,[2] interpolations were required to yield $\leq$2% error. For the high-energy regime, this should be reduced to $\leq$1%. Interpolated data are indicated by **boldface** and follow the methodology described in Sec. III.B.
(j) Similar to TG-43U1S1, $_{CON}g_L(r)$ values were tabulated on a common mesh for all source models of the same radionuclide. In contrast, the mesh used for $_{CON}F(r,\theta)$ follows the one(s) included in the selected publication(s). $_{CON}g_L(r)$ starts from the minimum available distance and continues with the common mesh [0.25, 0.5, 0.75, 1, 1.5, 2, 3, 4, 5, 6, 8, 10] cm, according to Sec. III.B, to ensure linear-linear interpolation accuracy within 1%. Further, for the case of ^{60}Co, high-resolution radial distance data are required in the vicinity of the source. The minimum r-value in the consensus dataset may be different as a function of the source model considered, physical processes in play based on photon energy, and the method used to simulate or measure dose in this region.
(k) According to Sec. III.B, the recommended angular mesh for $_{CON}F(r,\theta)$ is 0° to 10° (1° increments), 10° to 20° (5° increment), 20° to 160° (10° increment), 160°

to 170° (5° increment), 170° to 180° (1° increments). Consensus data were selected based on having an angular mesh closest to the recommended one.

(l) Extrapolation of consensus datasets was performed following the methodology described in Sec. III.B. Extrapolated values are underlined in dataset tables.

(m) Upon derivation of the consensus TG-43 dataset, an away-along dose rate table was obtained ($cGy \cdot h^{-1} \cdot U^{-1}$) for TPS quality assurance purposes. Range and resolution of this table are away [0, 0.25, 0.5, 0.75, 1, 1.5, 2–7 (1 cm increment)] cm and along [0, 0.5, 1, 1.5, 2–7 (1 cm increment)] cm.

(n) To provide a consistent convention for all brachytherapy sources, the angle origin is selected to be the source tip, i.e., θ is defined such that 0° is in the direction of the source tip. For the case of asymmetric LDR sources (without driven cable), the angle origin will be clearly identified for each source model. The origin of coodinates is selected to be the center of the active volume for all sources. Published data with a different angle/coordinate origin were transformed accordingly. This convention is recommended for future studies.

The criteria used to evaluate dosimetry parameters for each source were similar to those of the TG-43U1 report and are as follows:

1. Internal geometry and description of the source
2. Review of pertinent literature for the source
3. Measurement medium to liquid water medium corrections (if applicable)
4. Experimental method used
5. Geometry function used; active length assumed for the line source approximation
6. Name and version of MC code
7. MC cross-section library
8. Variance reduction techniques used (for s_K and dose in water)
9. Electron emission inclusion
10. Photon emission spectrum
11. MC benchmarking according to the HEBD prerequisites[17]
12. Phantom shape and size used in MC and EXP
13. Agreement between MC and experimental dosimetry (if applicable, according to the HEBD prerequisites)[17]

IV.A. Dose rate constant

As pointed out in TG-43U1,[2] MC and experimental studies complement one another and when combined can average out possible biases of each individual methodology. In contrast to the low-energy case, the high-energy ${}_{CON}\Lambda$ is obtained from the average of MC values alone, while available ${}_{EXP}\Lambda$ are used to validate MC. For the sources considered in this report, the ${}_{EXP}\Lambda$ agrees with the ${}_{MC}\Lambda$ to within 2%. This approach is justified because unlike for lower energy sources, the influence of source geometry on the dose distribution is less important at higher energies. It also has the advantage of utilizing the smaller uncertainties of the MC method, thus providing reduced uncertainty in the value of ${}_{CON}\Lambda$. In the case of sources within the category of "conventional encapsulated, similar to existing ones" for which just one study was available, the Λ value was compared with those for sources of similar design, first removing the geometrical dependence by forming the ratio $\Lambda/G_L(r_0,\theta_0)$, as discussed in Sec. III.C. Based on trends observed during the compilation of this report, the agreement between MC- and EXP-derived Λ values should be $\leq 1\%$.

IV.B. Radial dose function

For each source, MC and experimental $g_L(r)$ results were graphically compared. When a published study used a geometry function which was different than the simple linear geometry function, $g_L(r)$ was recomputed. Based on trends observed during the compilation of this report, the agreement between MC and EXP $g_L(r)$ values should be $\leq 3\%$. The most complete and smooth MC dataset was selected that also considered electronic disequilibrium and the dose from electron emissions.

IV.C. 2D anisotropy function

For each source, published $F(r,\theta)$ values from MC and EXP results were graphically compared. If a geometry function different than the simple linear geometry function was used, $F(r,\theta)$ was recomputed. Based on trends observed during the compilation of this report, the agreement between MC and EXP $F(r,\theta)$ values, when available, should be $\leq 5\%$.

V. RECOMMENDATIONS ON DOSIMETRY CHARACTERIZATION METHODS FOR HIGH-ENERGY PHOTON-EMITTING BRACHYTHERAPY SOURCES

The TG-43U1 report[2] on low-energy brachytherapy contains many methodological recommendations and suggestions that should be followed by investigators who would like their published work, whether based upon experimental or computational methods, to be considered as a reference-quality dataset for inclusion in the consensus dose-distribution formation process. In general, all TG-43U1 guidelines and recommendations are also applicable to high-energy source dosimetry, unless otherwise specified in the sections below. Thus, the present recommendations emphasize mainly variances from the TG-43U1 LEBD recommended methodology for obtaining brachytherapy dosimetry parameters.

In 2007, AAPM/ESTRO recommendations on dosimetric prerequisites for routine clinical use of photon-emitting brachytherapy sources with average energies higher than 50 keV were published.[17] These recommendations similar to the AAPM LEBD recommendations[16] apply to brachytherapy sources that are intended for routine clinical use and were intended to define minimum requirements for future

source dosimetry studies so that the accuracy and consistency of the consensus datasets may be improved.

In the current report, only the deviations from the TG-43U1 recommendations[2] (Sec. V, p. 650) necessitated by the higher photon energies or different physical configurations of the sources are noted. These are categorized as (A) preparation of dosimetry parameters, (B) reference data and conditions for brachytherapy dosimetry, (C) and (D) methodological recommendations, (E) uncertainty analyses, (F) publication of dosimetry results, and (G) non-MC computational methods.

V.A. Preparation of dosimetry parameters

For the high-energy sources, e.g., HDR ^{192}Ir sources, dosimetric parameters should be tabulated for 2D formalisms. Exceptions include spherical pellets (e.g., ^{137}Cs Selectron from Nucletron) where a 2D model cannot be formulated. Regardless of the dimensionality of the formalism adopted, the line-source approximation should always be used for computing the geometry function (with the obvious exception of spherically symmetric sources), $G_L(r,\theta)$, and $g_L(r)$. For reader convenience, we include the following alternative expression for $G_L(r,\theta)$:

$$G_L(r,\theta)=\frac{\cos^{-1}\left(\frac{r\cos\theta-\frac{L}{2}}{\sqrt{r^2+\left(\frac{L}{2}\right)^2-Lr\cos\theta}}\right)-\cos^{-1}\left(\frac{r\cos\theta+\frac{L}{2}}{\sqrt{r^2+\left(\frac{L}{2}\right)^2+Lr\cos\theta}}\right)}{Lr\sin\theta}. \quad (6)$$

This expression of $G_L(r,\theta)$ has been included in which cos and $\cos^{-1}$ are used as alternatives to tan and $\tan^{-1}$. The practical reason is that a negative argument of $\tan^{-1}$ results in a negative angle, instead of an angle between 90° and 180° as required by the TG-43 formalism polar coordinate system.

For $F(r,\theta)$ and $g_L(r)$, the minimum–maximum range for r and θ, and the resolution within this range where dose rate shall be calculated or measured, has been discussed in Sec. IV. If polynomial fits are presented, care should be taken to assure agreement within 0.5% between the polynomial fit prediction and the original tabulated data over the whole range. Special care must be taken rounding-off parameters from the fit. To assure that $g(r_0)=1$ with enough precision, the sumation of all the parameters must be "1.0000." Further, the range over which the fit is applicable should be stated. In addition to the TG-43 dosimetry parameters, a derived away-along table should be included for TPS QA testing purposes as described in Sec. IV.

V.A.1. Air-kerma strength

As similarly recommended in the TG-43U1 report,[2] source strength for high-energy sources should be expressed in terms of air-kerma strength or RAKR, not apparent activity, mg-Ra-eq, or other antiquated units. Exceptions may result in patient harm.

V.A.2. Dose rate constant

All TG-43U1 recommendations are applicable to high-energy sources, with the exception that for conventionally encapsulated, ^{192}Ir, ^{137}Cs, and ^{60}Co sources, only a single source is required for experimental purposes. To ensure validity of the source model used by MC simulations, pinhole autoradiography,[94] multislit techniques[95] and transmission radiography should be utilized to confirm the manufacturer's specifications for active length, uniform activity distribution, and physical-to-active source-tip offset. Experimental determinations of absolute dose rates to water from high-energy sources should have direct traceability of S_K to a primary or secondary standard dosimetry laboratory such as the National Institute of Standards and Technology (NIST) or an Accredited Dosimetry Calibration Laboratory (ADCL). Experimentally, Λ is evaluated by taking the ratio $\dot{D}(r_0,\theta_0)/S_K$.

V.A.3. Radial dose function

In addition to the TG-43U1 recommendations, investigators must consider using coupled photon–electron MC codes for short distances where secondary charged particle equilibrium failures imply a deviation of dose from collisional kerma in excess of 2%. As discussed in Sec. III.B, deviations greater than 1% may occur at distances less than 7, 3.5, 2, and 1 mm from the center of ^{60}Co, ^{137}Cs, ^{192}Ir, and ^{169}Yb sources, respectively. Similarly, β-ray transport must be simulated at distances where dose-to-kerma ratio deviations exceeding 1% are possible.

V.A.4. 2D Anisotropy function

The recommendations of the AAPM TG-43U1 report are to be followed.

V.B. Reference data and conditions for brachytherapy dosimetry

V.B.1. Radionuclide data

The influence of photon spectrum choice on brachytherapy dosimetry parameters such as Λ and $g(r)$ has been studied by Rivard *et al.*[96] For ^{192}Ir sources, they found that the uncertainties propagated to these parameters by photon-spectrum uncertainties were much less than 1% ($k=1$). Given the standardization of radionuclide data available

from the NNDC and the rigorous infrastructure for performing and maintaining the dataset evaluations, the AAPM and ESTRO recommend that NNDC data be used for clinically related applications of *all* brachytherapy sources.[20]

V.B.2. Reference media

As recommended by TG-43U1, pure degassed liquid water (H_2O) with a mass density of 0.998 g/cm^3 at 22.0 °C should be used for MC as the medium for both specification of absorbed dose and dose distributions. As clarified in the TG-43U1S1 report,[3] dry air (0% humidity) is recommended for S_K in contrast to the TG-43U1 report which recommended air at 40% relative humidity. The composition of dry air is given in Table XIV of the TG-43U1 report.

V.C. Methodological recommendations for experimental dosimetry

Historical reviews of experimental dosimetry for interstitial brachytherapy sources, including high-energy sources, appear in Williamson[97] and, for ^{192}Ir only, in the original TG-43 report.[1] Starting from the earliest work of Meredith *et al.*,[98] who used a cylindrical perspex ion chamber to measure exposure in air and water for ^{192}Ir interstitial sources, and progressing to dose measurements using LiF TLDs in solid phantoms, these papers and their associated references give an excellent perspective on experimental dosimetry methodologies in this field. A more detailed and contemporary review of experimental brachytherapy dosimetry methods, including emerging detector technologies such as radiochromic film, gels, and liquid-filled ionization chambers, has been given by Williamson and Rivard.[99]

V.C.1. Detector choice

Experimental determination of dose distributions around high-energy brachytherapy sources face the same challenges as their low-energy counterparts: high dose gradients near the source and low-dose rates further away. Moreover, at close distances to the brachytherapy source, detector size can significantly influence dose measurement accuracy due to averaging in the presence of high dose gradients and source self-attenuation. Thus, a suitable detector should possess a wide dynamic range, high sensitivity, flat energy response, and small geometric dimensions. A number of detectors (e.g., diodes, radiochromic films, and TLDs) satisfy the above criteria and therefore have commonly been used. Dosimeters used for reference data should satisfy the following criteria:

1. A relatively small active volume such that effects resulting from averaging of high-gradient dose fields over this volume are negligible or are accurately accounted for by correction factors.
2. A well-characterized energy-response function such that differences between the calibration energy and experimentally measured energy are either negligible or may be quantitatively accounted for.
3. Sufficient precision and reproducibility to permit estimation of dose rate in medium with $k = 1$ Type A (statistical) uncertainties $\leq$3% and $k = 1$ Type B uncertainties $\leq$6%.

While no practical detector system perfectly fulfils the three requirements above, among the established dosimetry techniques, LiF TLD-100 detectors provide a good tradeoff between flat energy dependence, small size, and detector dynamic range for both high- and low-energy brachytherapy sources and thus has been used most frequently.[99,100] For example, silicon diodes, which have smaller active detector volumes and larger sensitivities (reading per unit dose in water), violate requirement 2 above. They have sensitivities that vary by as much as 60% with respect to source-detector distance[101,102] for ^{169}Yb and ^{192}Ir sources due to variations in photon spectra. Thus, the AAPM and ESTRO currently do not recommend silicon diode detectors for reference-quality dose measurement for sources with mean energies exceeding 50 keV. Among validated and fully developed dosimeter technologies, TLD dosimetry has the least position-dependent sensitivity for high-energy sources. TLD energy response has been reported to vary 10%–15% over the 1 to 10 cm distance range for ^{192}Ir sources.[103] Similar magnitude but opposite direction variations have been reported for older (MD-55-2 and earlier) radiochromic film models.[104,105] Newer models of radiochromic film [EBT (Refs. 106–108) and EBT2 (Refs. 109–112)] include small concentrations of a medium atomic number loading compound designed to compensate for the absorbed dose under-response of the diacetylene monomer active sensor medium. EBT film type has a nearly energy-independent dose response.[113,114] MD-55-2 radiochromic film has been used successfully to measure high resolution (<0.25 mm) absolute dose distributions around HDR ^{192}Ir sources[115] and LDR ^{137}Cs sources[116] with $k = 1$ total uncertainties of 4%–4.6%, among the lowest ever reported for such measurements around a brachytherapy source using a secondary detector. However, these detectors must be considered under development at this time because of numerous artifacts (nonuniformity, dose rate dependence, film darkening kinetics, scanner artifacts) which require rigorous correction. TLD dosimetry techniques for both general radiotherapy applications[100,117] and reference-quality brachytherapy dosimetry have been reviewed extensively.[99,100]

V.C.2. Phantom material and energy response characterization

For low-energy brachytherapy dosimetry, accurate knowledge of the atomic composition of the phantom is critical for proper results.[100] The TG-43U1 report allows use of either single-component high-purity industrial plastics or polyamine-based epoxy resin mixtures (e.g., commercial solid water), which can have somewhat variable atomic compositions in their makeup. Therefore, it is suggested that the composition be independently determined by elemental composition assays of representative samples. In all cases, phantom-to-liquid-water corrections (based upon MC calculations) must be applied to the measurements.

For ^{192}Ir and other high-energy sources, absorbed-dose water equivalence is less dependent on phantom

composition,[103] so that commercial plastics such as polymethylmethacrylate (PMMA) as well as single-component resin mixtures can be used with lower correction uncertainties due to knowledge of their composition. Experimentally, Meli *et al.*[103] found that PMMA, polystyrene, and Solid Water™ introduced corrections ranging from −4% to +2% relative to liquid water at distances of 3–6 cm. MC calculations, simulating monoenergetic point sources embedded in 1 m radius phantoms composed of liquid or Solid Water, demonstrated that the latter introduced corrections of less than 5% at 10 cm distance for photon energies greater than 100 keV. A more recent MC study[118] of ^{192}Ir phantom correction factors for cylindrical phantoms of PMMA, polystyrene, and Solid Water™ found that corrections depended on phantom dimensions as well as phantom media. For a phantom size of 20 cm diameter and height (typical for experimental purposes), correction factors were <4% for $r \leq 10$ cm. For larger 40 cm phantoms, larger corrections (up to 6% at 10 cm for PMMA) were noted.

Industrial plastic phantoms (PMMA, polystyrene, or polycarbonate) for high-energy brachytherapy dosimetry are recommended. Single-component resin phantoms are recommended and should be accompanied by the appropriate phantom to water correction factors or should include an estimate of the uncertainties associated with the nonwater equivalence of the phantom for sources with average photon energies greater than 0.2 MeV, but should be avoided for average energies between 0.05 and 0.2 MeV unless validated by elemental composition assays. While atomic composition measurements are mostly unnecessary in this energy range, density measurements should be performed. MC-based medium corrections (phantom-to-liquid water conversion factors based upon the assumed composition and actual geometry and density of the experimental phantom) should be used. However, the dosimetry investigator should also consider the dependence of detector response as a function of source distance within the phantom due to differences in response between the phantom and reference medium, i.e., liquid water.

The TG-43U1 report recommended the experimental dosimetry formalism introduced by Williamson and Meigooni,[119] which has been updated[100] and whose notation is used below. An important correction factor is the relative energy response correction, $E(Q_0, \mathbf{G}_0 \rightarrow Q_{\text{ref}}, \mathbf{G}_{\text{ref}}, \mathbf{r}; \mathbf{G}_{\text{exp}})$, which accounts for the difference in detector sensitivity between the megavoltage photon beam used to calibrate the detector and the brachytherapy source irradiation geometry. $\mathbf{G}_0, \mathbf{G}_{\text{ref}}$, and $\mathbf{G}_{\text{exp}}$ are vectors corresponding to the energy-response correction factors associated with measurements taken during the calibration setup, the reference geometry ($\mathbf{G}_{\text{ref}}$ = unbounded water phantom with point detectors), and the source irradiation setup [e.g., $G_{\text{exp}} = (25\,\text{cm})^3$ PMMA phantom, $(1 \times 1 \times 1)$ mm^3 TLD-100 detectors], respectively. Q_0, Q_{exp}, and Q_{ref} denote the corresponding spectra in these irradiation geometries. The relative energy response correction can be factored into three separate corrections[100]

$$E(Q_0, \mathbf{G}_0 \rightarrow Q_{\text{ref}}, \mathbf{G}_{\text{ref}}, \mathbf{r}; \mathbf{G}_{\text{exp}}) = \frac{k_{\text{bq}}^{\text{rel}}(Q_0 \rightarrow Q_{\text{exp}}; M_0) \cdot f^{\text{rel}}(Q_0, \mathbf{G}_0 \rightarrow Q_{\text{exp}}, \mathbf{G}_{\text{exp}}, \mathbf{r})}{p_{\text{phant,wat}}(Q_{\text{exp}}, \mathbf{G}_{\text{exp}} \rightarrow Q_{\text{ref}}, \mathbf{G}_{\text{ref}}; \mathbf{r})}, \tag{7}$$

where the intrinsic relative energy response correction is given by

$$k_{\text{bq}}^{\text{rel}}(Q_0 \rightarrow Q_{\text{exp}}; M) \equiv \frac{k_{\text{bq}}(M, Q_0)}{k_{\text{bq}}(M, Q_{\text{exp}})} = \frac{(M_0/\bar{D}_{\text{det}})(\mathbf{r}, Q_{\text{exp}}, \mathbf{G}_{\text{exp}})}{(M_0/\bar{D}_{\text{det}})(Q_0, \mathbf{G}_0)}, \tag{8}$$

where M is the detector reading and $\bar{D}_{\text{det}}$ is the mean absorbed dose to the active detector volume. $k_{\text{bq}}^{\text{rel}}(Q_0 \rightarrow Q_{\text{exp}}; M)$ describes the efficiency with which the detector-response mechanism transforms energy imparted to its active collection volume by the brachytherapy radiation field into an observable response, relative to its efficiency in the calibration beam. The relative absorbed-dose energy dependence is given by

$$f^{\text{rel}}(Q_0, \mathbf{G}_0 \rightarrow Q_{\text{exp}}, \mathbf{G}_{\text{exp}}, \mathbf{r}) \equiv \frac{f(\mathbf{r}, Q_0, \mathbf{G}_0)}{f(\mathbf{r}, Q_{\text{exp}}, \mathbf{G}_{\text{exp}})} = \frac{(\bar{D}_{\text{det}}/D_{\text{wat}})(\mathbf{r}, Q_{\text{exp}}, \mathbf{G}_{\text{exp}})}{(\bar{D}_{\text{det}}/D_{\text{med}_0})(Q_0, \mathbf{G}_0)}, \tag{9}$$

f^{rel} is that component of relative detector response which is due only to the efficiency with which the brachytherapy spectrum imparts energy to the active detector volume relative to the corresponding efficiency in the calibration beam, when both efficiencies are normalized to dose in medium in the absence of the detector. f^{rel} includes the displacement and volume-averaging corrections. The dose-measurement phantom correction factor, $p_{\text{phant,wat}}(Q_{\text{exp}}, \mathbf{G}_{\text{exp}} \rightarrow Q_{\text{ref}}, \mathbf{G}_{\text{ref}}; \mathbf{r})$, corrects for differences between the irradiation geometry used to perform the measurements and the reference geometry in which the final dose distribution is to be specified. As a ratio of geometric point doses in homogeneous media, it is independent of the detector geometry, composition, and underlying mechanism and depends only on the reference and experimental phantom dimensions, composition, and positioning relative to other sources of scattered radiation near the measurement phantom.

The controversies surrounding the choice of $k_{\text{bq}}^{\text{rel}}(Q_0 \rightarrow Q_{\text{exp}}; M_0)$ corrections for TLD dosimetry of low-energy brachytherapy sources, where recent experiments suggest energy response correction factors ranging from 1.05 to 1.10, have been reviewed by Williamson and Rivard.[100] While $k_{\text{bq}}^{\text{rel}}$ values are closer to unity for high-energy brachytherapy sources, two recent publications found anomalously high values of $k_{\text{bq}}^{\text{rel}} = 1.018$–$1.038$ for ^{137}Cs relative to ^{60}Co.[120,121] Overall energy-response corrections for HDR ^{192}Ir brachytherapy sources have been measured but without result comparisons to MC calculated absorbed-dose energy-dependent factors. Because definitive factors are not yet available, it is recommended that $k_{\text{bq}}^{\text{rel}}$ be taken as unity for high-energy photon dosimetry, while $p_{\text{phant,wat}}$ and f^{rel} should be carefully calculated for the experimental geometry

used. Without additional information, a $k=1$ uncertainty of 3% may be assigned to the overall energy response correction factor.

For radiochromic film, it is not clear if dosimetrically significant energy response corrections exist or not. Based on a study of Model MD-55-2, Bohm *et al.*[104] concluded that MC f^{rel} accounted for measured $E(Q_0 \rightarrow Q_{\mathrm{ref}})$ values within 5%. On the other hand, Sutherland *et al.*[122] found relatively poor agreement between their MC calculations and previously reported measurements.[123,124] Since radiochromic film response is highly dependent upon film composition and depends on a host of other factors, including temporal history and temperature,[125,126] it is recommended that this detector be used cautiously.

V.C.3. Specification of measurement methods

Recommended methodologies for using TLD dosimetry in brachytherapy have been reviewed elsewhere.[99,100] All recommendations in Sec. V.D 3 of the 2004 AAPM TG-43U1 report[2] should be followed for high-energy brachytherapy dosimetry. Careful correction for volume averaging, source and/or detector displacements, and phantom composition/size should be applied so that the final dose rates represent absorbed dose rates to water per unit S_K at geometric points in an unbounded liquid water medium. The location of dose measurement points should be referenced to the geometric center of the active source core.

V.D. Methodological recommendations for Monte Carlo based dosimetry

Codes that have been widely used for high-energy source dosimetry include PTRAN, MCNP, GEANT4, PENELOPE, and EGSNRC. At the time of publication of this report, all these codes are based upon modern cross-section libraries and complex and accurate physics models to simulate transport of electrons and photons through complex media. All these codes have been benchmarked against experimental measurements or by code intercomparisons. For high-energy sources, collisional kerma approximates dose at distances from the source surface where electronic equilibrium is reached. However, electronic equilibrium at close distances from ^{192}Ir, ^{137}Cs, and ^{60}Co sources is not reached, and beta and internal conversion electrons emerging from the source capsule require detailed electron transport if accurate dose rate estimates near the sources are required (Sec. III.B). Errors exceeding 2% will occur if photon-only MC transport simulation is used to estimate dose for distances at or below 1.6, 3, and 7 mm for ^{192}Ir, ^{137}Cs, and ^{60}Co sources, respectively.[28]

In general, the AAPM and ESTRO recommend that MC investigators utilize well-benchmarked codes for brachytherapy dosimetry studies intended to produce reference-quality dose rate distributions for clinical use. A benchmarked code is able to reproduce MC simulations comparable to those obtained by other codes validated experimentally or a code whose results have been validated experimentally. However, all investigators should assure themselves that they are able to reproduce previously published dose distributions for at least one widely used brachytherapy source model. The 2007 HEBD prerequisites[17] stated that MC transport codes should be able to support dose rate estimation with expanded uncertainties ($k=2$) no greater than the 3%–5% characteristic of the MC transport codes currently used for low-energy source dosimetry. Also, the 2007 report included methods to benchmark the MC calculation method. Agreement between the MC results and the benchmark data should be within 2% for Λ, 5% for $g_L(r)$, and 10% for $F(r,\theta)$ within 5° from the source long axis.[17] Unlike for low-energy sources, the range of secondary electrons from high-energy sources will require electron transport at short distances.[28]

V.D.1. Specification of Monte Carlo calculation methods

The nine points in the list in Sec. V.E 1 of the TG-43U1 report are applicable to high-energy sources with the following changes:

1. Limit consideration to emitted photon energies above 10 keV (for simulations in both water and in-air or *in vacuo*). Based on typical PDR/HDR source encapsulations, 10 keV should be an adequate cut-off and is commonly used in publications. A lower energy cutoff does not produce more accurate results for most dosimetry applications but prolongs the calculation time required to achieve a fixed Type A uncertainty level (or prevents finer spatial resolution with associated volume averaging).
2. All photons emitted with an energy above the 10 keV cutoff must be included in dosimetry calculations. At least one publication has reported that high-energy photons with low emission probabilities can influence results significantly.[96] Therefore, reference spectra must be used in their entirety in MC simulations, i.e., NNDC reference spectra[20] must not have low intensity lines removed.
3. If charged particle transport is simulated, the underlying transport algorithm should be described clearly, if only by reference. The quantity used to approximate dose (e.g., collisional kerma) or any variance reduction techniques should be clearly specified. Whether beta-ray and internal conversion electron transport is included, along with the initial beta spectrum used, should be specified.

V.D.2. Good practice for Monte Carlo calculations

1. Reference-quality absorbed dose rate to water distributions should be computed in liquid water in a phantom which approximates full scatter conditions characteristic of an unbounded phantom. For ^{192}Ir, ^{137}Cs, and ^{169}Yb sources, a spherical phantom with radius $R=40$ cm (or the equivalent cylindrical phantom dimensions) should be used, while $R=80$ cm is required for ^{60}Co (Refs. 29–31) sources.
2. A suficient number of histories should be calculated to ensure that the dose rate per simulated history $\dot{d}(r,\theta_0)$ and $\dot{k}_{\mathrm{air}}(d,\theta_0)$ calculations for derivation of s_K have Type A uncertainties $(k=1)<0.1\%$ for distances ≤ 5 cm and Type A uncertainties $(k=1)<0.2\%$ for distances $\leq$

10 cm. In evaluating s_K, the confounding influence of contaminant low-energy photons below 10 keV (and contaminant electrons as well if charged-particle transport is simulated) should be assessed and corrected for if necessary. By convention, $\dot{k}_{\text{air}}(d,\theta_0)$ and s_K must be specified in dry air.

3. The influence of photon cross section uncertainties on dose estimation accuracy has not been comprehensively studied in the high-energy brachytherapy regime. Until careful studies demonstrate otherwise, TG-43U1 recommendations should be followed. This includes use of post-1980 cross-section libraries, preferably those equivalent to the current NIST XCOM database such as DLC-146 or EPDL97. Older cross-section libraries based on Storm and Israel data[127,128] must be avoided. Electron binding effects on coherent and incoherent scattering should be simulated using the form factor approximation. In the presence of high atomic number absorbers, atomic relaxation processes resulting in characteristic x-rays exceeding 10 keV should be simulated. Mass–energy absorption coefficients used to convert energy fluence into collisional kerma must be consistent with the interaction physics models and photon cross sections used for transport.
4. Collisional kerma and dose estimators (scoring tally)[129] and detector volumes should be chosen to limit volume-averaging artifacts to <0.1%. To minimize the impact of voxel size effects[130–132] while maintaining reasonable efficiency for track-length and analog estimators, maximum voxel sizes in cartesian coordinates could be chosen in the following way: $(0.1\,\text{mm})^3$ voxels for distances in the range of $r_{\text{source}} < r \leq 1\,\text{cm}$, $(0.5 \times 0.5 \times 0.5)\ \text{mm}^3$ voxels for $1\,\text{cm} < r \leq 5\,\text{cm}$, $(1 \times 1 \times 1)\ \text{mm}^3$ voxels for $5\,\text{cm} < r \leq 10\,\text{cm}$, and $(2 \times 2 \times 2)\ \text{mm}^3$ voxels for $10\,\text{cm} < r \leq 20\,\text{cm}$, where r is defined as the distance from the center of the source. Rectilinear or toroidal voxels of similar radial dimensions should have similar volume-averaging effects.
5. Especially for photon sources in the 50 to 300 keV energy range, the manufacturer-reported dimensions of encapsulation and internal components should be verified through the use of physical measurements, transmission radiography, and autoradiography. For all sources, transmission radiography and pinhole radiography should be used to verify the active source dimensions and location relative to the physical source dimensions and that the radioactivity is approximately uniformly distributed. The impact of internal source component mobility[133] on the dose distribution should be assessed.
6. Some MC studies consider the effect of electronic nonequilibrium conditions near a brachytherapy source, or the beta-ray contribution to the dose distribution near the source. In these cases, secondary electron transport should be simulated. To avoid inconsistencies and systematic errors in the results, the following precautions should be heeded. Because brachytherapy simulations involve rather extreme conditions (very small detector thicknesses, low energies, etc.) that may invalidate the approximations upon which the charged particle transport algorithms are based, they may produce artifacts that are evident only in extreme cases but that are masked in other situations. The following precautions cover different aspects including physics models implemented in the codes and electron tracking techniques, among others:

(a) Usually, the simplest strategy is to perform test simulations starting with standard simulation parameters recommended for the code under consideration, followed by other test runs that vary these parameters to study their influence on the final results.
(b) Electron step size is a critical parameter that influences deposited doses in small geometry regions. It should be handled with care in each simulation and, if adjustable, parametric studies should be performed to demonstrate that the dosimetric results are not sensitive to this parameter choice.
(c) Some multiple scattering (MS) theories place limits on the minimum number of mean collisions that must occur in each condensed history step for validity to be maintained. The existence of steep dose gradients at the distances of interest necessitates high spatial resolution for dose computation. Consequently, shells to score dose are very thin close to the source. The Molière MS minimum step size imposes a restriction on the spatial resolution of MC simulation. Care must be taken to maintain the dimension of the scoring region above this limit.[134] This limitation affects mainly codes derived from EGS4.
(d) The user must be sure that the number of interactions in a voxel is large enough (a minimum of 10) for the result to be statistically well behaved.
(e) Some codes handle boundary crossing algorithm corrections poorly while others generate artifact-free corrections. Switching to single-scattering mode near boundaries is the preferred solution. For example, Type-1 transport algorithms (MCNP, ITS, ETRAN), which use Goudsmit–Saunderson multiple-scattering formalism parameters, stopping powers, and energy-straggling corrections precalculated on a fixed logarithmically spaced energy-loss grid, are particularly subject to boundary crossing algorithm artifacts as media and detector interfaces truncate condensed history steps at arbitrary intermediate values. The influence of such partial steps cannot be recovered by interpolation of precalculated data. Chibani and Li[135] demonstrate that pre-2000 versions of MCNP-determined low-energy electron dose distributions were sensitive to choice of energy-indexing (boundary crossing algorithm interpolation scheme).
(f) Variance reduction techniques are often implemented in the codes, and although they are generally robust, they should be used with care. In particular, the user is advised to check that results are unbiased.

V.E. Uncertainty analyses

Both experimental and MC determinations of reference-quality single-source dose rate distributions should include formal uncertainty analyses that adhere to the methodology of NIST Technical Note 1297.[136] While a number of publications[100,137] including the TG-43U1 and TG-138 (Ref. 14) reports give detailed guidance on applying this methodology to low-energy brachytherapy, complete and rigorous uncertainty analyses for high-energy brachytherapy are generally lacking. However, extensive uncertainty analyses are given by Raffi *et al.*[41] for HDR ^{192}Ir experimental and MC and Granero *et al.*[138] for HDR ^{192}Ir MC simulations. These papers include both Type A and Type B uncertainties. These uncertainties are in agreement with those in the AAPM TG-138 report and are over a factor of two lower than those in Table XII of the TG-43U1 report for low- and high-energy sources. While ^{169}Yb has been considered by some manufacturers, the S_K calibration uncertainties are still a matter of study and are of the order of 3% ($k = 1$). As similarly recommended in Sec. V.D 3(10) and Sec. V.E 1(9) of the AAPM TG-43U1 report for measurements and simulations of low-energy photon-emitting brachytherapy dosimetry studies, respectively, the AAPM recommends that high-energy brachytherapy source dosimetry investigators perform detailed uncertainty analyses in a manner similar to Raffi *et al.* and Granero *et al.* yet specific to the source model and conditions examined in their investigation.

V.F. Publication of dosimetry results

As recommended by TG-43U1 (Ref. 2) and the HEBD prerequisites,[17] commercially distributed high-energy sources used in routine clinical practice should be supported by two independent dosimetry studies that adhere to the methodological recommendations of this report. As defined by TG-43U1, "independence" requires (a) that dosimetry investigators be free of affiliations or other conflicts of interest with the source vendor and (b) the two studies be scientifically independent of one another. The Li *et al.* recommendations[17] require that one study be experimental (usually TLD-based) and that the other be theoretical (MC). The studies must be published in the peer-reviewed literature. A technical note format is acceptable as is publishing the two independent studies in the same publication. Given publication length limitations, AAPM committees do not require that all expected or needed documentation and method description be included in the published paper. However, it must be either posted electronically with the online version of the paper or made available by the authors via a personal communication upon request. Conventionally encapsulated ^{192}Ir, ^{137}Cs, and ^{60}Co sources require only a single MC-based study for comprehensive dose characterization.

Some TPS algorithms correct the dose from full scatter to the clinical specific conditions and require dosimetry parameter data based on full scatter conditions. For some of these TPS algorithms, it has been proposed that the primary- and scatter-component functions be obtained from TG-43-based dose rate tables and will need to be handled independently by the TPS dose calculation algorithm.[139]

V.G. The role of non-Monte Carlo computational tools in reference dosimetry

Over the years, a variety of computational tools, in addition to MC simulation, have been proposed or even widely used for the determination of single-source dose distributions in the high-energy photon regime.

Heuristic analytical model algorithms were not introduced as dosimetry or dose-estimation tools, but as treatment-planning tools for computing more realistic and accurate dose distributions for clinical multisource implants in the presence of tissue-composition and density heterogeneities, applicator shielding and attenuation, and interseed attenuation. Accelerated MC simulation codes[140] have also been adapted for clinical dose computation. The potential for these innovations in clinical dose computation has been reviewed by Rivard *et al.*[33] and is the subject of the active AAPM Task Group 186.

Prior to community-wide acceptance of the 1995 AAPM TG-43 report, nearly every general-purpose brachytherapy planning system utilized the 1D path-length or Sievert model to generate single-source dose distributions around encapsulated line sources such as intracavitary brachytherapy tubes. Comparisons with MC simulation demonstrate that with properly selected input parameters and realistic modeling of the source geometry, accurate results (2% transverse axis and 5% longitudinal axis differences) can be achieved for ^{137}Cs tubes and needles.[57,141] However, for lower energy sources, including LDR ^{192}Ir seed and HDR ^{192}Ir sources, accurate modeling of 2D anisotropy corrections cannot be achieved.[142] Simple extensions of the Sievert model can restore accuracy in many cases such as by separating primary and scatter components and modeling the latter as an isotropic distribution.[142,143] However, comparisons between benchmark calculations from MC or analytical methods such as the Sievert integral are required to ensure dose prediction accuracy for new source designs. Hence, 1D path-length models are not endorsed by this report for estimation of reference-quality dose distributions for any category of high-energy sources.

A number of more sophisticated scatter separation algorithms, which involve 1 -, 2 -, or even 3-dimensional integration of the scatter dose distribution over the implant geometry have been proposed.[73,144–147] Closely related are superposition/convolution algorithms[148] of which the most fully developed is Carlsson–Tedgren's[149,150] brachytherapy adaptation of the external-beam collapsed cone approach. As with the simpler Sievert–style algorithms, these approaches require significant fine tuning and validation against more definitive MC simulations to avoid excessive systematic dose computation errors, and thus are not acceptable as substitutes for MC simulation for estimation of reference-quality single-source dose distributions.

A more empirical scatter-separation method was introduced[151] for CT-based planning for HDR ^{192}Ir

brachytherapy; the primary and scatter dose distributions for each dwell position are calculated first as if the patient is an infinite water phantom. Corrections for photon attenuation, scatter, and spectral variations along medium- or low-Z heterogeneities are made according to the radiological paths determined by ray tracing. The scatter dose is then scaled by a correction factor that depends on the distances between the points of interest, the body contour, and the source position. Dose calculations were evaluated for phantoms with tissue and lead (Pb) inserts, as well as patient plans for head-and-neck, esophagus, and balloon breast brachytherapy treatments. PTRAN_CT-based MC calculations were used as the reference dose distributions. For the breast patient plan, the TG-43 formalism overestimated the target volume receiving the prescribed dose by about 4% and $_{skin}D_{0.1cc}$ by 9%, whereas the analytical and MC results agreed within 0.4%.

Deterministic transport equation solvers, most commonly discrete ordinates methods simulations, have also been investigated for their potential use in brachytherapy planning applications.[152,153] A grid-based Boltzmann solver (GBBS) was introduced as a supported option in a commercially available brachytherapy planning system.[154,155] In contrast to the more sophisticated heuristic algorithms [class 1(b) above], GBBS directly solves the underlying Boltzmann transport equation on a systematically discretized seven-dimensional phase-space mesh. Because GBBS algorithms use random sampling on a very limited basis if at all, GBBS results do not suffer from statistical noise and very slow convergence rates. However, many application-specific parameters need to be optimized including density of the angular mesh, energy group structure and weighting functions, as well as spatial mesh geometry and angular-flux interpolation technique. Inadequate optimization can lead to substantial systematic errors and artifacts, e.g., ray effects. While very promising tools for radiotherapy planning purposes, inherently more accurate MC benchmarks are required for GBBS tuning and validation. Hence, GBBS and related techniques[156] are not suitable reference-quality dosimetry tools.

In summary, of the computational tools developed to date, only MC simulation is an acceptable method for estimating reference-quality dosimetry parameters. This is a consequence of the fundamental mathematical nature of MC simulation, which yields a statistically imprecise, but exact first-principles solution of the transport equation. While statistical noise in some settings can be a limiting problem, in the context of brachytherapy reference dosimetry, it can be eliminated as a practical issue through long run times, efficient sampling techniques, or proper selection of variance-reduction strategies.[100] Although approximations are often used within MC codes, the ideal of convergence to an unbiased solution of the Boltzmann equation is approximated to a high degree of accuracy in practice. Residual errors, e.g., volume averaging, are straightforward to correct or eliminate using modern codes. In contrast, both deterministic heuristic and transport-solution algorithms, while free of statistical uncertainty, are always subject to complex, geometry-dependent patterns of systematic error.

VI. RECOMMENDED DOSIMETRY DATASETS FOR HIGH-ENERGY PHOTON-EMITTING BRACHYTHERAPY SOURCES

Recommended consensus datasets for high-energy sources have been obtained for sources that were commercially available as of January 2010. Data are presented according to the AAPM TG-43U1 formalism, with upgraded interpolation and extrapolation techniques in Table III for $F(r,\theta)$ and $g(r)$. Additionally, the radial and angular ranges of the datasets are chosen to accurately represent the dosimetric characteristics given linear interpolation by TPS. A common mesh was introduced for $g_L(r)$, and the mesh of the selected publication has been kept for $F(r,\theta)$. For each source model, and the selection procedure is explained with additional discussion included (Appendix A).

For TPS that use the TG-43 dose calculation formalism and permit user input of dosimetry parameters, the medical physicist should enter the dosimetry parameters and check the accuracy of the dose calculation.[13] These tasks should be well documented. For some TPS, dosimetry parameters are entered by the manufacturer, without the possibility of user modification. In these cases, users should verify the correct entry and document these commissioning findings before releasing the TPS for clinical use.

Clinical implementation of these datasets should follow the recommendations included in Sec. VI of the TG-43U1 report.[2] A medical physicist should implement the dose calculation data and techniques recommended by this report on the TPS and quantitatively assess the influence of this action on dose delivery. In cases where data are introduced as coefficients in an equation, e.g., a polynomial function for $g_L(r)$, it is necessary to evaluate the quality of the fit over the intended calculation range. Users must verify that the TPS follows the TG-43U1 formalism and should also document the TPS methods for interpolation and extrapolation (applying the recommendations introduced in TG-43U1S1 and also more specifically in this report) of dose calculations within and beyond the range of provided dosimetry parameters. The dose rates calculated by the TPS from a single source should be compared with the dose rate distribution derived from the tabulated consensus values presented in this report. To facilitate this comparison, dose rate tables in a Cartesian coordinate system have been included as has been recommended previously by the AAPM (TG-40,[157] TG-53,[158] TG-56,[13] and TG-43U1.[2]) This comparison should yield agreement within $\pm 2\%$ over all angles and over the range of radial distances commissioned. Discrepancies exceeding 2% should be documented and critically examined since better agreement is expected.

VI.A. AAPM-RPC source registry

In 2001, the RTOG approached the RPC with the request to make available a list of brachytherapy sources that met appropriate criteria and could be considered usable for clinical trials. The RPC collaborated with the AAPM which had issued a report entitled "Dosimetric prerequisites for routine clinical use of new low energy photon interstitial

brachytherapy sources," by Williamson *et al.*[16] Sources that met these dosimetric prerequisites were judged to be sufficiently well characterized, have adequate traceability to national standards, and be manufactured under processes subjected to appropriate quality control standards. Shortly afterward, the joint AAPM/RPC Source Registry was established on the RPC web page and has been maintained ever since. Institutions considering enrolling patients in clinical trials sponsored by the U.S. National Cancer Institute (NCI) that involve low-energy seeds must use sources that are listed on the Registry. The Registry includes tables of dosimetry parameters that have been compiled from peer-reviewed publications and issued as consensus data deemed suitable for clinical use by the AAPM.

Development of a new RTOG protocol requiring use of high-energy photon-emitting brachytherapy sources prompted expansion of the Registry in 2009 to include such sources. For high-energy sources to be included in the Registry, there must be compliance with the HEBD prerequisites.[17] The BTSC and BSR have identified a number of high-energy sources that meet these prerequisites. In response, the RPC has added these sources to the Registry.

The differences in radionuclide characteristics stimulated some changes in the requirements between low- and high-energy photon-emitting brachytherapy sources. Whereas source manufacturers must submit low-energy sources at least annually to NIST or other primary standards labs for S_K calibration consistency, a calibration comparison frequency of 2 yr for ^{60}Co, ^{137}Cs, and ^{192}Ir sources is recommended. Vendors of sources containing these high-energy radionuclides should comply with this comparison frequency and are monitored for compliance by the AAPM and ESTRO. For ^{192}Ir, ^{137}Cs, and ^{60}Co sources of conventional design, the Registry only requires a single published dataset. This must be a MC study of dose to water in water medium as stated in Sec. IV.

A special case exists for *orphaned* sources: those no longer commercially available but still in regular use in hospitals. These must be sources with long half-lives and suitable dose rates that consequently comprise only certain models of ^{137}Cs and ^{60}Co sources. In the case of these sources, there is no manufacturer available to submit the Registry application forms. For these orphaned sources, the AAPM and RPC have developed an approved alternative procedure for Registry application: a hospital that wishes to participate in a clinical trial that involves brachytherapy sources not currently posted on the Registry may submit the application, listing the dosimetric studies available and the dosimetry parameters to be used for treatment planning. The hospital must also describe their method of source strength traceability for review by the RPC to assure the correct calibration of the sources. In the special case of source trains, in which individual sources cannot be removed for calibration with a well chamber, the hospital may describe a method of calibration at a distance in a phantom, in accordance with calibration procedures described in the peer-reviewed literature.

As extensively described by Rivard *et al.*,[159] while posting of a source model on the Registry does not imply existence of an AAPM-endorsed consensus dataset, clinical use of Registry-posted data represents a reasonable choice for medical physicists, the source vendor, and clinical trial investigators for implementing newly marketed seed products. AAPM consensus datasets are typically issued within 3 yr after posting on the Registry and then included on the RPC website.

In the absence of AAPM-issued consensus datasets, ESTRO manages a database for brachytherapy dosimetry parameters and other related data.[160] For low-energy LDR brachytherapy sources for which AAPM-endorsed consensus datasets are available, ESTRO recommends adopting these datasets and the ESTRO website includes a link to the Registry website. A similar policy is implemented for high-energy sources once consensus data are published.

Another online venue for brachytherapy dosimetry parameter data is the Carleton University website.[161] Data for this website includes results of MC simulations for ^{125}I, ^{103}Pd, ^{192}Ir, and ^{169}Yb sources. A key difference between this site and the other three venues is that the data were derived from a common MC radiation transport code, BrachyDose.[132] In addition to the TG-43 dosimetry parameters, dose rate tables for high-energy sources are also presented separately for primary, single-scattered, and multiple-scattered photons. For ^{192}Ir sources, these datasets have been evaluated in this report.

VI.B. Consensus datasets

Sources meeting the 2007 AAPM prerequisites[17] are considered in this section. The publications pertaining to each source have been evaluated following the guidelines described in Sec. IV. Details about source characteristics including source schematic diagram, criteria for selecting consensus data among those published, and a brief discussion about the publications related to each source are available in Appendix A of the full report available online on the AAPM website. In the following section, a brief summary for each source is presented.

VI.B.1. HDR ^{192}Ir sources

The HDR ^{192}Ir brachytherapy sources for which consensus datasets have been obtained are as follows:

(a) Nucletron model mHDR-v1 (classic) source
(b) Nucletron model mHDR-v2 source
(c) Varian Medical Systems model VS2000 source
(d) Eckert & Ziegler BEBIG GmbH model Buchler source
(e) Varian Medical Systems model GammaMed HDR 12i source
(f) Varian Medical Systems GammaMed HDR Plus source
(g) Eckert & Ziegler BEBIG GmbH model GI192M11 source
(h) Eckert & Ziegler BEBIG GmbH model Ir2.A85-2 source
(i) SPEC, Inc. model M-19 source
(j) Isodose Control model Flexisource

VI.B.2. PDR ^{192}Ir sources

The PDR ^{192}Ir brachytherapy sources for which consensus datasets have been obtained are as folows:

(a) Varian Medical Systems GammaMed PDR 12i source
(b) Varian Medical Systems GammaMed PDR Plus source
(c) Nucletron model mPDR-v1 source
(d) Eckert & Ziegler BEBIG GmbH model Ir2.A85-1 source

VI.B.3. LDR ^{192}Ir sources

The LDR ^{192}Ir brachytherapy sources for which consensus datasets have been obtained are as follows:

(a) Best Industries model 81-01 seed
(b) Eckert & Ziegler BEBIG GmbH 0.5 and 1.0 cm long wires

VI.B.4. LDR ^{137}Cs sources

The LDR ^{137}Cs brachytherapy sources for which consensus datasets have been obtained are as follows:

(a) Eckert & Ziegler BEBIG GmbH model CSM-3 source
(b) Isotope Product Laboratories model IPL source
(c) Eckert & Ziegler BEBIG GmbH model CSM11 source

VI.B.5. HDR ^{60}Co sources

The HDR ^{60}Co brachytherapy sources for which consensus datasets have been obtained are as follows:

(a) Eckert & Ziegler BEBIG GmbH model GK60M21 source
(b) Eckert & Ziegler BEBIG GmbH model Co0.a86 source

VI.C. Reference overview of sources without consensus datasets

In addition to the sources enumerated in Sec. VI.B for which consensus data have been produced, there are other sources that have been used in the past in clinical practice or are even still being used at the time of publication of this report. However, these sources were no longer commercially available as of January 2010, and consensus datasets are not issued. However, since there may be retrospective dosimetry trials involving these sources, and also to guide medical physicists still using them clinically, references are provided from which dosimetry data can be obtained (these are justified in Appendix B of the online report). Any manipulation of these datasets is the responsibility of the individual user or company.

These sources are as follows:

(a) LDR ^{137}Cs: pellet, CSM2, CSM3-a, CDCS-J, 6500/6D6C, Gold-matrix series 67-800, CSM1, CDCS-M, CDC.K1-K3, CDC.K4, CDC 12015 to CDC 12035, and CDC.G and CDC.H
LDR ^{192}Ir: Platinum-clad seed
(b) HDR ^{192}Ir: Varian classic
(c) PDR ^{192}Ir: Nucletron
(d) HDR ^{60}Co: Ralstron Type-1, Type-2, and Type-3

NOMENCLATURE

1D	One-dimensional
2D	Two-dimensional
AAPM	American Association of Physicists in Medicine
ADCL	Accredited Dosimetry Calibration Laboratory
BRAPHYQS	ESTRO Brachytherapy Physics Quality assurance System
BSR	AAPM Brachytherapy Source Registry Working Group
BTSC	AAPM Brachytherapy Subcommittee
CTV	Clinical target volume
EC	Electron capture
ESTRO	European Society for Radiotherapy and Oncology
EXP	Experimental measurement
GBBS	Grid-based Boltzmann solver
HDR	High-dose rate
HEBD	AAPM High Energy Brachytherapy Source Dosimetry Working Group
IC	Internal conversion
ISA	Inter-source attenuation
LDR	Low-dose rate
LEBD	AAPM Low Energy Brachytherapy Source Dosimetry Working Group
MC	Monte Carlo
MRI	Magnetic resonance imaging
NIST	U.S. National Institute of Standards and Technology
NNDC	National Nuclear Data Center
PDR	Pulsed-dose rate
POI	Points-of-interest
RPC	Radiological Physics Center
RTOG	U.S. Radiation Therapy Oncology Group
TG-43	AAPM Task Group No. 43 brachytherapy dose calculation formalism
TG-43U1	2004 update to the TG-43 report
TG-43U1S1	2007 supplement to the 2004 AAPM TG-43U1 report
TLD	Thermoluminescent dosimeter generally composed of LiF (TLD-100)
TLS	Two length segmented method
TPS	Treatment planning system(s)
U	The unit of air-kerma strength equivalent to μGy m^2 h^{-1} or cGy·cm^2·h^{-1}.
β	Angle subtended by $P(r,\theta)$ and the two ends of the brachytherapy source active length; as used in the line-source approximation, β has units of radians
d	Distance to the point of measurement from the source center in its transverse-plane, typically measured *in-air* or *in-vacuo*; units of cm
$\dot{d}(r_0,\theta_0)$	The dose rate per history estimated using Monte Carlo methods at the reference position
$\dot{D}(r,\theta)$	Dose rate in water at P(r,θ); the dose rate is generally specified with units cGyh^{-1} and the reference dose rate, $\dot{D}(r_0\theta_0)$, is specified at P(r_0,θ_0) with units of cGyh^{-1}
δ	Energy cut-off parameter used for air-kerma rate evaluation, with units of keV

$F(r,\theta)$ 2D anisotropy function describing the ratio of dose rate at radius r and angle θ around the source, relative to the dose rate at $r_0 = 1$ cm and $\theta_0 = 90^\circ$ when removing geometry function effects; dimensionless units

$G_X(r,\theta)$ Geometry function approximating the influence of the radionuclide physical distribution on the dose distribution; $G_X(r,\theta)$ is calculated by the following:

$$G_P(r,\theta) = r^{-2}$$

point-source approximation

$$G_L(r,\theta) = \begin{cases} \dfrac{\beta}{Lr\sin\theta} & \text{if } \theta \neq 0^\circ \\ (r^2 - L^2/4)^{-1} & \text{if } \theta = 0^\circ \end{cases}$$

line-source approximation

with units of cm^{-2}

$g(r)$ Radial dose function describing the dose rate at distance r from the source in the transverse plane relative to the dose rate at $r_0 = 1$ cm; dimensionless units

$g_L(r)$ Radial dose function determined under the assumption that the source can be represented as a line segment; dimensionless units

$g_P(r)$ Radial dose function determined under the assumption that the source can be represented as a point; dimensionless units

${}_{CON}g(r)$ Radial dose function derived from consensus dataset; dimensionless units

$\dot{k}_\delta(d)$ Air-kerma rate *in vacuo*, per history as estimated using Monte Carlo methods, due to photons of energy greater than δ

$\dot{K}_\delta(d)$ Air-kerma rate *in vacuo* on the source transverse plane due to photons of energy greater than δ, with units of $\text{cGy}\cdot\text{h}^{-1}$

Λ Dose rate constant in water, with units of $\mu\text{Gy}\cdot\text{h}^{-1}\cdot\text{U}^{-1}$; Λ is defined as the dose rate at $P(r_0,\theta_0)$ per unit S_K

${}_{CON}\Lambda$ Notation indicating that the reported value of Λ is the consensus value determined by the AAPM from published data, with units of $\text{cGy}\cdot\text{h}^{-1}\cdot\text{U}^{-1}$

${}_{EXP}\Lambda$ Notation indicating that the reported value of Λ was determined by experimental measurement

${}_{MC}\Lambda$ Notation indicating that the reported value of Λ was determined using Monte Carlo calculations

L Active length of the source (length of the radioactive portion of the source) with units of cm

L_{eff} The effective active length of the source; L_{eff} is used for brachytherapy sources containing uniformly spaced multiple radioactive components; $L_{\text{eff}} = \Delta S \times N$, where N represents the number of discrete pellets contained in the source with center-to-center spacing ΔS

$P(r,\theta)$ Point-of-interest, positioned at distance r and angle θ from the geometric center of the radionuclide distribution

$\phi_{\text{an}}(r)$ 1D anisotropy function; at any radial distance r, $\phi_{\text{an}}(r)$ is the ratio of dose rate averaged over 4π steradian integrated solid-angle to the dose rate at the same distance r on the transverse-plane; dimensionless units

r The distance from the source center to $P(r,\theta)$, with units of centimeter

r_0 The reference distance, generally 1 cm

s_K The air-kerma strength per history estimated using Monte Carlo methods

S_K Air-kerma strength: the product of the air-kerma rate $\dot{K}_\delta(d)$ and the square of the distance d to the point of specification from the center of the source in its transverse-plane; S_K is expressed in units of $\mu\text{Gy m}^2\text{ h}^{-1}$, a unit also identified by U

θ The polar angle between the longitudinal-axis of the source and the ray from the active source center to the calculation point, $P(r,\theta)$

θ_0 The reference polar angle, generally 90° or $\pi/2$ radians

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