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Addendum to the AAPM's TG-51 protocol for clinical reference dosimetry of high-energy photon beams

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An addendum to the AAPM's TG-51 protocol for the determination of absorbed dose to water in megavoltage photon beams is presented. This addendum continues the procedure laid out in TG-51 but new k_Q data for photon beams, based on Monte Carlo simulations, are presented and recommendations are given to improve the accuracy and consistency of the protocol's implementation. The components of the uncertainty budget in determining absorbed dose to water at the reference point are introduced and the magnitude of each component discussed. Finally, the consistency of experimental determination of $N_{D,w}$ coefficients is discussed. It is expected that the implementation of this addendum will be straightforward, assuming that the user is already familiar with TG-51. The changes introduced by this report are generally minor, although new recommendations could result in procedural changes for individual users. It is expected that the effort on the medical physicist's part to implement this addendum will not be significant and could be done as part of the annual linac calibration. © *2014 American Association of Physicists in Medicine*. [http://dx.doi.org/10.1118/1.4866223]

Key words: photon beams, dosimetry protocol, ionization chamber, beam quality conversion factors, kQ, uncertainty analysis, absorbed dose calibration coefficients

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1. INTRODUCTION AND RATIONALE

The AAPM's TG-51 (Ref. 1) protocol for the determination of absorbed dose to water in megavoltage photon and electron beams was published in 1999. In replacing the previous exposure based protocol [TG-21 (Ref. 2)] it adopted the k_Q formalism[3] whereby linac calibrations were based on a ^{60}Co absorbed dose to water calibration coefficient, traceable to a primary standard, and calculated k_Q factors were used to convert from ^{60}Co to linac photon and electron beam qualities.

In the years following its publication, TG-51 was extensively tested and compared with other protocols based on both absorbed-dose and air-kerma standards.[4–11] Although its accuracy and applicability to the calibration of clinical linacs has not been challenged, developments have occurred in the fifteen years since the TG-51 protocol was published that necessitate updating and expanding the protocol:

(i) The protocol lists calculated k_Q factors for only 18 cylindrical chamber types, which represented the majority of reference chambers available at the time of publication. Today, a user has a choice of more than 30 different designs. TG-51 addresses how one might use the data given in the protocol to obtain k_Q factors for other chambers (see Sec. 11 of the protocol), but it is not an ideal method to have to follow and one that could lead to errors in the determination of absorbed dose.

(ii) Monte Carlo radiation transport algorithms such as EGSnrc (Ref. 12) and PENELOPE (Ref. 13) have been developed and accurately benchmarked[14–17] for calculations of detailed chamber geometries.[18–21] The calculations given in TG-51 are based on a semianalytic approach that does not take all the details of the chamber geometry into account.[22] Factors determined using a full Monte Carlo simulation of the chamber details better reflect the true geometry of the chamber and should be more accurate than the semianalytical algorithm used in TG-51.

(iii) Significant progress has been made by many primary standards laboratories on the development of primary standards for megavoltage photon and electron beams.[23,24] Measured k_Q factors could replace the calculated values as given in TG-51 or at least allow a better estimate of the accuracy of such calculations.

(iv) What is a suitable reference-class ionization chamber is not always easy to establish. In North America, reference dosimetry in the radiotherapy clinic has traditionally been carried out using one class of geometrically similar ionization chambers, based on the design of Aird and Farmer.[25] Of the 18 chamber types listed in Table I of the TG-51 protocol, nine are this "Farmer-type" or derivatives. Some of the chamber types now available have specific applications (for example, the microionization chambers with volumes less than 0.01 cm^3 designed for measurements in very small radiosurgery beams), but it might not be obvious from the manufacturer's data sheets if such a chamber is suitable for other applications such as reference dosimetry. In an effort to use a "one size fits all" detector, a user might therefore be tempted to employ a small chamber for reference dosimetry, beam scanning, and small-field applications. An increase in the calibration of such chambers at the US ADCLs supports such a concern. The main issue is that there is little in the literature available on the application of such chambers to reference dosimetry.

In light of these issues the AAPM set up a Working Group to review TG-51. The review of photon beam dosimetry has resulted in this Addendum to TG-51 that addresses the following:

(i) The need for calculated k_Q factors for new reference ionization chambers developed after the publication of TG-51 and revised values for other chambers.
(ii) A comparison of these (and similar) calculations with measured k_Q factors obtained at primary standards laboratories.
(iii) Specification of a reference chamber.
(iv) Guidance on the implementation of TG-51, including information relevant to new developments in linac technology that still fall within the application of TG-51 (specifically, flattening-filter-free linacs).
(v) A discussion of uncertainties, with emphasis on the components of an uncertainty budget and how the clinical physicist can affect the combined uncertainty in the measurement of absorbed dose to water.

To prevent potentially confusing overlaps with other AAPM working groups and Task Groups the decision was taken to maintain the reference field definition of TG-51 (specifically, SSD or SAD setup (usually 100 cm), field size $= 10 \times 10$ cm). This means that the protocol and addendum *cannot* be applied directly to the calibration of photon beams from certain treatment machines (e.g., Gamma Knife®, CyberKnife®, TomoTherapy®). As noted above, in reviewing the literature, the Working Group concluded that flattening-filter-free linacs did fall within the TG-51 specification, although the peaked radial dose distribution presents an additional challenge.

Section 3 provides new recommended k_Q data and a comparison with previous calculations and recent measurements. Section 4 provides some guidance and clarification on the implementation of the TG-51 protocol. Section 5 discusses uncertainties and uncertainty analysis as it relates to the measurement of absorbed dose to water in the user's beam. Appendix A specifies criteria for reference-class ionization chambers for the measurement of absorbed dose in megavoltage photon beams using the TG-51 protocol. Appendix B presents background on primary standards for megavoltage photon beams and other national and international dosimetry protocols.

This addendum has been reviewed and approved by the AAPM Calibration Laboratory Accreditation Subcommittee and by the AAPM Therapy Physics Committee. If certain commercial products are identified in this report, such identification does not imply recommendation or endorsement by the AAPM or the National Institute of Standards and Technology (NIST), nor does it imply that the product is necessarily the best available for these purposes. The intended audience of this report is the clinical medical physicist concerned with reference dosimetry of radiotherapy beams using the methodology of the AAPM TG-51 protocol.

2. NOTATIONS AND DEFINITIONS

(1) The notations and definitions as laid out in the TG-51 protocol are used in this report except that $N_{D,w}$ is now referred to as a calibration coefficient, consistent with current international practice.

(2) The majority of users will obtain ^{60}Co absorbed dose to water calibration coefficients from one of the US ADCLs, and this is therefore the default scenario in any following discussion. However, as users could obtain the required calibration from a primary standards laboratory (e.g., the National Research Council in Canada) or another secondary standard dosimetry laboratory, "ADCL" should be read as shorthand for all such calibration laboratories.

(3) Notations introduced in this addendum:

C_{init}: the component of the ion recombination correction factor, P_{ion}, to take account of initial recombination.

C_{gen}: the coefficient of general (volume) recombination. The product of C_{gen} and the dose per pulse, D_{pp}, is the component of the ion recombination correction factor, P_{ion}, to take account of general recombination. C_{init} and C_{gen} are defined such that the ion-recombination correction factor, $P_{ion} = 1 + C_{init} + C_{gen}D_{pp}$.

D_{pp}: the absorbed dose in the ionization chamber's sensitive volume per beam pulse from the linear accelerator. For a therapy-level ^{60}Co beam general recombination is assumed to be negligible.

P_{leak}: the correction factor to take account of leakage (defined as any contribution to the measured reading that is not due to ionization by the radiation beam in the chamber's collecting volume).

P_{rp}: the correction factor to take account of the variation of the radial dose distribution that is averaged by the detector.

3. k_Q FACTORS

3.A. Basis of calculated k_Q factors

This addendum provides the most accurate extensive set of k_Q factors calculated to date for ionization chambers. These are based on Monte Carlo calculations by Muir and Rogers[21] who carried out simulations using the EGSnrc Monte Carlo code system with the egs_chamber user-code of Wulff *et al.*[26] Geometries were modeled with the egs++ geometry package.[27] Chambers were modeled according to specifications from manufacturers' user manuals, catalogs, blueprint specifications where available, or models previously described in the literature.

The k_Q values were also calculated using the same computer program and physics as used for the original TG-51 values.[22] As Muir and Rogers show, the Monte Carlo calculated k_Q factors show generally very good agreement with the original TG-51 calculations, with differences of 0.5% or less. These new k_Q values have also been compared with other Monte Carlo calculated values in the literature.[18–20,28] There is very good agreement between the different Monte Carlo calculations with differences typically less than 0.4%.

Systematic uncertainties were also investigated by Muir and Rogers,[21] including those from uncertainties in photon cross sections, stopping powers, chamber dimensions, the use of photon spectra instead of full linac head models, and possible variation of $(W/e)_{air}$ with beam energy. Most relevant for the application of the protocol is the sensitivity of the calculated k_Q factors to changes in chamber dimensions (i.e., to address the question, "What if the user's chamber is not exactly as the blueprint specifies?"). The results indicated that, while geometry variations (especially volume) clearly affect the calibration coefficient, the k_Q factors are insensitive at the 0.1% level to reasonable changes (5%–10%) in both chamber-wall thickness and chamber volume.

3.B. Comparison of calculated and experimental k_Q factors

In addition to the investigations discussed in Appendix A, McEwen[29] carried out a wide-ranging comparison of

measured k_Q factors with those in TG-51. The quality conversion factors were obtained for 27 different types of cylindrical ionization chamber. Chambers were classified as "Farmer-type" (0.6 cm^3 thimble chambers and derivatives), "Scanning" (~0.1 cm^3 chambers typically used for beam commissioning with 3D scanning phantoms), and "Micro" (very small volume ionization chambers ~0.01 cm^3 used for small field dosimetry). As might be expected, Farmer-type chambers showed the most predictable performance, and experimental k_Q factors were obtained with a relative standard uncertainty of 0.3%. The performance of scanning and microchambers (specified below) was somewhat variable. Some chambers showed very good behavior but others showed anomalous polarity and recombination corrections that are not fully explained at present. Muir *et al.*[30] directly compared the latest MC-calculated k_Q factors with these measurements. Overall they found very good agreement, with differences typically less than 0.4% (and closer to 0.2% for low-energy MV beams).

The ability of the EGSnrc package to accurately model ionization chamber response is also demonstrated in the recent work of Swanpalmer and Johannson[31,32] in ^{60}Co and MV photon beams. Agreement between measurement and Monte Carlo calculation is reported at the 0.1%–0.2% level. Benmakhlouf and Andreo have rightly counseled that "the MC technique is not a magic black box"[33] but the level of agreement with experiment shown by Swanpalmer and Johansson and Muir *et al.* significantly increases confidence in the use of MC-calculated k_Q factors.

3.C. Recommended chambers

The range of chambers available from manufacturers has grown significantly since TG-51 was published. A number of chambers have also seen small design changes (often not visible to the user) that have resulted in new model numbers. The decision taken by this Working Group in providing new k_Q factors was to consider only chambers *currently available*. For users having chambers for which k_Q factors are not given in either the original TG-51 protocol or in this addendum, the recommendations of Sec. 11 of TG-51 still apply. With the extensive list of chambers given here it should be relatively straightforward to determine k_Q factors for any chamber not included. Manufacturers can provide guidance on how obsolete chamber designs relate to those currently available.

This addendum follows the TG-51 recommendation that only cylindrical chambers should be used for photon-beam reference dosimetry. Although Christ *et al.*[34] claim very good performance of some parallel-plate chambers in ^{60}Co beams and stability comparable to Farmer chambers, it does not necessarily mean that performance in MV photon beams will be similarly acceptable. McEwen *et al.*[35] reported poor performance of parallel-plate chambers in linac photon beams, compared to thimble chambers. More recently, Kapsch and Gomola[36] reported somewhat better results for two specific chamber types, but chamber-to-chamber variations were still larger than for cylindrical chambers. Muir *et al.*[37] investigated a wide range of parallel-plate chambers and compared measured and calculated k_Q factors. Their conclusions were that some chamber types met the requirements of a reference-class ionization chamber but that there were still concerns over aspects of chamber performance, particularly chamber-to-chamber variations and long-term stability.

However, this recommendation does not preclude the use of parallel-plate chambers for beam-quality measurements. McEwen *et al.*[38] showed that such chambers are ideal for measuring depth-dose curves as they have a well-defined effective point of measurement and provide the best agreement with Monte Carlo calculations, particularly in the build-up portion of the depth-dose curve.

To provide guidance in the context of the increased range of cylindrical chambers available, this addendum defines a specification (see Appendix A) for a reference-class ionization chamber (modified from that proposed by McEwen[29]). Although the specification can be applied at the user level to test individual chambers, its primary purpose is to classify types of ionization chamber as fit for the purpose of reference dosimetry. Application of this specification results in no data being presented here for chambers with a measurement volume, V, less than 0.05 cm^3. Chambers with $V < 0.05$ cm^3 include micro or Pinpoint^{TM} chamber types. Data reported by McEwen indicate that such small chambers do not show expected polarity or recombination behavior and are more sensitive to leakage currents and irradiation history. In addition, calculations show that those small volume chambers with high-Z electrodes exhibit behavior in beams without flattening filters[39] that are not well specified by $\%dd(10)_x$ (or $TPR_{20,10}$).

3.D. k_Q factors for MV photon beams

Table I lists the k_Q data for *chambers meeting the specification given in Appendix A*. The fit parameters from Muir and Rogers[21] are provided so that users can accurately evaluate k_Q factors for any specific beam quality according to:

$$k_Q = A + B \cdot 10^{-3} \cdot \%dd(10)_x + C \cdot 10^{-5} \cdot (\%dd(10)_x)^2$$
$$63 < \%dd(10)_X < 86. \quad (1)$$

The fits have an average rms deviation of 0.07% compared with the explicitly calculated k_Q values for each chamber in ten beams. Note that the fit is valid only for values of $\%dd(10)_X$ in the range $63 < \%dd(10)_X < 86$. For convenience, tabulated values for the most common beams, as collated by the Radiological Physics Center (Houston), are also given. For beam qualities below $\%dd(10)_X = 63$, users should linearly interpolate between the value tabulated for that beam quality and $k_Q = 1.000$ at ^{60}Co ($\%dd(10)_X = 58$).

3.E. Chambers not listed in Table I of this report

Table I lists only chambers that are currently being manufactured. There are therefore chambers listed in the TG-51 protocol, and potentially still in clinical use, for which new k_Q factors have not been calculated. In addition, there are a small number of new chambers for which MC-calculated k_Q

TABLE I. Recommended k_Q fitting parameters and factors as a function of the beam-quality specifier, $\%dd(10)_X$. These parameter values are taken from Monte Carlo calculations of Muir and Rogers (Ref. 21) and Muir *et al.* (Ref. 30) and are valid for $63 < \%dd(10)_X < 86$. Tabulated k_Q values are given for the most common beams (according to the RPC). Users are referred to the manufacturers' data sheets for the specifications of the chambers listed here (wall material and thickness, central electrode, etc.). Chambers requiring a waterproof sleeve were modeled with a 1 mm PMMA sleeve and are indicated below with a *.

Chamber type		Comment	Fitting parameters for Eq. (1) ($63 < \%dd(10)_X < 86$)			k_Q values for the most common beams (as function of beam-quality specifier $\%dd(10)_x$)				
			A	B	C	63	67	73	77	81
Capintec	PR-06C/G*	0.6 cc Farmer-type	0.9519	2.432	−2.704	0.998	0.993	0.985	0.979	0.971
Exradin	A19	Water proof Farmer	0.9934	1.384	−2.125	0.996	0.991	0.981	0.974	0.966
Exradin	A12	0.6 cc	1.0146	0.777	−1.666	0.997	0.992	0.983	0.976	0.968
Exradin	A12S	0.2 cc "short Farmer"	0.9692	1.974	−2.448	0.996	0.992	0.983	0.976	0.968
Exradin	A18	0.125 cc waterproof	0.9944	1.286	−1.980	0.997	0.992	0.983	0.976	0.969
Exradin	A1	0.06 cc waterproof	1.0029	1.023	−1.803	0.996	0.991	0.981	0.975	0.967
Exradin	A1SL	0.06 cc waterproof	0.9896	1.410	−2.049	0.997	0.992	0.983	0.977	0.969
NE	NE2561 *	0.3 cc NPL Sec. Std	0.9722	1.977	−2.463	0.999	0.994	0.985	0.978	0.971
NE	NE2571 *	0.6 cc Farmer	0.9882	1.486	−2.140	0.997	0.992	0.983	0.976	0.968
PTW	PTW30010*	0.6 cc Farmer-type	1.0093	0.926	−1.771	0.997	0.992	0.983	0.976	0.968
PTW	PTW30011*	0.6 cc Farmer-type	0.9676	2.061	−2.528	0.997	0.992	0.983	0.976	0.969
PTW	PTW30012*	0.6 cc Farmer-type	0.9537	2.440	−2.750	0.998	0.994	0.985	0.979	0.971
PTW	PTW30013	Waterproof Farmer	0.9652	2.141	−2.623	0.996	0.991	0.982	0.975	0.967
PTW	PTW31013	0.25 cc waterproof	0.9725	1.957	−2.498	0.997	0.992	0.982	0.975	0.967
IBA	FC65-G	Waterproof Farmer	0.9708	1.972	−2.480	0.997	0.992	0.983	0.976	0.968
IBA	FC65-P	Robust Farmer	0.9828	1.664	−2.296	0.997	0.991	0.982	0.975	0.967
IBA	FC23-C	0.2 cc "short Farmer"	0.9820	1.579	−2.166	0.996	0.991	0.982	0.975	0.968
IBA	CC25	0.25 cc waterproof	0.9551	2.353	−2.687	0.997	0.992	0.984	0.977	0.969
IBA	CC13	0.13 cc waterproof	0.9515	2.455	−2.768	0.996	0.992	0.983	0.976	0.969
IBA	CC08	0.08 cc waterproof	0.9430	2.637	−2.884	0.995	0.990	0.982	0.975	0.967

factors are not available. Some of the most common chamber types not included above are discussed here:

(i) Based on manufacturer's specifications, the previously manufactured **PTW31003** (listed in the TG-51 protocol) is identical to the PTW31013. The values for the PTW31013 in the table above can therefore be used for the type 31003.

(ii) The CC13 is the closest replacement for the **IC10** from IBA (Wellhöfer) listed in the TG-51 protocol, and there appears to be no significant change in construction. The k_Q values for the CC13 chamber in the table above can be used for the IC10 ionization chamber.

(iii) The **PTW30001**, **30002**, **30004,** and **30006** were replaced by the PTW30010, 30011, 30012, and 30013, respectively. This was more than a simple change in numbering; there was a significant change in the thimble design. The earlier chambers used a PMMA thimble with conductive graphite "dag" on the inner face. The dag was replaced in the newer design by a solid graphite liner to the PMMA thimble.[40] Users of these older PTW ionization chamber models (30001, 30002, 30004) should use the k_Q data in the original TG-51 protocol. Section 11 of TG-51 indicates that for the 30006 model (not listed in Table I of TG-51) one can use the k_Q data given for the 30001.

(iv) The **Capintec PR-05/PR-05P** as listed in TG-51 is still manufactured, but there is little or no data in the literature on its performance for reference dosimetry. Data from the ADCLs indicate that this chamber type is still quite widely used, so continued use of such chambers is allowed using the data in the TG-51 protocol as long as the user verifies that the chamber meets the requirements of Table III in Appendix A.

(v) The **NE2581** chamber is no longer recommended for reference dosimetry. It has been shown by Mijnheer[41] that an A-150 walled chamber (such as the NE2581) can exhibit significant changes in the chamber volume as a function of the relative humidity, and Mayo and Gottschalk[42] showed that the temperature coefficient of A-150 could result in a significant change in thimble volume with temperature. This recommendation also applies to other A-150 chambers (e.g., the Exradin T1, an A-150-walled version of the A1 chamber) that are not already explicitly excluded in Sec. 4.C based on experimental evaluation.

4. IMPLEMENTATION GUIDANCE

4.A. Implementation of TG-51 addendum

Implementation of this addendum will be very simple compared to the effort that was required in moving from TG-21 to TG-51, for which the new formalism and method was very different from the former. The Radiological Physics Center (RPC) in Houston has monitored the implementation of TG-51 and reports that the vast majority of clinics in the US and Canada are now using TG-51 for linac reference

dosimetry, so this working group anticipates a rapid take-up of this addendum.

The changes introduced by this report in the determination of absorbed dose to water in megavoltage photon beams are very minor:

(1) The formalism has not changed in any way, and the procedure as set out in TG-51 remains the same.
(2) The major contribution is the provision of new, highly accurate values of k_Q for chambers in TG-51 as well as chambers developed since TG-51's publication. Second, recommendations about some chamber types to be avoided are also given.
(3) Some recommendations are provided in Sec. 5, which could result in procedural changes for individual clinics.
(4) More detailed measurements of ion recombination might be required for certain chamber types.

It is expected that the effort on the medical physicist's part to implement this addendum will not be significant and could be done as part of the annual linac calibration.

4.B. Reference-class ionization chamber

Any ionization chamber used to realize absorbed dose to water using the TG-51 protocol and this addendum should meet the specifications of Table III. Since these are operational (rather than mechanical/geometric) specifications, it is the user's responsibility to verify the performance of the chamber they intend to use. Note also that all chambers have a finite lifetime, and therefore chamber performance should be verified on a regular basis. It is not sufficient to make measurements when the chamber is commissioned and then assume that corrections such as polarity, recombination, and leakage continue to apply. Based on the practices of the ADCLs and the RPC, and the recommendations of AAPM Report TG-142 (Klein *et al.*[43]), this Working Group recommends that all aspects of chamber performance be verified at the annual TG-51 calibration *and* whenever a new irradiation beam is commissioned.

4.C. Equipment needed

The Appendix to TG-51 lists the minimum equipment required to implement this protocol. This working group endorses that list, where it applies to photon dosimetry, with the following comments:

(1) The ionization chamber must meet the reference-class specification as detailed in Sec. 4.B above.
(2) Information on redundancy/stability checks is given in Sec. 5.B.4.
(3) See Sec. 5.A.5 below for resolution specifications and expected precision for barometers and thermometers.
(4) A lead foil is no longer mandatory for *certain beams* with energies greater than 10 MV, and this is detailed in Sec. 4.H below.

4.D. k_Q data sets

For chambers listed in both this addendum and the original TG-51 protocol, the k_Q factors listed in Table I of this document should be used. For chambers that are not listed in either the original TG-51 protocol or in this addendum, the recommendations of Sec. 11 of TG-51 should be followed as long as the chamber meets the requirements of Table III.

4.E. Choice of polarizing voltage

It is the user's responsibility to choose the correct polarizing voltage for the calibration and use of an ionization chamber, noting that:

(a) The calibration coefficient is valid only for the polarizing voltage stated on the calibration certificate issued by the calibration laboratory (i.e., the same polarizing voltage should always be used).
(b) High polarizing voltages (e.g., in an attempt to minimize the magnitude of P_{ion}) can lead to excessive charge collection, resulting in an incorrect measured and/or applied ion-recombination correction P_{ion} and, ultimately, electrical breakdown within the air cavity.
(c) A manufacturers' safe rating for a chamber is not the same as the correct value to use for accurate dosimetric measurements (i.e., a value at which the ionization chamber is behaving according to the standard recombination theory).

Based on data in the literature this addendum recommends an upper limit of 300 V for cylindrical chambers used for reference dosimetry.

4.F. Measurement of polarity correction, P_{pol}

The polarity correction should be measured for any new chamber and beam combination, and then at least annually. The polarity effect is associated with a net deposition of charge in the chamber (electrodes and/or insulators and/or cable), and in the situation of transient charged-particle equilibrium there is no net charge deposited. P_{pol} values between 0.996 and 1.004 should therefore be expected. This range is slightly larger than given in TG-51 and is based on the chamber specification defined in Appendix A. For values outside this range, the procedure in Sec. 7.A of TG-51 should be followed. The measurement of P_{pol} is a very simple QA check of the chamber/electrometer system as it confirms that the polarizing voltage is applied correctly between the chamber's electrodes, and any change with time indicates a possible change in chamber response.

4.G. Effective point of measurement

For TG-51 photon-beam dosimetry, the effective point of measurement (EPOM) is required *only* for the measurement of depth-dose curves. In Sec. 8 of the TG-51 protocol, a shift from the geometric center of the chamber toward the radiation

source of $0.6r_{cav}$ for cylindrical chambers was recommended (curve II in Fig. 1 of the TG-51 protocol). Kawrakow[44] carried out a Monte Carlo-based investigation of the effective point of measurement of cylindrical ionization chambers and showed that the shift, or EPOM, was a function of cavity length and central-electrode diameter. Tessier and Kawrakow[45] then calculated the effective point of measurement for a wide range of chambers currently available, and Looe *et al.*,[46] using an experimental method based on radiochromic film, presented measured results in good agreement with these calculations. In many cases, a shift significantly different from $0.6r_{cav}$ was required to yield the "true" depth-dose curve.

As published EPOM data are not available for all the chambers listed in Secs. 3.D and 3.E, the baseline recommendation for determining $\%dd(10)_X$ for use with the TG-51 Addendum is to use a shift of $0.6r_{cav}$ toward the radiation source. This maintains consistency with the current TG-51 recommendation. Using $0.6r_{cav}$ compared to the more accurate values in the literature has a small effect on the subsequent selection of k_Q factors. Hence the effect on the dose determined at the reference point is less than 0.1% for the chambers in Table I.

For accurate measurement of the full photon beam depth-dose curve for other clinical applications, the chamber-specific EPOM values given in the literature (if available) should be used.

4.H. Use of lead foil to determine $\%dd(10)_X$

Determination of the photon beam-quality specifier, $\%dd(10)_X$, for high-energy beams (around 10 MV and above) requires the placement of a 1 mm lead foil to fully intercept the beam. Although TG-51 clearly states that the foil must be removed for the dose measurement step, there is anecdotal evidence of confusion as to when the lead foil must be used. TG-51 also provided an "interim measure" [Eq. (15) in TG-51] to convert directly from $\%dd(10)$ (measured in the open field, no lead foil required) to $\%dd(10)_X$, and Tailor *et al.*[47] report that using this relation introduces an error of no more than 0.2% in the selection of k_Q. Since the simplified experimental procedure should be less prone to operator error, the Working Group recommends that Eq. (15) in TG-51 (no lead foil required) can be used as the default procedure, with the provisos that (i) it be used only for linacs with flattening filters, and (ii) an increased uncertainty component be included to take account of its use compared to the more accurate method using the lead foil.

4.I. Use of small-volume chambers in relative dosimetry

As discussed above, very small chambers (volumes <0.05 cm^3) are not recommended for reference dosimetry. In addition, it should be noted that many of the issues reported for these chambers (anomalous recombination behavior, large polarity effect, or complex effects from high-Z electrodes) can also impact relative dosimetry measurements (such as measurement of depth-dose curves or beam profiles). Hence careful characterization of such chambers (e.g., as described by McEwen[29]) must be done before use in any situation.

4.J. Non-water phantoms prohibited

This addendum upholds the recommendations of TG-51 that a water phantom *must* be used for the output calibration of the linear accelerator. The main advantage of solid phantoms is in the ease of setup, but this is countered by the uncertainty in the correction factor to convert from dose in plastic to dose in water (particularly for generic materials). Despite the developments in the formulation of water-equivalent phantoms that have been documented in the literature,[48–51] the additional uncertainty in using such materials still negates any ease-of-use issue (in particular, temperature equilibration times and temperature gradients can be much greater than for a water phantom[52]). Small calibration water phantoms are available from a range of manufacturers. These are simple to use, weigh less than 30 kg when full and provide the necessary positioning capability to measure the beam-quality specifier and to position the chamber at d_{ref}.

4.K. Application to flattening-filter-free linacs

The factors and methods listed in the TG-51 protocol and this addendum can be used for flattening-filter-free (FFF) linacs, assuming that the standard reference field (i.e., SSD or SAD = 100 cm, field size = 10 × 10 cm) can be produced. Xiong and Rogers[53] have shown that $\%dd(10)_X$ is a suitable beam-quality specifier, and therefore TG-51 can be followed to determine the reference absorbed dose to water. Additional points to note for FFF beams are:

(1) The ion-recombination correction is larger than found for "standard" linacs. The effect of the higher dose per pulse from a FFF linac on the behavior of the user's ionization chamber should be investigated. Kry *et al.*[54] showed that, at least for one type of FFF linac, the ion-recombination correction is consistent with the correction for lower dose-per-pulse beams and is accurately determined using the same techniques. A working limit of $P_{ion} \leq 1.05$, as suggested in TG-51, is recommended for FFF linacs, which could, for certain FFF beams, restrict the choice of reference chamber type.
(2) The significant radial nonuniformity of the beam can have an effect on volume averaging within the ion-chamber volume. It is recommended that chambers used in FFF beams should have a short collecting volume (the chamber should still meet the specification of a reference-class instrument, as given in Appendix A). Alternatively, a correction must be applied to account for the beam nonuniformity across the chamber volume (see Sec. 5.C.7). Such a correction should include an associated uncertainty component.
(3) Measurement of $\%dd(10)_X$ requires the use of the 1 mm Pb foil, as Eq. (15) in the TG-51 protocol and

Sec. 4.H above are known to be valid only for "flat" beams, whereas the equations for using lead foils are based on calculations with filtered and unfiltered beams.[55] It is recommended that the lead foil be used in beam quality measurements for all FFF beams (including those below the 10 MV limit of TG-51) to eliminate the potential effect of accelerator-produced electron contamination. The geometry of the detector should not have a significant effect on the measurement of beam quality.

(4) The k_Q values presented in Table I are based on calculations for beams with flattening filters only.[20,28] In a study of central-electrode effects,[37] it was shown that these same values apply for FFF beams within 0.1% or so for chambers with low-Z or aluminum electrodes. However, for chambers with high-Z electrodes, values of k_Q can vary by more than 1% in FFF beams for a given $\%dd(10)_X$. This is another reason these microchambers are not recommended for reference dosimetry.

4.L. Best-practice guidelines

This addendum is not intended to be a comprehensive best-practice guide. Over the last decade the RPC has, through its role in providing dosimetry-audit services, collated a large amount of information on implementing TG-51 correctly in the clinic and common mistakes to be avoided. Users are referred to the publications of Tailor *et al.*[47] and Followill.[56]

4.M. TG-51 corrigenda

There are a number of minor typos and formatting issues in the original TG-51 document:

(1) The equation for general recombination in a ^{60}Co beam on p. 1862 (Worksheet A, p. 2) is missing an exponent in the denominator and a closing bracket. See Eq. (11) in the main text of TG-51 for the correct equation.

(2) In the figures for k_Q, some of the curves are mislabeled (the tabulated data are correct). However, for the majority of chambers, the photon k_Q data in TG-51 are superseded by the data presented in this addendum.

5. UNCERTAINTIES

As part of any measurement it is essential to estimate the uncertainty in the result obtained. There is much in the literature that describes the evaluation of measurement uncertainties. Readers are referred to the ISO Guide on the Uncertainty in Measurement [ISO GUM,[57] also known as the JCGM Report 100 (Ref. 58)] for a comprehensive approach to uncertainties in general, to Mitch *et al.*[59] for detail on uncertainties in radiation dosimetry and to Castro *et al.*[60] for an example uncertainty budget derived for radiotherapy.

Here we examine the primary components of uncertainty in an absorbed dose to water measurement and describe how uncertainties can depend on the user's experimental method, the choice of chamber, etc. It cannot be overstated that all measurement uncertainty components ***must*** be evaluated by the user. Two basic equations of TG-51 are our focus for this discussion:

$$D_w^Q = M k_Q N_{D,w}^{^{60}Co}, \tag{2}$$

$$M = M_{raw}(x, y, z, SSD, FS) P_{TP} P_{ion} P_{pol} P_{elec} P_{leak} P_{rp}. \tag{3}$$

The meanings of all but two of these parameters are given in the TG-51 protocol. P_{leak} is the correction factor defined as any contribution to the measured reading that is not due to ionization released by the radiation beam in the chamber's collecting volume, and P_{rp} is the correction factor to take account of any off-axis variation in the intensity profile of the radiation field over the sensitive volume of the ionization chamber. The effect of chamber positioning is indicated in Eq. (3) by explicitly showing that the chamber reading is a function of position within the water phantom. The chamber reading is also a function of the source-surface distance, *SSD* (or source-axis distance for an SAD setup) and field size [*FS* in Eq. (3)], and these can be sources of uncertainty.

In the following, each term in Eqs. (2) and (3) is discussed in turn. Where uncertainty values are assigned, these are for clinical measurements and may be: (i) as reported in the literature, or (ii) based on the experience of the members of this Working Group. Unless stated otherwise, all values are given as standard uncertainties (as defined by the GUM). Users should evaluate all components themselves as applied to measurements made in the photon beams being calibrated at their own facilities. The aim here is not to provide a detailed analysis of how to carry out an uncertainty analysis but to give some guidance on measurement practice and indicate where the clinical physicist can make a significant impact on the overall measurement uncertainty. Two ***example*** uncertainty budgets are presented in Table II representing two realistic situations:

(i) All components of the uncertainty in the dose determination are evaluated; reference-class equipment is used.

(ii) Uncertainty components of the dose determination are evaluated using "typical" assumptions, and the performance of the equipment could be questionable.

In the text below we attempt, where possible, to indicate how these two values for each component are derived. Note that (ii) does *not* represent a worst case, but is intended to reflect a realistic clinical situation.

The user's uncertainty budget should not be constructed only once, but should be reevaluated whenever there is a significant change in the procedure and/or equipment used. In this discussion, a vertical-beam geometry is assumed, as this is the most common calibration setup in the radiotherapy clinic. It is also assumed that linac QA, as set out in AAPM's

TABLE II. Example uncertainty budgets for two *realistic* situations in which the absorbed dose to water is determined at the reference depth[a]. All values are given as relative standard uncertainties (i.e., $k = 1$). The combined estimate assumes that all components are uncorrelated.

Component of uncertainty	Section in this report	Situation (i): all components of the dose determination are evaluated; reference-class equipment is used	Situation (ii): uncertainty components of the dose determination are evaluated using "typical" assumptions, and the performance of the equipment could be questionable[b]
		Value	Value
		Measurement	
SSD setting	V.A.1	0.10%	0.4%
Depth setting	V.A.2	0.17%	0.5%
Field-size setting	V.A.3	0.10%	0.5%
Charge measurement	V.A.4	0.23%	0.5%
P_{TP} correction	V.A.5	0.10%	0.4%
Humidity	V.A.6	0.05%	0.15%
		Calibration data	
^{60}Co $N_{D,w}$ [c]	V.B.1	0.75%	0.75%
k_Q factor	V.B.2	0.4%	0.5%
Assignment of k_Q factor	V.B.3	0.10%	0.6%
Stability of reference chamber	V.B.4	0.05%	0.5%
		Influence quantities	
P_{pol}	V.C.1	0.05%	0.5%
P_{ion}	V.C.2	0.10%	0.5%
Preirradiation history	V.C.3	0.10%	1.0%
Leakage current	V.C.4	0.05%	0.3%
Linac stability	V.C.5	0.05%	0.2%
P_{elec}	V.C.6	0.07%	0.25%
P_{rp}	V.C.8	0.05%	0.4%
COMBINED ($k = 1$)		0.9%	2.1%
User-dependent part[d]		0.3%	1.8%

[a]These values are given as examples only and must *not* be taken as representative of any real calibration situation.
[b]Situation (ii) does not reflect a "worst case" but is based on reported and anecdotal activities in a realistic clinical situation in which time is limited and assumptions that might not be valid are made about the equipment.
[c]The uncertainty in the ^{60}Co $N_{D,w}$ coefficient is the value given by the AAPM ADCLs. It may be different for other calibration laboratories.
[d]The "user-dependent part" refers to the components that depend on the procedure and equipment of the clinical physicist.

TG-142 report,[43] is being carried out (e.g., with regard to mechanical isocenter). All uncertainties are quoted as standard uncertainties according to JCGM Report 100 (Ref. 58).

5.A. Measurement

5.A.1. SSD setting, $M_{raw}(x,y,z,SSD,FS)$

With the mechanical positioning devices available with modern linacs (e.g., a mechanical front pointer or distance stick) and careful technique, the *SSD* can be determined with a measurement precision of 0.5 mm. If we take this as the standard uncertainty in setting the *SSD*, then for a typical 100 cm *SSD* machine, the inverse-square law yields a relative uncertainty of 0.1% in $M_{raw}(x,y,z)$. The use of the light-field distance indicator is not recommended for reference dosimetry measurements, as the error in using such a system with a water surface (as in a vertical-beam geometry) is likely to be in the range 1.5–3 mm. Laser systems provide a convenient method of rapidly verifying the *SSD*, but, as the AAPM TG-106 (Ref. 61) report notes, the accuracy of a laser system must be verified first.

The assumption here, and in the AAPM's TG-40 (Ref. 62) and TG-142 (Ref. 43) reports, is that the mechanical front pointer correctly indicates the *SSD*. This is verified at linac commissioning and should be confirmed as part of annual linac QA checks. *SSD* consistency between beam calibration and regular use is more important than a pointer setting of "100" denoting an exact *SSD* of 100 cm.

If an SAD setup is used, then the concerns are the same since some position has to be mechanically defined and so the same equipment is likely to be used.

Table II values based on: (i) 0.5 mm; (ii) 2 mm uncertainty in *SSD* precision.

5.A.2. Positioning the chamber at d_{ref}, $M_{raw}(x,y,z,SSD,FS)$

Modern calibration phantoms typically have a resolution of 0.1 mm, but the largest uncertainty is in defining the origin for the phantom positioning system-–usually with the chamber at the water surface. Das *et al.*[61] suggest a simple optical method to do this, while Tailor and Tello[63] propose a mechanical device to position the chamber directly at d_{ref} with a claimed uncertainty of 0.1 mm. An analysis of Fig. 6 of TG-106 suggests that an uncertainty in d_{ref} of 0.33 mm should be achievable without any specialized equipment, but 0.5 mm is perhaps more realistic. With relative dose gradients at d_{ref} in MV photon beams in the range of 3–5% per cm, a 0.5 mm uncertainty in position leads to a relative standard uncertainty of 0.25% in M_{raw}. Ververs *et al.*[64] show that the origin can be determined retrospectively from depth-dose curves measured through the water surface, with an uncertainty potentially better than 0.25 mm, but this approach will be applicable only if depth-dose curves are acquired at the same time as making reference absorbed dose measurements.

Table II values based on: (i) 0.33 mm, (ii) 1.0 mm uncertainty in setting chamber at reference depth.

5.A.3. Setting the field size, $M_{raw}(x,y,z,SSD,FS)$

The fluence incident on the ionization chamber is sensitive to the field-size, so the absorbed dose to water determined using TG-51 is defined for a specific field size (i.e., 10 × 10 cm). It is a simple measurement to determine the sensitivity of M to the light-field size setting (*not* that indicated by the machine software). A value of around 1% per 1 cm change in field size (i.e., from 9.5 × 9.5 cm to 10.5 × 10.5 cm) is typical.

This measurement assumes that the light and radiation fields are congruent. A more accurate, but time-consuming, option would be to measure the radiation field size each time (e.g., using film, detector array, etc).

It is recognized that certain linacs cannot reproduce exactly a 10 × 10 cm field for a 100 cm *SSD* setup. The effect of variations at the 0.5 cm level from this specification for the determination of $\%dd(10)_x$ are typically less than 0.5%, and therefore the effect on the selection of k_Q values is less than 0.1%. The applicability of the ^{60}Co calibration coefficient (obtained in a 10 × 10 cm field) to such an accelerator field does not require an additional uncertainty component. Note that the uncertainty in setting the calibration field, as described in paragraph 1 of Sec. 5.A.3 must still be considered.

Table II values based on: (i) field size verified to be 10 × 10 cm at the 1 mm level in both dimensions; (ii) field size verified to be 10 × 10 cm at the 5 mm level.

5.A.4. Charge measurement, M_{raw}

The uncertainty in M_{raw} is due to the three components of the measurement system—chamber, extension cable, and electrometer (ignoring any variability in beam delivery discussed in Sec. 5.C.5). Klein *et al.*[43] indicate that a value of 0.05% is reasonable for the ionization-chamber component, and any extension cable should not contribute significantly (<0.02%), assuming that there is no fault in its operation (e.g., mechanical or radiation damage). IEC standard 60731 (Ref. 65) deals with the requirements for a reference-class electrometer, but the specification is rather lenient and allows a combined relative uncertainty of 1.6%. This potentially represents the dominant uncertainty in absorbed-dose measurement. The Institute for Physics and Engineering in Medicine (IPEM) recognized this, and Morgan *et al.*[66] set out a more stringent specification for a reference-class electrometer, including statements on repeatability, long-term stability, and nonlinearity. A check source (see Sec. 5.B.4 below) can be used to monitor long-term stability of the chamber-electrometer system as a whole. Alternatively, a current source could be used to monitor the electrometer independently of the ionization chamber.[67]

Due to the wide range of electrometers available, with differing modes of operation, it is not possible to provide detailed guidance on the uncertainty in the charge measurement. Morgan *et al.* provide much useful information on electrometer performance, and ADCLs can also provide charge calibration data to characterize the device.

Table II values based on: (i) Table III of Morgan *et al.*; (ii) electrometer meets the IEC 60731 standard, but there is limited evaluation of its performance.

5.A.5. Correction for cavity air temperature and pressure, P_{TP}

The evaluation of this correction requires a calibrated thermometer and barometer, both of which are readily available from a number of manufacturers. The recommended resolutions of measuring devices are 0.1 °C and 0.1 kPa, respectively. The use of air-pressure data from external sources (weather stations, airports, etc.) is not recommended. The water phantom should be in equilibrium with the room temperature to minimize temperature gradients within the phantom (e.g., use water stored at room temperature), and the temperature of the water must always be measured (even at equilibrium the water temperature will differ from the room-air temperature). A working definition of equilibrium is that the drift in the water phantom temperature is less than 1 °C/h. Christ *et al.*[68] give much detail on the correction for air density. For megavoltage photon beams, problems with the pressure correction for low-energy photons[69–71] are not significant. With calibrated equipment having sufficient resolution (0.1 °C, 0.1 kPa), it is estimated that P_{TP} can be evaluated with a relative standard uncertainty of 0.1%.

Related to the temperature-pressure correction is the thermal history of the chamber. As has been shown,[72,73] the time required for an ionization chamber to come to thermal equilibrium with the water phantom can be significant. Sufficient time must be allowed, after the ionization chamber is positioned in the water phantom (which will be chamber- and situation-dependent), to ensure that the uncertainty due to this effect is negligible. Combining the results from Das

and Zhu[73] with those of McEwen[29] on the time required for pre-irradiation of ionization chambers (Sec. 5.C.3), the conclusion is that positioning the chamber at the reference depth and preirradiating for 10 min will achieve both stability requirements.

To be complete, one must also consider the thermal expansion of the chamber thimble.[42,73,74] However, although measurable, the effect is small ($<0.04\%\ ^\circ C^{-1}$). Unless the water temperature is "extreme" (defined as outside the normal range of 15 °C–25 °C), this effect should be insignificant.

Table II values based on: (i) calibrated equipment used to evaluate P_{TP}, chamber and water allowed to come to thermal equilibrium with environment; (ii) uncalibrated equipment used, system not in thermal equilibrium.

5.A.6. Humidity

The effect of humidity is generally ignored in clinical reference dosimetry because over a wide range of humidity it leads to, at most, a 0.15% error.[75] Calibration laboratories usually define a range of humidity values for which the calibration coefficient is valid, and users should monitor the relative humidity of their treatment rooms to ensure that this range is met (affordable systems are readily available). The effect of extreme humidity values on equipment should also be considered (increased leakage, corrosion, etc.). Ideally, the relative humidity should be maintained in the range 40%–60%, and then the relative uncertainty due to humidity is 0.05%. A more realistic range of achievable relative humidity values is 20%–80% (the range specified in TG-51), and this leads to a relative uncertainty of about 0.15%.

There is the implicit assumption that the relative humidity measured in the room is the same as that of the chamber air volume in the water phantom. Although long-term immersion in water could lead to higher humidity levels in the chamber, there is unlikely to be any effect when following the TG-51 protocol.

Table II values based on: (i) humidity maintained in range 40%–60%; (ii) humidity varying over the larger range of 20%–80%.

5.B. Calibration data

5.B.1. Calibration certificate, $N_{D,w}^{^{60}Co}$

With the possible exception of the choice of calibration laboratory, the user has no influence on the uncertainty of this component. Note, however, that calibration certificates usually report the expanded ($k = 2$) uncertainty. This uncertainty should be converted to $k = 1$ for combination with the other uncertainty components in this analysis.

Table II values based on values disseminated by US ADCLs.

5.B.2. Quality conversion factor, k_Q

This component refers to the uncertainty inherent in the calculations used to provide the data in Table I. Rogers[22] estimates that the relative uncertainty in k_Q values in TG-51 is 0.5% based on the level of agreement with measured values, and the extensive measured data of McEwen are consistent with this value *for recommended chambers*. Muir and Rogers[21] and Wulff *et al.*[76] have investigated the systematic uncertainties in Monte Carlo calculations of k_Q factors and indicate that the relative standard uncertainty in k_Q for an NE2571 chamber is of the order of 0.3%, ignoring any variation in the energy dependence of $(W/e)_{air}$. Muir *et al.*[30] used the comparison between MC-calculated k_Q factors and measurements to obtain an upper limit on the variation of W/e from ^{60}Co to 25 MV. Combining this value of 0.29% with the previous analysis yielded a standard relative uncertainty in k_Q for reference chambers of 0.4%.

Table II values based on: (i) Wulff *et al.* and Muir *et al.* analyses for MC-calculated k_Q factors; (ii) uncertainty estimated by Rogers for k_Q factors listed in TG-51.

5.B.3. Assignment of k_Q factor

This uncertainty component must be determined by the user. It is a composite uncertainty, comprising the following steps:

(i) measurement of $\%dd(10)_X$;
(ii) selection of k_Q data for the user's chamber from Table I;
(iii) interpolation of tabulated data to user's beam quality.

The selection of k_Q factors is rather insensitive to the absolute value of $\%dd(10)_X$: a 1% change in $\%dd(10)_X$ leads to a ~0.15% change in k_Q. However, the user is cautioned to use the $\%dd(10)_X$ measured at the time of calibration and not use a "standard" value (e.g., the planning system's clinical value) as this could introduce a larger error. The largest uncertainty in this factor can be the correction for electron contamination, but data from the RPC suggest the relative uncertainty in determining $\%dd(10)_X$ is at most 2%, which corresponds to a relative uncertainty in k_Q of about 0.25%.[77] The largest uncertainty in assigning k_Q is most likely due to the selection of tabulated data and interpolation. For chambers listed in Table I, the relative standard uncertainty achievable should be approximately 0.1% (or less if using the polynomial fits); for chambers not listed, the relative uncertainty can be as large as 0.5%, depending on the user's application of Sec. 11 of TG-51, although likely to be much less in a 6 MV beam.

Table II values based on: (i) accurate determination of $\%dd(10)_X$, use of fit parameters in Table I; (ii) 2% uncertainty in $\%dd(10)_X$, k_Q factor derived for unlisted chamber.

5.B.4. Stability of reference chamber

Between calibrations at an ADCL, the long-term stability of the reference chamber should be monitored by the user. The best method for such monitoring is to use a ^{60}Co beam with the reference conditions as defined in TG-51. However, with the elimination of ^{60}Co units from most North American radiotherapy clinics, this is not a realistic option. The method recommended is a regular comparison with other reference-class ionization chambers in a linac beam. At least three

chambers are required to make such a comparison method robust (although it is not necessary to have all three calibrated at an ADCL). A third option is to use a ^{90}Sr check source, although the uncertainties reported for such measurements[78,79] are significantly larger than those reported for ^{60}Co measurements.[29] Degradation of an ionization chamber can be monitored by the comparison of calibration coefficients obtained in 50 kV and in ^{60}Co beams. Measurements in kV beams are more sensitive than MV beams to documented issues such as contamination of the chamber thimble or corrosion of the central electrode[80] and are available at the US ADCLs. A change in the ratio of calibration coefficients would indicate a possible problem with the chamber.

Without such stability checks, the additional relative uncertainty due to this component (for a reference-class instrument) will be typically 0.3%–0.5 %, but can be significantly larger depending on the specific chamber type.

Table II values based on: (i) reference chambers regularly monitored using a ^{60}Co beam; (ii) no monitoring of chamber between recalibrations, chamber drift outside range defined in Table I.

5.C. Influence quantities

5.C.1. Polarity effect, P_{pol}

It should be possible to determine the polarity correction with an uncertainty similar to that of M_{raw} (Sec. 5.A.4). If the measurement of the polarity effect is carried out correctly, the relative uncertainty in the correction can be as low as 0.05%. However, the uncertainty can increase significantly if sufficient time is not allowed for the chamber to reach equilibrium after the polarizing voltage is changed. A longer series of measurements might be required to confirm this. Ideally, one should also repeat the first polarizing voltage to correct for any overall drift of the chamber (or linac) response.[31]

Table II values based on data reported in the literature[29,60] for different chamber types.

5.C.2. Ion recombination, P_{ion}

The TG-51 protocol recommends the use of the 2-voltage technique to evaluate the recombination correction, but the underlying assumption of this method is that the chamber is behaving as predicted by Boag theory.[81] The 2-voltage technique itself cannot demonstrate such behavior, and a number of authors have shown that chambers can show non-ideal behavior.[82–84] In extreme examples, this can lead to relative errors in the evaluation of P_{ion} of several percent. Characterization at a range of polarizing voltages and dose-per-pulse values (e.g., as described by Bruggmoser *et al.*[85] and Palmans *et al.*[86]) should be performed before a chamber is used for reference dosimetry measurements. As part of this, the optimal value of the polarizing voltage for the particular chamber should be identified (see Appendix A). Note that this might be lower than the "standard" value used in the clinic or calibration laboratory. If a chamber has been shown to behave ideally, the 2-voltage technique yields the recombination correction with a relative uncertainty of 0.1% or better,[29] assuming that enough time is allowed for the chamber to equilibrate at each value of the polarizing voltage (see Sec. 5.C.3).

The introduction of flattening-filter-free linacs introduces a further problem in that the dose per pulse for these machines can be very high, leading to a large recombination correction. Care is required to accurately characterize the recombination correction for such linac beams, for which the mode of operation can be very different from that of a conventional (with-flattening-filter) accelerator.[54]

Table II values based on: (i) full characterization of recombination behavior of ionization chamber; (ii) poor procedure and/or chamber that does not behave according to theory.

5.C.3. Pre-irradiation history

It has previously been shown[87] that ionization chambers can show vastly different equilibration behaviors under irradiation. Some chambers achieve a stable reading very quickly while others take much longer. McEwen[29] reports that in pulsed linac beams, chambers reach equilibration within 10 min at a nominal dose rate of 2.5 Gy min^{-1}. However, the chamber irradiation history can have a significant effect (e.g., a longer waiting time might be required for a chamber that has not been used for some time or after a change in the polarity of the polarizing voltage). If this effect is correctly taken into account (i.e., sufficient data are acquired to ensure that equilibration has occurred) the relative uncertainty should be of the order of 0.1%. However, if it is ignored, some chambers have shown effects that would lead to a relative error of up to 1%.

Table II values based on: (i) chamber irradiated to dose of $>$10 Gy prior to measurement, stability monitored during dose measurements; (ii) no pre-irradiation, no time allowed after a switch in polarizing voltage, error expressed as Type B uncertainty.

5.C.4. Leakage currents

The leakage current is usually measured with all the equipment in place, the accelerator on, but with no beam. Contributions can come from the chamber itself, the ionization chamber cable (e.g., due to damage or long-term radiation-induced degradation), or the electrometer. In such a measurement, extra-cameral currents and radiation-induced leakage (e.g., in the cable) are not evaluated, although the definition of P_{leak} does include those components. Extra-cameral currents can be estimated by shielding the ionization chamber while irradiating the cable, and Campos and Caldas[88] demonstrated a method to evaluate such radiation-induced leakage. Each component of the chamber-cable-electrometer system should be evaluated separately, when possible. A reasonable and achievable target for the value of P_{leak} is that given in the IAEA TRS-398 Code of Practice:[89] the leakage should contribute less than 0.1% to the charge reading. For any particular system, the value of the leakage current could be significantly larger, and a value greater than 0.5% must be investigated. If the leakage current is at or below the 0.1% level then it is

reasonable to set $P_{leak} = 1.000$ (no correction for leakage) with an associated relative uncertainty of 0.1%.

Table II values based on data obtained from dosimetry calibration laboratories.

5.C.5. Stability of linear accelerator

The TG-51 protocol lays out a method to determine the absorbed dose to water for a single irradiation. Practically, this has no useful meaning unless it can be related to some measure of the treatment delivery (e.g., exposure time for ^{60}Co, number of MUs for a linear accelerator). An uncertainty component, therefore, is included to address the short-term repeatability of the linac when delivering a series of fixed MU runs [IEC/TR 60977 (Ref. 90)]. From reviewing the literature this component appears to be small for modern linacs, typically less than 0.05% (Ref. 43). Note that this component specifically deals with the linac stability for the period of the measurements required to carry out the dose-determination part of TG-51. An additional component might be required if the beam quality and dose measurements are not measured at the same time (i.e., to take account of any variation in beam quality with time). Long-term reproducibility of the accelerator output is not part of this procedure and is covered by other publications, such as TG-142 (Ref. 43).

The number of MU (monitor units) for each irradiation has typically been chosen to be the same as used in treatment. This has historically led to calibration deliveries of 100–200 MU. This range is recommended, as longer irradiations can lead to over-ranging of the electrometer, while shorter irradiations can be significantly affected by beam start-up.

Table II values based on: (i) data reported for modern linacs; (ii) linac with reduced stability (but still within manufacturers operating limits).

5.C.6. Electrometer calibration coefficient, P_{elec}

The US ADCLs calibrate ionization chambers and electrometers separately, and therefore the calibration certificate will give a value for P_{elec} and its associated uncertainty. However, some calibration laboratories (e.g., NRC in Canada) calibrate a complete system—ionization chamber plus electrometer—so there is no separate electrometer factor (and therefore no associated uncertainty).

Table II values based on: (i) US ADCL calibration capability; (ii) value reported by Castro *et al.*[60]

5.C.7. Radial beam profile, P_{rp}

An ionization chamber averages over some volume (dependent on the particular chamber) but the measurement is directly related to the absorbed dose to water at a point. At the reference depth, the axial dose fall-off is close to linear, so the focus is the radial beam uniformity or flatness. The AAPM's TG-142 report[43] defines flatness criteria for the whole field, but does not deal with the averaging effect of the ionization chamber. The TG-106 report[56] shows the large effect volume averaging has on the measurement of the penumbra, but the effect on the determination of the reference dose should also be considered. The issue of the "horns" on a MV beam profile is usually noted for large fields, but the operation of the accelerator combined with the design of the flattening filter can result in a significant nonuniformity of the beam (1%–2%) even for a 10 × 10 cm field.[91,92] The effect is greatest for long-thimbled Farmer-type chambers.

This correction factor is included in Eq. (3) above, as a correction to the raw ionization-chamber reading. Conceptually, however, it does not address some non-ideal behavior of the ionization-chamber/electrometer system but rather corrects for the non-ideal dose distribution from the radiation field. One could therefore equally consider it as a component of Eq. (2), the conversion from ionization-chamber reading to absorbed dose to water at a point. It was felt to be simpler, and less confusing, to introduce the two new correction factors (P_{leak} and P_{rp}) in the same equation.

However, there is an additional complication because the k_Q factors presented in this report are based on simulations of the photon beam and ionization-chamber geometry. The impact of the radial-dose profile of the simulated beams on the ionization-chamber reading is already taken into account, and therefore a correction to the measured ionization-chamber reading would seem not to be needed. This is not the case, however, as the calculated dose profiles are much more uniform than typical measured profiles. The effect on the calculated k_Q factors of the dose profile is estimated to be of the order of 0.05% (worst case) which, for the purposes of this discussion, is considered to be insignificant.[110] In contrast, the correction for the measured P_{rp} can be an order of magnitude larger.

To determine P_{rp} in the clinic, one calculates the average of the radial dose profile over the dimensions of the active part of the chamber. For linacs with a flattening filter, a simple 1D integration (average) along the thimble should be sufficient as this dominates over averaging across the air cavity. The magnitude of P_{rp} can be significantly larger for FFF beams because the unflattened beam is strongly peaked along the central axis[93] and therefore a more detailed 2D measurement and calculation might be required for such beams. Note that the measurement of the radial dose profile might require a water phantom different from that normally used for implementation of TG-51. In addition, the effect on the radial-dose profile due to small changes in linac beam steering can be larger for a FFF beam than for a flattened beam;[94] thus, constancy of the correction factor P_{rp} over time for either type of beam should not be assumed.

Table II values based on: (i) P_{rp} derived from detailed 2D radial beam scan; (ii) field assumed to be uniform over dimensions of chamber cavity.

5.D. Combined uncertainty

What level of uncertainty is achievable? This is very much dependent on the user and their equipment. There is some dependence on the calibration laboratory used to obtain the ^{60}Co absorbed-dose calibration coefficient, but the variability in published uncertainties among calibration laboratories (assuming that it is accredited to ISO17025,[95] or is part of the

AAPM ADCL or IAEA SSDL networks) is small. The variability among users, however, can be very significant. With care, and following the recommendations in this addendum, a combined relative standard uncertainty in the determination of absorbed dose to water at the reference point can be as low as 1%. However, if the procedure is followed without due care with a chamber that has not been characterized, a figure two to three times greater is possible. Example uncertainty budgets are shown in Table II. *These values are given as examples only and must not be taken as representative of any real calibration scenario*. For any particular situation there can be additional uncertainty components. For example, calibration in a horizontal-beam geometry would require a component to take into account the uncertainty associated with the non-water material of the phantom wall.

Table II shows that the clinical physicist can have a significant impact on the combined uncertainty. By using calibrated equipment and the correct procedures, the contribution from the measurement protocol becomes a small fraction of the total uncertainty.

6. CONCLUSION

This addendum for reference dosimetry of megavoltage photon beams is to be used in conjunction with the AAPM's TG-51 dosimetry protocol. The procedure outlined in that document is used as the basis of this addendum, but new k_Q data for photon beams, based on Monte Carlo simulations, are presented and recommendations are given to improve the accuracy and consistency of implementation that might change clinical practice slightly. The components of the uncertainty budget in determining absorbed dose to water at the reference point are introduced and the magnitude of each component discussed. Finally, the experimental determination of $N_{D,w}$ coefficients worldwide is discussed.

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APPENDIX A: SPECIFICATION OF A REFERENCE-CLASS IONIZATION CHAMBER FOR THE MEASUREMENT OF ABSORBED DOSE IN MEGAVOLTGE PHOTON BEAMS USING THE TG-51 PROTOCOL

This specification was developed from that by McEwen.[29] It is not a requirement of this addendum that users must test individual chambers against the specification (although this would constitute a comprehensive commissioning of the instrument). The specification has been used specifically to determine which types of ionization chambers are included in Table II. The aspects of chamber performance identified as being crucial to determining reference-class behavior were:

Chamber stabilization – both in terms of the irradiation time required to obtain an equilibrium reading and the difference between the initial chamber reading per monitor unit when the radiation beam is switched on and the final equilibrium reading. Note: this does not refer to the stabilization of the linac dose-rate, which generally has no effect on the measured dose/MU ratio.

Leakage current – defined as any contribution to the measured reading that is not due to ionization released by the radiation beam in the chamber's collecting volume.

Polarity correction (P_{pol}) – both in terms of the magnitude of the correction (as discussed in the TG-51 protocol) and the variation in the correction with photon energy.

Recombination correction (P_{ion}) – the maximum polarizing voltage for which the 2-voltage technique applies must be determined for each specific chamber and the initial and general recombination components need to be evaluated.

Chamber stability – long-term chamber stability needs to be consistent with the calibration interval and/or user-specific monitoring procedures.

As McCaffrey *et al.*[87] have shown, chamber stabilization can be significant for certain chamber types. There are two major concerns: (i) if the time required to reach equilibrium is significant, then the limited time often available in the clinical setting could impact the measurement, and it is more likely that an error could be introduced due to a difference in how the chamber is calibrated by the ADCL and how the chamber is used. To minimize the impact of any settling effect one could set a reasonable limit that there must be less than a 0.5% change in reading per monitor unit from beam-on to stabilization and that equilibration should be achieved in less than 5 min.

Leakage currents (as discussed in Sec. 5.C.4) are generally not significant, but can be an issue for smaller-volume chambers and can vary significantly depending on the equipment used (electrometer, cable, connectors, etc.).

The polarity correction for thimble chambers in photon beams is often assumed to be negligible, and one expects a value close to unity as no net charge is deposited in the ionization chamber (a different situation exists for electron beams where the charge of the incident beam can impact the polarity measurement). In the TG-51 protocol document, a typical range for P_{pol} of 0.997–1.003 is given. Based on the data for chambers available now, the recommendation is that the polarity correction should differ from unity by less than 0.4% (in either direction) at any energy *and* that the total variation in the polarity correction across the photon energy range of interest (^{60}Co to 25 MV) be less than 0.5%.

The determination of the ion-recombination correction for cylindrical chambers has generated significant liter-

ature over recent years,[81–86] highlighting a number of issues that impact the selection of a chamber as reference quality:

(i) The TG-51 protocol uses the 2-voltage technique, which has the underlying assumption that the chamber follows the Boag theory of ion recombination.[81] The practical demonstration of this is that a plot of 1/charge-reading (1/*M*) *versus* 1/polarizing-voltage (1/*V*) should be linear for a range of polarizing voltages, and the polarizing voltage used to obtain M_{raw} [Eq. (3)] should be within the linear region for the user's chamber. The upper limit on the polarizing voltage is dependent on the chamber type, but can also vary between chambers of the same type. Deviation of the 1/*M vs.* 1/*V* plot from a straight line could indicate charge multiplication, and Palmans *et al.*[86] give a detailed procedure on how to evaluate this. It is possible that no linear region of the 1/*M vs.* 1/*V* plot can be found (e.g., due to the particular chamber type or the range of voltages available from the user's electrometer), and in that case the procedure of Palmans *et al.* should be followed.

(ii) P_{ion} should vary linearly with the dose per pulse (and the slope should be positive and consistent with the electric field within the air cavity). McEwen[29] showed that linearity was generally obtained but found that the gradient for certain chamber types was not what was expected, based on the chamber geometry and polarizing voltage. It might be possible to change the dose per pulse through the linac control system but the simplest method is to change the *SSD* and/or measurement depth while maintaining approximately the same field size (to minimize variations in stem/cable irradiation).

(iii) The third point relates to initial recombination. From the definition of P_{ion} in Sec. 2 ($P_{ion} = 1 + C_{init} + C_{gen}D_{pp}$), C_{init} in a pulsed linac beam can be obtained from the intercept of a plot of P_{ion} *vs.* dose-per-pulse. Muir *et al.*[37] have shown that C_{init} obtained this way for plane-parallel chambers is indeed the same as the recombination correction obtained in a continuous ^{60}Co beam, where D_{pp} can be taken to be zero (i.e., general recombination in therapy-level ^{60}Co beams can be considered to be negligible), and this can also be assumed to be the case for reference-class cylindrical chambers. This means that a user can verify consistency of chamber recombination between calibration at the ADCL and use in the clinic. One expects the initial component to be small ($C_{init} \leq 0.002$), and a large value would indicate nonstandard behavior.[77] Note that C_{init} is inversely proportional to the polarizing voltage: e.g., $C_{init} = 0.002$ at 300 V becomes $C_{init} = 0.004$ at 150 V.

(iv) The final point is that researchers have shown that the recombination correction can be polarity dependent (e.g., see Fig. 6 of McEwen[29]). As it is not practical to define a universal measurement polarity to ensure consistency of P_{ion} determination, this introduces another component of the chamber specification.

In addition to these performance characteristics, and perhaps most importantly, a reference chamber should be stable over the period between calibrations at a primary standards laboratory (PSDL), ADCL, or other secondary standard calibration laboratory (typically two years). The minimum requirement[47] is that the change in calibration coefficient should be less than 0.3%.

Taking these components together, the Working Group proposes the specifications in Table III. A reference-class instrument should meet all of the criteria.

TABLE III. Specification of a reference-class ionization chamber for megavoltage photon-beam dosimetry. Note that upper-limit values at the reference depth are given, not standard uncertainties.

Measurand[a]	Specification
Chamber settling	Should be less than a 0.5% change in chamber reading per monitor unit from beam-on for a warmed up machine, to stabilization of the ionization chamber.
P_{leak}	< 0.1 % of chamber reading ($0.999 < P_{leak} < 1.001$)
P_{pol}	< 0.4 % correction ($0.996 < P_{pol} < 1.004$)
	< 0.5 % maximum variation in P_{pol} with energy (total range)
$P_{ion} = 1 + C_{init} + C_{gen}D_{pp}$[b]	
General	P_{ion} should be linear with dose per pulse.
Initial	Initial recombination should be less than 0.2%, that is, $C_{init} < 0.002$, for the TG-51 reference conditions[c].
Polarity dependence	Difference in initial-recombination correction between opposite polarities should be less than 0.1%.
Chamber stability	Should exhibit less than a 0.3%[d] change in calibration coefficient over the typical recalibration period of 2 years.

[a]Refer to McEwen (Ref. 29) for details on how each parameter was evaluated.
[b]Both initial and general recombination need to be considered.
[c]Value derived from data presented by McEwen (Ref. 29).
[d]This value is derived from calibration data from dosimetry calibration laboratories.

APPENDIX B: REVIEW OF PROGRESS IN REFERENCE DOSIMETRY SINCE THE RELEASE OF THE TG-51 PROTOCOL

1. Development of standards

The field of primary standards for megavoltage photon beams has progressed steadily in the last decade. Interested readers are referred to the proceedings of absorbed-dose workshops[23,96,97] or the review papers published in recent years.[24,98,99] Different standards are maintained by different PSDLs, based on ionometry, graphite and water calorimetry, and chemical dosimetry. Within the international network of PSDLs, this variety provides a more robust basis for the dissemination of absorbed dose than is currently the case for ^{60}Co air-kerma standards (for which all national standards are based on the same measurement technique and therefore share any systematic errors). As the focus of this report is reference dosimetry in North America, it should be noted that both the National Institute of Standards and Technology (NIST) and the National Research Council of Canada (NRC) maintain primary-standard water calorimeters.

International comparisons of absorbed dose (and air kerma) are co-ordinated through the Bureau International des Poids et Mesures (BIPM) and have focused to date on kV x-ray beams and ^{60}Co. Figure 1 shows data for 13 PSDLs, relative to the BIPM's standard (which is defined as the reference, equal to unity, for comparison purposes). The unweighted mean value for the quantity (D_{lab}/D_{BIPM}) is 0.9976, with a standard deviation of 0.0020. Alternatively, one can remove the reference value and look at the agreement between the different PSDLs, yielding an average difference between *any two* laboratories (D_{lab_m}/D_{lab_n}) of 0.21%, and this value is independent of the type of primary standard (i.e., no difference between graphite and water calorimeters). The conclusion, therefore, is that primary laboratories worldwide realize absorbed dose to water in a ^{60}Co beam with a consistency of better than 0.3%. Between the US (NIST) and Canada (NRC), in particular, the standards agree with each other within 0.15%.

For MV photon beams there have been a number of *ad hoc* bi-lateral comparisons of standards[100,101] based on the exchange of calibrated ionization chambers but, until recently, no co-ordinated program. In 2009, the BIPM embarked on a large-scale linac-based comparison using a transportable graphite calorimeter to directly compare absorbed-dose standards, and results from one of the first comparisons are very encouraging, showing agreement between the BIPM and participating PSDL within the combined uncertainties in three high-energy photon beams (6–25 MV) (Ref. 92).

As part of the development of an NCS Code of Practice (Report 18) for high-energy photon and electron beams, Aalbers *et al.*[102] carried out a comprehensive analysis of published k_Q data. Figure 2 is taken from that report, showing data for the NE2571 chamber.

The data covered a large time period, involved both primary and secondary dosimeter systems, and used quite different methodologies, so it is difficult to determine the differences between individual laboratories. However, there is good agreement overall between the k_Q values obtained at different laboratories. As was the case for ^{60}Co, the data do not support a difference between different types of standards (e.g., water and graphite calorimeters). The linear fit shown in Fig. 2 is an unweighted fit to all the data, and the rms relative deviation is about 0.3%. Muir *et al.*[30] show data from a more recent comparison of PSDLs (Ref. 103) in which the same ionization chambers were sent to all participants. This should lead to greater consistency in the determined k_Q factors, and the results from that comparison indicate a rms relative deviation of 0.24% between the fit and all data.

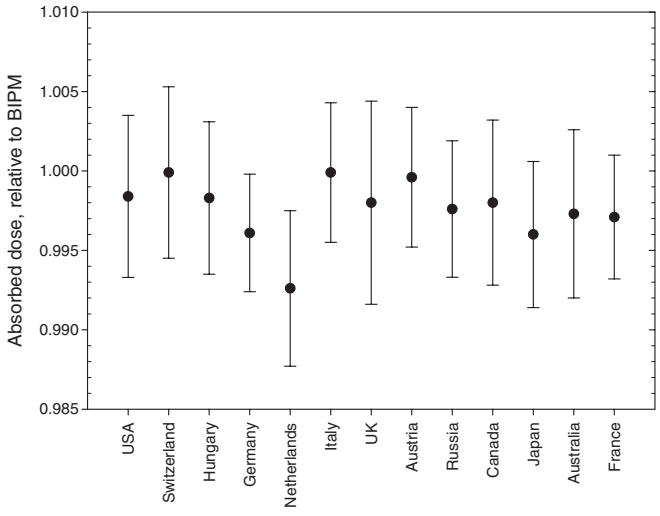

FIG. 1. Comparison of primary standards for ^{60}Co absorbed dose to water (as of the end of 2013). The uncertainty bars are given as the standard uncertainty and are the combined uncertainty for each standard compared to the BIPM standard. Data taken from BIPM's key comparison database (KCDB) database (www.bipm.org). This database is continually updated as comparisons are carried out.

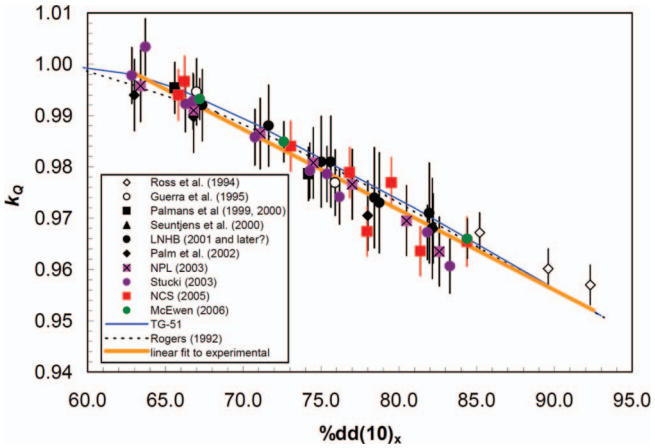

FIG. 2. k_Q factors for the NE2571 chamber as a function of beam-quality specifier $\%dd(10)_x$. The references are those given in the report by Aalbers *et al.* (Ref. 102). Uncertainties are given as one standard uncertainty (Ref. 58). ^{60}Co is assigned a $\%dd(10)_x$ value of 58.4. A value of 85.0 represents the highest linac energy typically found in radiotherapy clinics. Figure courtesy of NCS.

These comparisons allow an estimation of the best (current) relative uncertainty that could be achieved for ^{60}Co $N_{D,w}$ and k_Q of 0.29% and 0.24%, respectively. This would reduce the combined relative uncertainty in column (i) of Table II above to 0.6%.

An alternative option, available in a number of countries, is the direct calibration of ionization chambers in megavoltage beams. $N^{Q}_{D,w}$ then replaces $\{k_Q N^{^{60}Co}_{D,w}\}$ in the TG-51 formalism, the main advantage of which is that the ionization chamber is calibrated in a radiation beam very similar to that in which it will be used. Inherent in the k_Q formalism are the assumptions that "real" ionization chambers (i) behave in the way predicted by calculated k_Q factors, and (ii) exhibit the same behavior in the low-dose-rate continuous ^{60}Co beam at the calibration laboratory and in the (potentially) high-dose-rate pulsed linac beam in the clinic. By using pulsed MV beams for calibration (together with ^{60}Co), the chamber is verified to be "fit for purpose" and the applicability of calculated k_Q factors is confirmed. The combined uncertainty might not be significantly reduced, but as a QA process it can be very useful. Duane and Simon[104] summarized the results of direct calibrations in the UK. More recently, Andreo *et al.*[105] reviewed the dosimetry chain from primary standard to clinical dose measurement and concluded that chamber-specific calibration coefficients were preferable. However, in their analysis of the calculated k_Q factors presented in this addendum, they showed that there was excellent agreement, within the stated uncertainties, of these values with measured data from a number of PSDLs, confirming the findings of Muir *et al.*[30] Since the options for obtaining direct calibrations in MV beams are very limited, such comparisons of measured and calculated k_Q factors provide additional confidence in the TG-51 approach.

2. Photon beam-quality specifiers

A photon beam qualifier should, based on a simple measurement, allow the unique assignment of a calibration coefficient or k_Q factor. A number of different beam-quality specifiers have been proposed, with differing degrees of success, but in the end only two alternatives were adopted: $TPR_{20,10}$ (as used by IPSM 1990,[106] DIN6800-2,[107] IAEA TRS-398), and $\%dd(10)_X$,[108] as adopted in TG-51. Kalach and Rogers[109] showed that for realistic heavily filtered clinical spectra there was effectively no difference, dosimetricaly, in using either parameter, and this was confirmed by Aalbers *et al.*[102] Figure 3 shows data from Stucki *et al.*[103] This figure indicates that the heavily filtered beams used by PSDLs for the primary realization of absorbed dose to water and the calibration of ionization chambers are equivalent to typical clinical beams with flattening filters. From this, the relative uncertainty in transferring a MV calibration from the calibration laboratory to the clinic (i.e., how well the two beams match for dosimetry purposes) is estimated to be no more than 0.15%.

For flattening-filter-free (FFF) beams that are seeing increased use for IMRT deliveries, Xiong and Rogers[53] showed that $\%dd(10)_X$ remains an accurate beam-quality specifier

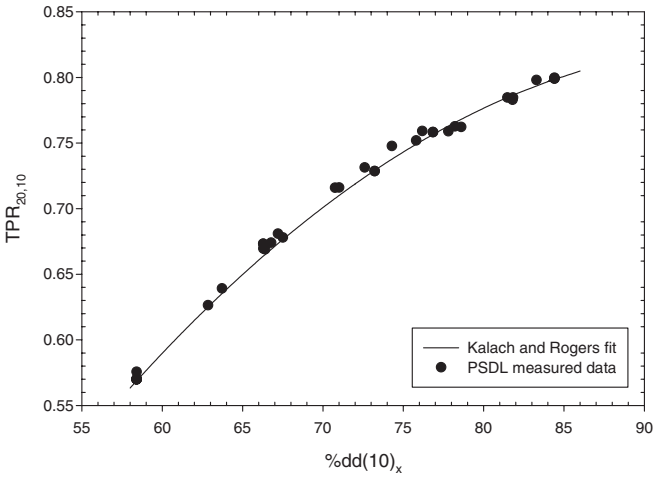

FIG. 3. Beam-quality data (Ref. 103) from six primary standard laboratories (individual laboratories are not identified) compared to the fit obtained by Kalach and Rogers (Ref. 107) for simulated data from heavily filtered clinical linac beams. The rms relative deviation from the fit is 0.7 %.

when selecting k_Q. There is one proviso to this statement: Muir and Rogers[39] show that for ionization chambers with high-Z electrodes (steel, copper, silver) neither $\%dd(10)_X$ nor $TPR_{20,10}$ works well for such lightly filtered beams. However, as detailed in Sec. 3 above, none of these chamber types are recommended for reference dosimetry.

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