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TOWARDS A DOSIMETRY SYSTEM BASED ON ABSORBED DOSE STANDARDS

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Abstract

TOWARDS A DOSIMETRY SYSTEM BASED ON ABSORBED DOSE STANDARDS.

A review is given of the rationale for establishing primary standards of absorbed dose in accelerator photon beams and the progress at the National Research Council Canada (NRC) towards establishing these standards. Attention is drawn to the problems that currently exist with primary standards of air kerma in ^{60}Co beams. The world system of air kerma standards is not very robust because they are all very similar and hence possibly subject to undetected common systematic errors. The uncertainties in these standards have been underestimated because the uncertainties in the value of $(W/e)_{air}$ and in the graphite to air stopping power ratios in a ^{60}Co beam have been underestimated. The value of the product $(W/e)_{air}s_{gr,\,air}$ which is used in air kerma standards is almost entirely based on graphite calorimeter measurements. A dosimetry system based on these standards is dependent on external physical data which are subject to change, in particular the electron stopping powers of graphite, water and air. In contrast, primary standards of absorbed dose to water are based on a variety of measurement techniques (graphite and water calorimeters, ionometric measurements and total energy absorption in Fricke solution). These constitute a very robust system which can eliminate systematic errors and a dosimetry system based on them is independent of changeable external data. The NRC primary standard for absorbed dose is based on water calorimetry and transfer to a point in water using Fricke dosimetry. Recent work has indicated larger effects from the Fricke vial walls than previously realized. Another serious problem to overcome is that of photon beam quality specification. It is proposed that percentage depth dose at 10 cm in a 10 × 10 cm^2 field at an SSD of 100 cm is a better specifier than TPR_{10}^{20} and effects of electron contamination can be taken into account using a 1 mm lead foil. Finally, it is argued that clinical dosimetry should be based on absorbed dose calibrations in a ^{60}Co beam and correction of the absorbed dose calibration factor to the beam quality of interest using a single correction factor, k_Q. This simplifies the calibration procedure compared to accelerator calibrations, and greatly simplifies clinical protocols, thereby improving accuracy in the field.

1. INTRODUCTION

Most current clinical dosimetry is based on air kerma standards and protocols to establish the absorbed dose to water in clinical radiotherapy beams. However, this approach has several problems. From the perspective of the clinical user, the dosimetry protocols are excessively complex and may reduce the accuracy of clinical dosimetry as a result of mistakes. From the perspective of a standards laboratory, the air kerma standards themselves have problems because of changes in the theory of cavity ion chambers and uncertainties in some of the physical data required. The basic problem is that all the standards are based on the same measurement technique and are thus potentially subject to common errors. These problems are discussed in Section 2.

In contrast, many primary standards laboratories are developing standards of absorbed dose to water. These are based on different approaches and thus a much more robust system is being put in place. In developing the water calorimetry based standard at the National Research Council Canada (NRC) a variety of problems have been studied (thermal heat defect of water, effects of Fricke vial walls and beam quality specification). The solutions to these problems will be discussed because they have broader implications.

Once a high energy absorbed dose calibration service is established, the question is how to use it most effectively. Since it is much more expensive to calibrate chambers in accelerator beams than in ^{60}Co beams, and because the vast majority of dosimetry standards laboratories do not have linear accelerators, any dosimetry protocol based on absorbed dose calibration factors should be based on calibrations in a ^{60}Co beam. Clinical dosimetry based on such a system is much simpler than that based on air kerma calibrations. The German dosimetry protocol is already based on such an approach [1]. The American Association of Physicists in Medicine (AAPM) has a task group (TG-51) which is investigating the feasibility of a dosimetry protocol based on this approach.

2. AIR KERMA STANDARDS AND DOSIMETRY BASED ON THEM

Major primary standards for air kerma in ^{60}Co beams are all based on graphite walled ion chambers. This is a problem because it is hard to detect systematic errors which may affect this type of measurement. Recent work on cavity chamber theory has uncovered two significant errors which have affected many primary standards. It has been shown that the linear extrapolation to zero wall thickness of ion chamber response leads to wall attenuation and scatter correction factors which are wrong by up to 1% [2]. Corrections for the point of measurement in a point source field have also been shown to be wrong by up to 1% [3, 4]. Taken together, these two changes imply that air kerma standards increase on average by

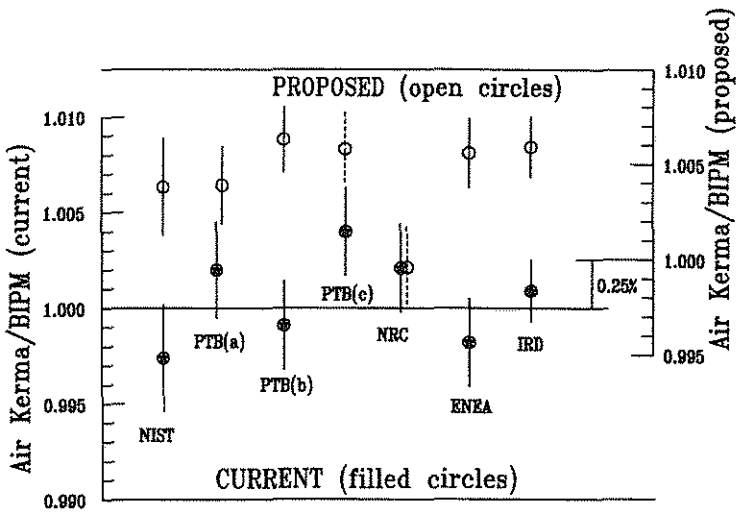

FIG. 1. Comparison of ratios of various primary standards of air kerma to that of the BIPM using either the original (left hand scale) or recently proposed (right hand scale) corrections for wall attenuation and scatter and point of measurement (from Ref. [5]). The average air kerma increases by about 0.6% but the spread and RMS deviation stay about the same.

about 0.6%, as shown in Fig. 1. The very good consistency between the primary standards is maintained after both changes are applied [5]. These changes demonstrate the fragility of a system in which all standards are based on the same measurement technique.

A similar concern is that all air kerma standards require knowledge of the product of $s_{gr,air}$, the graphite to air stopping power ratio in a ^{60}Co beam, and $(W/e)_{air}$, the energy deposited in dry air by electrons slowing down. A recent re-evaluation of the data on $(W/e)_{air}$ implies a decrease by 0.24% to 33.89 $\pm$ 0.07 J/C [6], which is a significant change compared to the previously stated uncertainty of 0.15% [7], and the uncertainty of $\pm$0.07 J/C (1σ) only applies when the product with $s_{gr,air}$ is used. The uncertainty on $(W/e)_{air}$ itself is $\pm$0.13 J/C ($\pm$0.38%).

Perhaps a more fundamental reservation concerns the internal logic of a system based on air kerma standards. Over 90% of the weight in determining the product $(W/e)_{air}s_{gr,air}$ is from measurements involving graphite absorbed dose calorimeters [6]. Thus air kerma standards can be said to be based on graphite calorimeters and one proposal at this symposium makes this fundamental relationship explicit [8]. If the air kerma standard is used as the basis of a dosimetry protocol which determines absorbed dose to water, then, at least as applied in a ^{60}Co beam, the air kerma standard amounts to one component of a very complicated transfer from absorbed dose to graphite to absorbed dose to water. This does not affect the ability of the system to assign absorbed dose accurately, but it certainly does make it complex!

A more practical concern about dosimetry systems based on air kerma standards is that they require the value of $s_{gr,air}$ in a ^{60}Co beam[1] and the uncertainty on this quantity is at least $\pm 0.7\%$, mostly from the uncertainty on the mean excitation energy of graphite given in Report 37 of the International Commission on Radiation Units and Measurements (ICRU) ($I = 78 \pm 7$ eV) [10]. More disturbing is a recent high quality measurement of the I value of graphite ($I = 86.9 \pm 1.2$ eV [11]) which implies a 1.2% reduction in the ^{60}Co graphite to air stopping power ratio and thus major changes in $(W/e)_{air}$ [6]. This emphasizes the point that the dosimetry chain based on air kerma standards is dependent on knowledge of quantities which are not measured or controlled within the measurement system itself. Furthermore, suggestions that $(W/e)_{air}$ may vary with beam quality [9] add further uncertainty to air kerma based dosimetry systems.

Ion chamber buildup caps which are never used in the clinic add another unnecessary complexity to the air kerma based systems since corrections for and details about the buildup cap play an important role in the protocol. The theory concerning these buildup caps is not well investigated (see e.g. Ref. [12]).

In short, clinical dosimetry based on air kerma standards is unnecessarily complex, conceptually awkward, based on a set of standards which are not very robust and dependent on externally determined radiation related quantities. However, one argument which is raised in defence of air kerma based systems is that they show remarkable consistency. This is a desirable goal, but the arguments given above indicate that some of this consistency may be fortuitous. More importantly, just because one can use any one of several primary standards or any one of several dosimetry protocols based on air kerma standards and obtain very similar results in any given beam quality, this tells us nothing about how consistent these results are for differing beam qualities — i.e. 1 Gy in a ^{60}Co beam may be very different from 1 Gy in a 20 MV photon beam and this is unacceptable.

3. ABSORBED DOSE STANDARDS

In contrast to the conceptual and practical problems with air kerma standards, there is a wide variety of absorbed dose to water standards being developed (for general reviews, see Refs [13, 14]). One significant advantage of these standards is that they are based on different techniques such as absorbed dose to graphite calorimeters, absorbed dose to water calorimeters, water calorimetry, energy absorption in Fricke solution and ionometric methods. Thus it is much more likely that systematic errors can be minimized or removed.

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[1] The product $(W/e)_{air} s_{gr,air}$ required for the air kerma standard is measured, but the final dose to any medium besides graphite requires knowledge of $(W/e)_{air}$ alone and hence one needs $s_{gr,air}$ to extract it (see Refs [6, 9]).

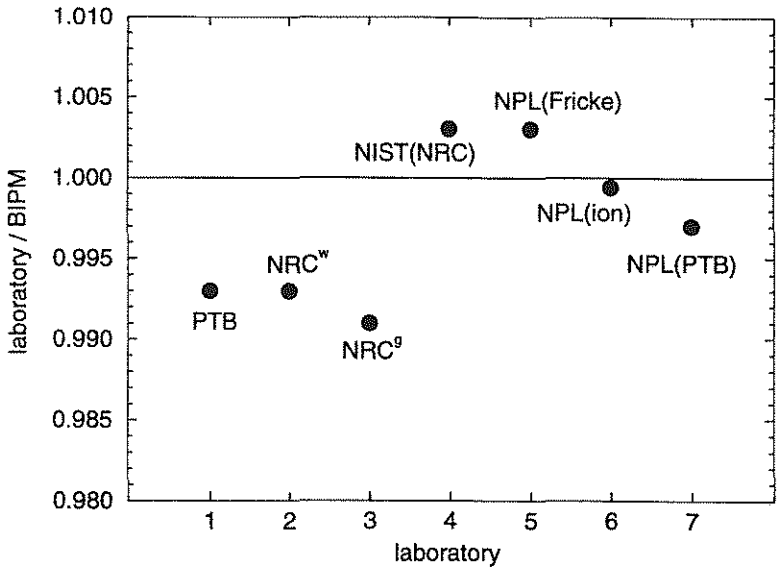

FIG. 2. Results of comparisons of ^{60}Co absorbed dose standards, shown as the ratio to the standard at the BIPM (data deduced from Ref. [14]). The NRC^{w} value is based on the NRC 20 MV water calorimeter calibration of Fricke and the NRC^{g} value is based on a graphite calorimeter with gap corrections. The NIST value is deduced from a NIST/NRC comparison and the NRC^{w}/BIPM result. The NPL(Fricke) and NPL(ion) values are based on two comparisons using Fricke or ion chambers as transfer instruments. The NPL(PTB) value is based on an NPL/PTB comparison and the known PTB/BIPM result.

A variety of comparisons of these standards have been done (e.g. Refs [14, 15]). Figure 2 presents a summary of results for comparisons in a ^{60}Co beam and less extensive data are also available for accelerator beams [14]. Although the apparent variations between the absorbed dose standards are currently greater than for air kerma standards, these comparisons are testing all of the various systematic uncertainties whereas the comparisons of air kerma standards are not testing many of the potential systematic uncertainties (values of $(W/e)_{air}$, $s_{gr,air}$, A_{wall}, point of measurement, etc.). Some of the differences found between the current absorbed dose standards reflect known problems which are described below. Not all of the solutions have been taken into account in all standards yet, but there should be significant progress in the next few years.

In the future the world will have a remarkably robust system of absorbed dose standards in accelerator photon beams with accuracies of $\pm 1\%$ or better (1σ). This implies a significant improvement in clinical dosimetry where the overall uncertainty using the International Atomic Energy Agency (IAEA) Code of Practice is given as ± 3 to $\pm 4\%$ [16]. At the same time, current indications are that the new standards

are in reasonable accord with the results of dosimetry protocols and thus introduction of their use should not cause major changes in a clinic which has been properly applying the current protocols (e.g. Refs [16, 17]).

4. NRC ABSORBED DOSE STANDARD

At NRC we have made considerable progress towards developing primary standards of absorbed dose to water in high energy photon beams. These are based on water calorimetry and the use of Fricke dosimetry [18, 19]. Calorimetric methods are used to measure the average temperature rise in a 100 mL volume of thermally isolated and stirred high purity water which is saturated with various gas mixtures. Using the well known heat capacity of water and a calculated thermal heat defect for the aqueous solution, one can deduce the average absorbed dose to the water. The container is then filled with about the same quantity of Fricke solution which is irradiated to a known absorbed dose to water. This calibrates the Fricke solution in the beam quality of interest after various corrections are applied (e.g. for excess heat transfer from the thin walled glass container to the water, the differences between mass energy absorption coefficients in the water and Fricke volumes). The Fricke solution is then used in a standard NRC Fricke vial to determine the dose at the reference point in the photon beam quality of interest. This process avoids the need to make any assumptions about variations in the value of ϵG with beam quality although to date the standard has only been established in a 20 MV beam.

The overall uncertainty in the standard is $\pm 0.7\%$ (1σ). The 20 MV NRC standard beam is generated by a 20 MeV electron beam hitting a fully stopping aluminium target and is flattened with an aluminium filter [18, 19, 15]. Assuming that ϵG does not vary between 20 MV and ^{60}Co beams, the 20 MV standard has been shown to be in good agreement with the NRC ^{60}Co absorbed dose to water standard based on a graphite calorimeter (to within 0.2% if gap effect corrections are included) or with our Fricke system (to within 0.3% if the value of ϵG given in ICRU Report 35 [20] is used [19]).

In developing this standard we have investigated several problems which are discussed briefly in the next three sections.

4.1. Thermal heat defect of water

All calorimeter based standards require knowledge of the thermal heat defect of the absorbing medium, i.e. the percentage difference between the energy deposited by radiation and the amount of heat released. In the case of water calorimeters this defect can be significant and it depends sensitively on the water quality. We have studied the relative thermal heat defect of seven aqueous solutions and found that the calculated values agree well with experimental values, especially

for solutions in which OH radicals are scavenged. Also, the calculated value is almost independent of the parameters in the model for H_2 and H_2–O_2 mixtures [19]. The overall uncertainty in the calculated thermal heat defect is $\pm 0.5\%$. For completely pure and isolated water the calculated thermal heat defect is almost 0, and measurements at the Physikalisch-Technische Bundesanstalt (PTB) are consistent with this value to within measurement uncertainties of 0.5% [21, 22]. However, to obtain completely pure water requires significant preirradiation of the entire water volume, which makes large stagnant water calorimeters impractical. To overcome this problem, Domen has developed a large water absorbed dose calorimeter with a small sealed container of highly pure water in which the sensing elements are placed [23].

4.2. Fricke vial wall corrections

Although the potential for glass walled Fricke vials to affect the dose measured in the vial has been recognized for a long time (see ICRU Report 35 and references therein [20]), many standards laboratories have used quartz- or Pyrex-walled Fricke vials because they are sent out to clinics for measurements and only with quartz vials can chemical storage effects be avoided. In collaboration with Ma and Nahum, a series of Monte Carlo calculations were done at NRC which showed vial wall effects of up to 2% for our standard coin shaped quartz vials in 24 MV beams and about 1% effects for the thinner walled vials used by the United Kingdom National Physical Laboratory (NPL) and the PTB [24]. Figure 3 shows a comparison of calculated and measured values of the ratio of the dose to Fricke solution in vials made of quartz and polyethylene. The good agreement strongly supports the calculated values. The NRC standard includes these calculated corrections and the PTB recently began to apply the calculated corrections for its vials [15, 14]. The vial wall effect has also been shown to explain at least part of the difference between ion chamber and Fricke dosimetry at the NPL [24].

4.3. Photon beam quality specification

Another important issue with accelerator based standards is to ensure the equivalence of the beam quality, Q, in which calibrations are done and that in which the chamber is to be used. Although TPR_{10}^{20}, the standard beam quality specifier, is thought to specify stopping power ratios for 'typical' clinical beams [25], this does not necessarily apply to the beams used in standards laboratories nor to those delivered by the new scanned photon beam accelerators [26]. For example, Monte Carlo calculations have shown that for beams with the same TPR_{10}^{20} near 0.78, the stopping power ratio in the beam of the NRC 20 MV standard (generated by 20 MeV electrons on a stopping target of aluminium and a conical aluminium flattening filter) is about 0.7% less than in a more typical clinical spectrum generated with high Z

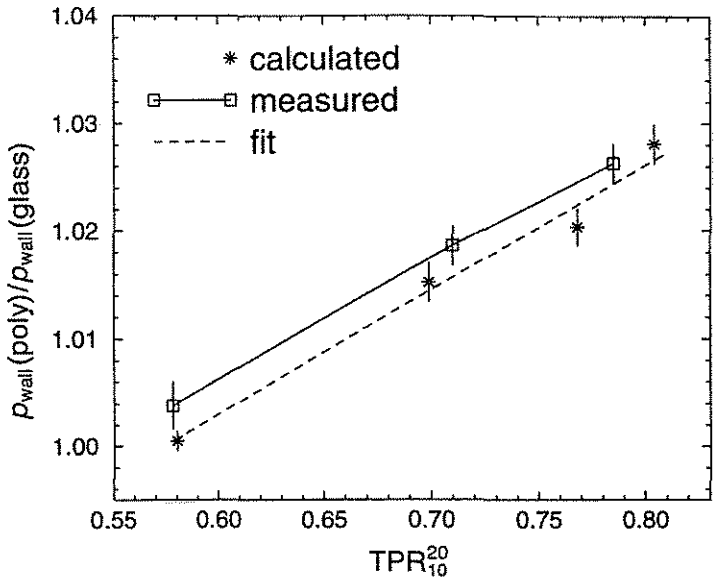

FIG. 3. Comparison of the measured and calculated ratio of the dose to Fricke solution in the standard NRC quartz walled vials to that in the polyethylene walled test detector for various photon beam qualities. The calculated value is given by the ratio of the vial wall correction for polystyrene divided by that for Pyrex walls. The dashed line is a least squares fit to the calculated data. The slightly higher measured results may be due to chemical effects from the polyethylene vial.

targets [26]. This might further affect the NRC–PTB comparison mentioned above because the PTB uses a clinical accelerator [15]. Measured data showing how beam quality affects absorbed dose calibration factors are presented in another paper at this symposium [27].

Kosunen and Rogers [26] have shown that the calculated percentage depth dose at 10 cm in a 10 × 10 cm^2 beam, %dd(10), is a better specifier of beam quality than TPR_{10}^{20} in the sense that it almost uniquely determines the water to air stopping power ratio that applies at the reference depth in any thick target bremsstrahlung beam (Fig. 4) and hence specifies the appropriate absorbed dose calibration factor which depends primarily on this stopping power ratio. A remarkable feature is that the relationship between the stopping power ratio and %dd(10) is linear and hence, unlike TPR_{10}^{20}, %dd(10) maintains its sensitivity in high energy beams. Of course, when measuring %dd(10) care must be taken to remove the effects of electron contamination from the measured value of the dose maximum. By using two very different measured sets of 'typical' clinical data which include electron contamination, a method to account for the electron contamination in clinical beams has been proposed [26]. The effects on %dd(10) are negligible below about 10 MV and increase to

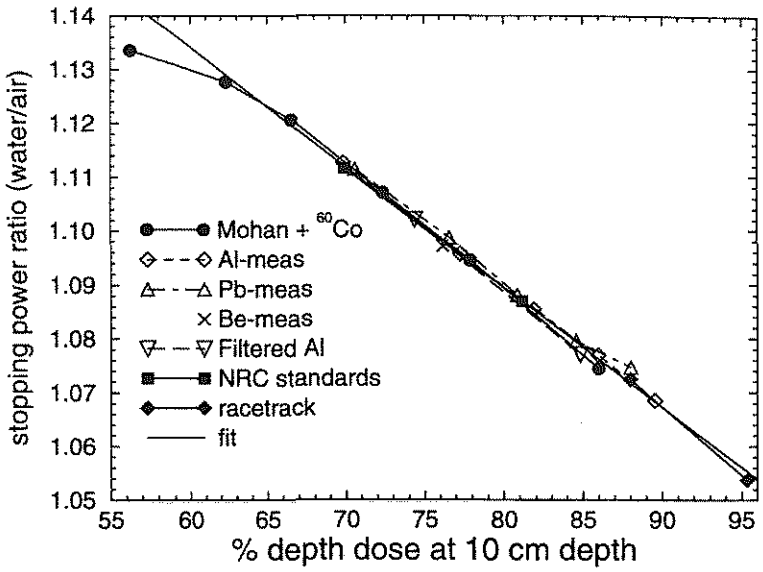

FIG. 4. Calculated Spencer–Attix water to air stopping power ratios versus %dd(10) (in a 10 × 10 cm² beam, SSD = 100 cm) for a wide variety of thick target bremsstrahlung and ^{60}Co spectra. A fit to all the bremsstrahlung beams gives

$$spr(water/air) = 1.2676 - 0.002\,224\,(\%dd(10))$$

with RMS deviation 0.0013 and maximum deviation 0.003. (Stopping powers from ICRU Report 37 [10]; figure from Ref. [26].)

about 2% in a 24 MV beam. Even a crude estimate of this contamination reduces the uncertainty in the stopping power ratio to a few tenths of a per cent. A more rigorous solution is to remove all accelerator specific electron contamination by using a scattering foil near the accelerator head and then determining the correction factor to account for the electron contamination and beam filtering effects of the foil and air [26]. The advantage of this is that these corrections are nearly the same for all accelerators of a given beam quality and need only be determined once. For a 1 mm lead foil the corrections are about 1% for a 24 MV beam [28] and allow stopping power ratios to be determined to within 0.3% for any thick target bremsstrahlung beam by measuring %dd(10) with the scattering foil in place. Routine use of the scattering foil would also improve the characteristics of clinical beams.

LaRiviere has shown that another advantage of %dd(10) is that it provides a meaningful, well defined and unique specification of nominal beam energy in megavolts — a number which manufacturers insist on using.

Another possible beam quality specifier (discussed by Ross et al. at this symposium [27]) is the dose perturbation near a high Z interface placed in the phantom in the accelerator beam of interest.

5. TRANSFER TO THE CLINIC

One significant problem with accelerator based calibrations is that they are much more expensive to provide than absorbed dose calibrations in a ^{60}Co beam (which require about the same effort as an air kerma calibration). Furthermore, many standards laboratories do not have accelerators. An approach which avoids these problems is to calibrate in a ^{60}Co beam and correct the absorbed dose to water calibration factor, N_D, to the beam quality, Q, using a factor called k_Q [1, 30], i.e.:

$$N_D^Q = k_Q N_D^{Co} \qquad \text{(Gy/C)} \tag{1}$$

and then, under reference conditions with the chamber placed with its centre at the point of measurement:

$$D_w^Q = M P_{ion} k_Q N_D^{Co} \qquad \text{(Gy)} \tag{2}$$

where: D_w^Q is the absorbed dose to water at the location of the centre of the ion chamber when the chamber is absent; the ion chamber reading M has been corrected to reference conditions of temperature and pressure; and P_{ion} corrects for lack of complete charge collection in the user's beam and must be measured for each beam quality. The value of k_Q can and will be measured for various ion chambers making use of the primary standards of absorbed dose at each beam quality. It can also be calculated, and using the AAPM TG-21 Protocol one finds:

$$k_Q = \frac{[P_{wall} P_{repl} (\bar{L}/\rho)_{air}^{w}]_Q}{[P_{wall} P_{repl} (\bar{L}/\rho)_{air}^{w}]_{Q_0}} \tag{3}$$

where P_{wall} corrects for any non-medium equivalent materials in the ion chamber, P_{repl} corrects for the cavity introduced into the medium and $(\bar{L}/\rho)_{air}^{w}$ is the Spencer–Attix water to air stopping power ratio. The measured values are to be preferred since they remove all uncertainties from our incomplete knowledge of the various quantities involved in Eq. (3) as well as any potential variations in $(W/e)_{air}$ with beam quality.

5.1. Photon beam dosimetry

Using Eq. (3), sets of k_Q values have been published as a function of TPR_{10}^{20} for all relevant ion chambers listed in the IAEA Code of Practice or AAPM TG-21 Protocol [30, 31]. The calculations were done using both the AAPM TG-21 and IAEA stopping power ratios, the former to allow the TG-21 Protocol to be applied using this formalism (TG-21 values of N_D/N_X were given), the latter because the

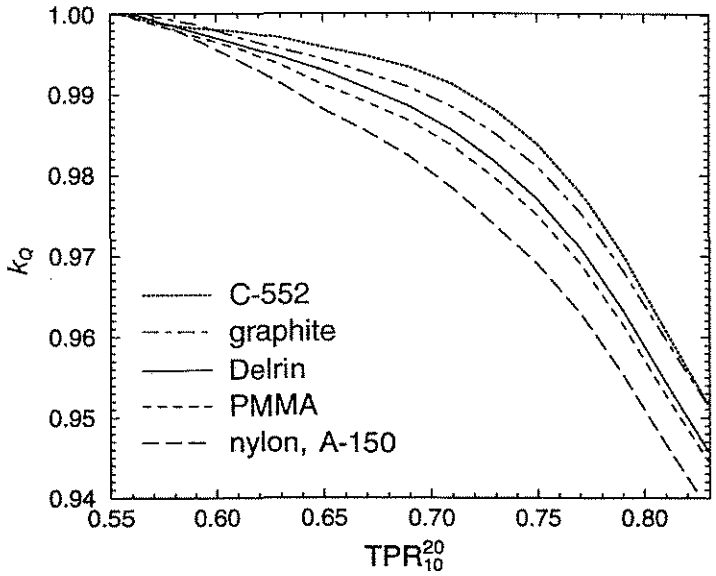

FIG. 5. Universal k_Q *curves based on ICRU 37 [10] stopping powers for all chambers of walls of the indicated materials and thicknesses less than 0.25 g/cm*^{2}*. These curves agree with the individual values for all chambers referred to in the AAPM and IAEA Protocols to within between 0.1% (for graphite, PMMA and Delrin) and 0.4%, depending on the material. (Figure from Ref. [30], which also presents analytic formulas for these curves.)*

IAEA values are in principle more accurate. The values of k_Q are by definition unity for ^{60}Co beams and decrease to about 0.96 at 24 MV. It has been shown that for chambers of a given wall material only one k_Q–TPR_{10}^{20} curve is needed for walls less than 0.25 g/cm^2 thick (Fig. 5) and these curves can be fitted with simple equations [30]. For example, for all commercial graphite walled ion chambers, the value of k_Q given by[2]

$$k_Q = 1 - 0.0877(\text{TPR} - 0.57) + 0.5279(\text{TPR} - 0.57)^2 - 3.536(\text{TPR} - 0.57)^3 \quad (4)$$

fits to within 0.12% all the individual values calculated using the AAPM value of P_{repl} and the IAEA stopping power ratios.

If k_Q values are calculated following the IAEA Code of Practice, there are some minor problems because the Code uses an effective point of measurement

[2] There is a typographical error in Ref. [30], where the rows in table III were inverted.

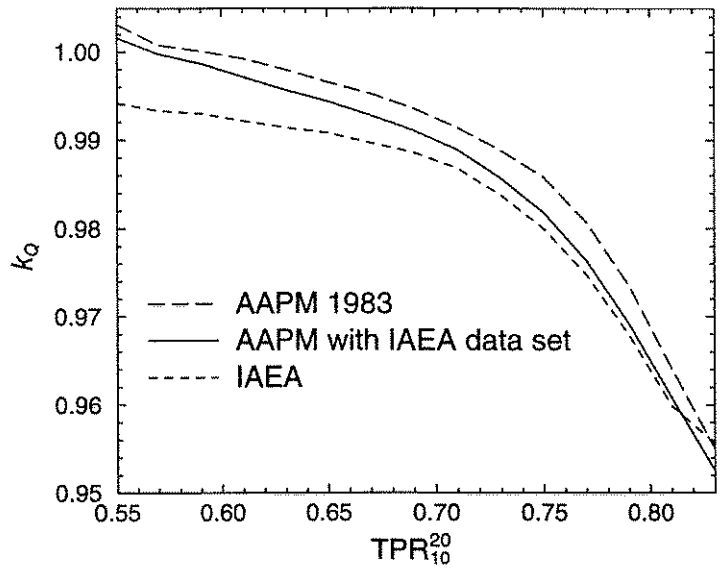

FIG. 6. Comparison of k_Q curves calculated using the original AAPM TG-21 Protocol, the TG-21 approach but with the IAEA data set, and the IAEA Code of Practice with an effective replacement correction factor which accounts for the offset in the point of measurement (does not go to unity for low energy beams because the point of measurement is defined differently for ^{60}Co beams). (From Ref. [30].)

instead of P_{repl} corrections. However, k_Q has been defined with the centre of the chamber as the point of measurement and thus one must calculate an effective P_{repl} [32]. As discussed elsewhere, the treatments of P_{repl} lead to the largest differences between the IAEA and AAPM Protocols [30], not only because of the different ways of handling the correction but also because of differences in the original data sets. However, in the present context a more significant issue is that the IAEA Code uses a different point of measurement for ^{60}Co beams than for any other beam and thus k_Q does not go smoothly to unity for low energy beams. Figure 6 compares k_Q values calculated in three ways. Except for the aluminium electrode effects at very high energies, the AAPM approach using the more recent IAEA stopping power ratios and the IAEA values appear to converge for high energy beams, but this is only because the differences in P_{repl} at ^{60}Co and in high energy beams nearly cancel.

The advantages of a system based on absorbed dose standards are as follows: (i) it is very simple to use and understand; (ii) it is in principle more accurate; and (iii) the system of primary standards upon which it is based is more robust. If a measured value of k_Q is used, then a further advantage is that the entire system is independent of a knowledge of stopping power ratios, $(W/e)_{air}$ or variations in

$(W/e)_{air}$ with beam quality, as long as the absorbed dose standard is independent of these quantities — which is usually the case.

The k_Q formalism is obviously simpler to use than the IAEA and AAPM Protocols, but this is an artificial difference because air kerma based protocols could also be written in terms of a single, ion chamber specific factor C_Q, such that:

$$D_w^Q = MP_{ion}N_K C_Q \tag{5}$$

and a complete set of C_Q values is available [31]. However, conceptually the k_Q approach is so much simpler to understand that it will be used more accurately in practice. Since many extraneous concepts and factors no longer play any role (e.g. the thickness and material of the buildup cap, the cavity length, $K_{comp}(k_m)$, $A_{wall}(k_{at})$, etc.), clinical physicists will have more time to understand and take into account more important dosimetry concepts such as the variation in chamber response away from the reference conditions being discussed here.

The second advantage of the k_Q approach, at least in photon beams, is that k_Q can be calculated more accurately than the corresponding C_Q factor because k_Q depends only on the change in various parameters, not their absolute values (e.g. we know relative stopping power ratios more accurately than absolute values). Similarly, k_Q can be measured more accurately than C_Q because we only introduce the uncertainty in the change in N_D (e.g. using the NRC standard, the uncertainty in the thermal heat defect of water cancels out when measuring k_Q). In contrast, to measure C_Q introduces the total uncertainty in both the absorbed dose and air kerma standards. This increased accuracy in measuring k_Q is useful for verification purposes but it must be remembered that when using measured k_Q or C_Q values, the uncertainty in the quantity of interest, namely D_w^Q, depends on the total uncertainty of the primary standard for absorbed dose in the beam of quality Q, and *not* on the ^{60}Co standards for either absorbed dose or air kerma.

The advantage of the robust nature of the system of absorbed dose standards has been dealt with above. Other advantages related to the reduction in uncertainty when using plastic phantoms or waterproofing sleeves are discussed elsewhere [30].

5.2. Electron beam dosimetry

The k_Q formalism makes the most sense for photon beams where Eq. (3) for k_Q represents ratios of the same quantity in different beam qualities. The same formalism can be applied in electron beams although the simplicity is lost because the ratios are of different quantities as well as qualities. Also, k_Q becomes a function of beam quality and depth in the phantom. For simplicity it may prove best to continue to use the cavity gas calibration approach of the IAEA and AAPM TG-21 Protocols but to determine N_{gas} (somewhat confusingly called N_D in the IAEA Code) from the absorbed dose to water calibration factor, N_D, i.e.:

$$N_{gas} = \frac{N_D}{[P_{wall} P_{repl} (\bar{L}/\rho)^{w}_{air}]^{60}Co} \quad \text{(Gy/C)} \tag{6}$$

This simplifies the calculation of N_{gas} considerably compared to using an air kerma calibration factor, and a complete set of calculated N_{gas}/N_D values is available [30, 31]. Although this approach means that electron and photon protocols would look different, this only reflects the reality that dosimetry for the two modalities is very different. Part of the difficulty with present protocols is that various parameters (e.g. P_{repl} and p_u) represent different quantities in electron and photon dosimetry.

6. CONCLUSIONS

Current clinical dosimetry based on air kerma standards has many problems, from intellectual obscurity, to practical complexity, to problems with the standards themselves which are subject to possible common errors. In contrast, clinical photon beam dosimetry based on absorbed dose standards is intellectually clear, simple in practice and founded on a very robust system of absorbed dose standards which are in the final stages of development. With these observations in mind, the AAPM TG-51, which is charged with producing a new or revised clinical dosimetry protocol, is investigating the feasibility of a protocol based on absorbed dose calibration factors. The German protocol is already using this approach [1].

On the basis of present understanding, changing to an absorbed dose based system will not significantly change the dose determined in careful clinical dosimetry, but because it is so much simpler to use and understand, it is bound to improve clinical practice.

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