Beam quality specification for photon beam dosimetry

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It is argued that $\%dd(10)$, the percentage depth dose at 10 cm in a 10×10 cm^2 photon beam at a SSD of 100 cm, is a better beam quality specifier for radiotherapy beams than the commonly used values of TPR_{10}^{20} or nominal accelerating potential (NAP). For radiation dosimetry purposes, TPR_{10}^{20} is not an ideal beam quality specifier because (i) stopping-power ratios for the same value of TPR_{10}^{20} can vary by up to 0.7% for thick-target bremsstrahlung beams; (ii) the value of TPR_{10}^{20} becomes insensitive to beam quality changes for high-energy beams; and (iii) it has little intuitive meaning. In contrast, $\%dd(10)$ in a pure photon beam specifies stopping-power ratios within 0.2% for all thick-target bremsstrahlung beams, maintains its sensitivity for high-energy beams, and has a simple physical and clinical meaning. It is shown that for all thick-target bremsstrahlung beams the $spr\ (water/air) = 1.2676 - 0.002\,224[\%dd(10)]$ with a rms deviation of 0.1%. The effects of electron contamination in typical high-energy clinical beams can be corrected for using previously published experimental results or by reducing electron contamination using lead scattering foils.

I. INTRODUCTION

Assessing beam quality plays a fundamental role in radiation dosimetry, but there has been no completely satisfactory single parameter that can act as a beam quality specifier. As a result, different approaches have become common for different purposes. In most radiation dosimetry protocols TPR_{10}^{20} values are used, whereas for manufacturer's specifications the nominal accelerating potential (NAP) in MV or percentage depth dose at 10 cm, $\%dd(10)$, are used.

Clinical dosimetry protocols usually use the value of TPR_{10}^{20}. This value is determined by measuring the absorbed dose on the beam axis at depths of 20 and 10 cm for a constant source-detector distance and a 10 cm$\times$10 cm field at the plane of the chamber. Strictly speaking, the definition is in terms of absorbed dose at two depths, but the ratio of ionization measurements in the two geometries gives an acceptably accurate value because stopping-power ratios and other factors required to convert ionization to dose do not change much with depth.

The idea of using measured ratios of ionization or dose at two depths was first introduced in the Nordic dosimetry protocol (NACP) to specify the accelerator energy.[1] The AAPM protocol[2] went one step further, based on the work of Cunningham and Schulz[3] and directly associated the beam quality index with the stopping-power ratios needed in the dose equation of the protocol. They found that there is a universal curve relating these two quantities based on analytic calculations of stopping-power ratios and TPR_{10}^{20} for a variety of clinical photon spectra. They needed to relate their calculated TPR_{10}^{20} with the experimental values in order to produce a universal curve relating experimental TPR_{10}^{20} values and stopping-power ratios because their calculations only included first-order scattering. Andreo and Brahme showed that for clinical photon spectra one could accurately calculate values of TPR_{10}^{20} using Monte Carlo techniques, and found that they obtained a universal curve relating stopping-power ratios and TPR_{10}^{20}.[4] They also showed that for a given nominal accelerator energy there are up to 1.3% variations in the stopping-power ratio. It is for this reason that TPR_{10}^{20} is used rather than nominal accelerating energies. However, this is not the final answer, since Andreo has shown that stopping-power ratios vary by up to 1.5% for bremsstrahlung beams with the same value of TPR_{10}^{20}.[5] This variation applies to extreme spectra and is much greater than the variation found in clinic-like beams. However, the National Physical Laboratory has reported different measured absorbed-dose calibration factors for ion chambers in different bremsstrahlung beams with the same value of TPR_{10}^{20} but generated with and without a thick aluminum filter in place behind a high-Z target.[6] The variation of up to 0.5% is related to the variation in stopping-power ratios for a given value of TPR_{10}^{20}. This is a significant problem for standards laboratories offering absorbed dose calibration services for accelerator beams and potentially represents the largest source of uncertainty in the calibration factor.

High-energy photon beams also lead to other problems with TPR_{10}^{20} as a beam specifier. As will be shown below, the high-energy swept beams from the new racetrack microtron accelerators have considerably different curves of stopping-power ratios versus TPR_{10}^{20} values compared to other "typical clinical beams." Furthermore, for high-energy beams, TPR_{10}^{20} is an insensitive quality specifier. For example, a 1% change in TPR_{10}^{20} for values near 0.8 leads to a 3-MV change in the nominal accelerating potential (near 20 MV) and a 0.4% change in water to air stopping-power ratio. In contrast, for values of TPR_{10}^{20} near 0.7 a 1% change corresponds to a 0.1% change in stopping-power ratio and only an 0.5-MV change in the NAP.

Manufacturers and others have often specified beam quality in terms of a nominal MV which is not well defined. The obvious definition in terms of the energy of the electrons from the accelerator is not very useful because this energy is usually not well known. More importantly, the beam quality is strongly affected by the type of beam flat-

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tening used (see LaRiviere[7] and references therein for a good discussion). LaRiviere proposed that the beam quality in MV should be specified in terms of $\%dd(10)$, the percentage depth dose at 10-cm depth in a 10×10-cm^2 field at a source to surface distance (SSD) of 100 cm with

$$Q=10^{[\%\mathrm{dd}(10)-46.78]/26.09}\ (\mathrm{MV}), \tag{1}$$

which is a good fit to the "experimental" data, although there is scatter in the initial data caused by the lack of a clear definition of Q. He also showed that a similar relationship between Q and the depth of maximum dose, $d_{\max}$, could be derived, but he argued that this is not as good as using $\%dd(10)$ values because the depth of dose maximum, as opposed to the dose maximum itself, is harder to measure and more subject to electron contamination problems. LaRiviere concluded that beam quality in MV should be specified using Eq. (1), but in practice his method amounts to using $\%dd(10)$ as the beam quality specifier.

However, LaRiviere took one more important step. Using a combination of calculations based on published spectra and experimental depth-dose curves, he showed an excellent correspondence between $\%dd(10)$ and the dose-weighted mean energy of the photon beams $\bar{E}_d$ (MeV), viz.,

$$\bar{E}_d=10^{[\%\mathrm{dd}(10)-55.37]/28.68}\ (\mathrm{MeV}). \tag{2}$$

Having discovered this strong correlation, he did not pursue it, but noted that $\bar{E}_d$ might correlate well with the value of TPR_{10}^{20}. Following this paper, Owen at the NPL showed that if measured absorbed-dose calibration factors are plotted as a function of a calculated value of $\bar{E}_d$, a universal curve is obtained instead of different curves of calibration factors which depended on how the beam is filtered and are not uniquely specified as a function of TPR_{10}^{20}.[6]

Following these leads, we have done an extensive series of Monte Carlo calculations for many different clinical and other realistic bremsstrahlung spectra. We have confirmed the variation in stopping-power ratio for a given TPR_{10}^{20} value as reported by Andreo.[5] We have found that $\%dd(10)$ does not correlate particularly well with the value of TPR_{10}^{20} as hoped for by LaRiviere, but we do find a very close correlation between $\bar{E}_d$ and stopping-power ratios, as suggested by the NPL work. Most importantly, for all thick-target bremsstrahlung beams we find a very tight linear relationship between calculated values of $\%dd(10)$ and stopping-power ratios. We show that our calculations are in good agreement with a large amount of experimental data, except for high-energy beams where contaminant electrons play a role. Suggestions on how to deal with this problem are made. The conclusion of this work is that manufacturers, clinicians, and those doing radiation dosimetry could all make direct use of $\%dd(10)$ as a beam quality specifier. This avoids many of the current problems while making use of a well-defined and routinely measured quantity.

II. CALCULATIONS

The original intention was to follow the technique of Andreo and Nahum[8] and use precomputed data for monoenergetic photon beams and fold these data with spectra from a variety of accelerator beams to produce the parameters of interest. To this end a code called DDSPR was written and documented.[9] However, it was found that several of the parameters of interest associated with the maxima of depth-dose curves are not derived with sufficient accuracy using this technique. This is because of the various interpolations required and the lack of spatial resolution about the regions of maximum dose in the databases employed. To overcome these shortcomings, it was found easiest to do the complete Monte Carlo calculation for each of the spectra involved.

The calculations are done using the EGS4 Monte Carlo code system for simulating electron and photon transport.[10] Central-axis depth-dose curves are calculated as a function of incident beam radius using the user code DOSRZ. The incident photon beams are circular, parallel, and incident normally. This code has been extensively benchmarked.[11] Calculations are done for a sufficient number of histories to reduce the statistical uncertainty on the dose in individual depth bins to a few tenths of a percent or less (up to 60×10^6 histories). The default PRESTA electron transport algorithm is used.[12] Electron histories are tracked down to a kinetic energy of 500 keV at which point their residual range is 1.8 mm. The values of TPR_{10}^{20} are calculated for 100-cm^2 parallel photon beams by least squares fitting the calculated depth-dose curve to an exponential between depths of 10 and 20 cm. To calculate the dose maximum, percentage depth dose at 10 cm and the depth at which the dose fell to 80% of $D_{\max}$, the parallel beam results are transformed to point source results at a SSD of 100 cm by using a simple $1/r^2$ correction, assuming a point source. The depth bins are adjusted for each spectrum, so that the maximum of the depth-dose curve has at least three bins of 3.3 mm thickness. In the selection of the maximum value of the dose, a distinct bias would be introduced by selecting the peak value on the depth-dose curve. Because of the statistical fluctuations one would usually pick a slightly too high value. To avoid this, the number of bins near $D_{\max}$ is kept at three or more and the depth-dose curve is fit using a least squares fitting routine which selected the best analytic representation of the curve from before the dose maximum to about 20-cm depth. In all cases excellent fits are obtained and $D_{\max}$ is taken as the maximum of the fitted function and the uncertainty is obtained from the statistical uncertainty on the fitted values. The residuals are comparable to the fitting uncertainty, which is typically 0.2%. Once the value of $D_{\max}$ is determined it is straightforward to obtain the dose at a depth of 10 cm as a percentage of $D_{\max}$—again the fitted dose at 10 cm is used to reduce the effects of statistical fluctuations in a single bin. The statistical uncertainty on the calculated value of $\%dd(10)$ is 0.2% or less. The analytic fit is also used to determine depth at which the dose fell to 80% of $D_{\max}$.

The Spencer–Attix stopping-power ratios are deter-

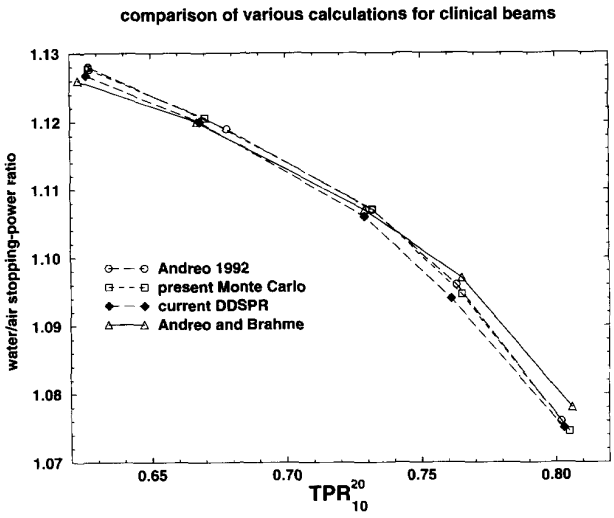

FIG. 1. Values of Spencer–Attix water to air stopping-power ratios versus TPR^{20}_{10}, calculated here using the full Monte Carlo calculations or the DDSPR code, and as calculated by Andreo and Brahme (Ref. 4) and Andreo (Ref. 15). The spectra used are those calculated by Mohan *et al.* (Ref. 18).

mined using an EGS4 user code called SPRRZ(V5).[13] This code also uses the PRESTA algorithm and calculates stopping-power ratios "on the fly". This code has been checked in detail against the more standard techniques used by Malamut *et al.* with the EGS4 code[14] and gives the same results to within 0.1%. It has also been checked in detail against the results for monoenergetic beams of Andreo[15] and found to agree within about 0.1%, except above 20 MeV where the present results are up to 0.3% lower (which is consistent with previous comparisons[14]). The electron collision stopping powers of ICRU Report No. 37[16] are used in the EGS4 code, as implemented by Duane *et al.*[17] For all Spencer–Attix stopping-power ratios, a value of $\Delta = 10$ keV is used. The stopping-power ratios are calculated for parallel beams of various radii incident on a water phantom. The results presented here are for 100-cm^2 beams and are at a depth of 10 cm, although variations in stopping-power ratios between 5 and 10 cm are less than 0.2% except for beams above 25 MV. The statistical uncertainty on the stopping-power ratios is 0.1% or less in all cases.

The input spectra used for the calculations come from a variety of places, which are described in detail in the Appendix. In many cases the complete accuracy of these spectra is not critical to the arguments here because, in the end, a beam quality specifier will be proposed which is close to unique. Thus, even if the actual beam from a particular accelerator differs from what is used here, one can use the measured beam quality specifier to determine the stopping-power ratio for that beam.

Figure 1 compares the present results for the calculated stopping-power ratios as a function of TPR^{20}_{10} using both the full Monte Carlo calculations and the results from DDSPR for the clinical spectra calculated by Mohan *et al.*[18] to the latest results of Andreo[15] (which supersede the results of Andreo and Brahme[4]). The comparison shows exceptionally good agreement for the latest calculations (at the 0.1% level) and reasonable agreement with the previous values or those done with the DDSPR code. These differences are much less than the systematic uncertainties in the calculations involved.

The dose weighted mean energy is calculated by the code DDSPR[9] using the equation

$$\bar{E}_d(z) = \frac{\Sigma_{j=1}^{n} D(z,E_j) E_j^{\text{mean}}}{D(z)}, \tag{3}$$

with

$$D(z,E_j) = \int_{E_{j,\min}}^{E_{j,\max}} \phi(E) K(z,E)\, dE, \tag{4}$$

where $\phi(E)$ is the incident photon fluence spectrum, $K(z,E)$ is the precomputed photon fluence to dose (at depth z) conversion factor for photon energy E, and $D(z,E_j)$ is the dose delivered by photons in energy bin E_j with mean energy E_j^{mean}. Note that the dose weighted mean energy refers only to the incident spectrum, is dependent on the depth referred to and tends to increase as the depth increases because higher-energy photons contribute relatively more dose at greater depths. In this work we consistently use $z = 10$ cm, although LaRiviere did not explicitly state what depth he was using in his calculations.[7]

III. RESULTS

A. Stopping-power ratios versus TPR^{20}_{10}

Figure 2 presents the calculated stopping-power ratios versus TPR^{20}_{10} for a variety of bremsstrahlung beams plus a series of monoenergetic beams. This figure confirms Andreo's results[5] which showed that TPR^{20}_{10} is not a unique specifier of beam quality for determining stopping-power ratios. Under extreme conditions there is a spread of over 2% in the stopping-power ratio for a given TPR^{20}_{10} value—but that is for a monoenergetic beam compared to a Schiff thin-target bremsstrahlung spectrum. A more relevant comparison is between the curve for clinical spectra represented by the curve for the spectra of Mohan *et al.*,[18] and the values for the spectra in the calibration beams at NRC. Here the difference is up to 0.7%. Thus, if an ion chamber were calibrated in the NRC 20-MV beam, the calibration factor would be roughly 0.7% low compared to an ion chamber calibrated in a "clinical" beam with the same TPR^{20}_{10} value. This situation occurred in a recent comparison of the ion chamber calibration factors determined using the German and Canadian primary standards of absorbed dose in a 20-MV photon beam. The German laboratory uses a clinical accelerator and to the extent that its beam matches those of Mohan *et al.*, it could be expected that the NRC ion chamber calibration factor would be about 0.7% lower for beams with the same value of TPR^{20}_{10} (assuming, of course, that the primary standards are independent of these effects, as is normally the case).[19]

Another important feature of Fig. 2 is that the two points for the racetrack microtron accelerator beams are

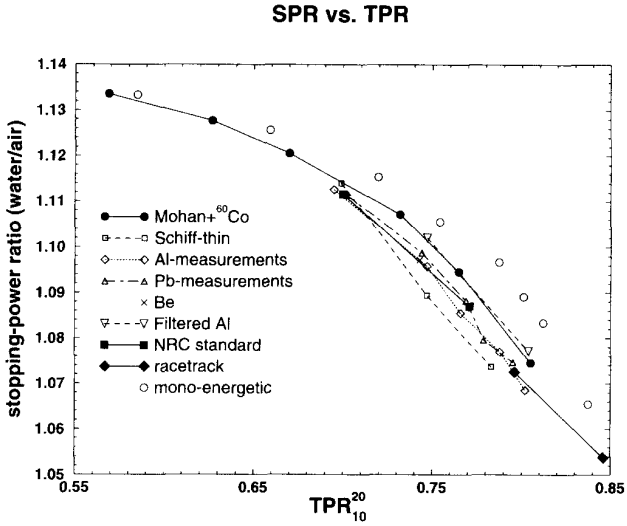

FIG. 2. Values of stopping-power ratios versus TPR^{20}_{10} for nine families of photon spectra. Monoenergetic beams with energies of 1.25, 2, 3, 4, 5, 6, 7, and 10 MeV (○); the Mohan *et al.* spectra (Ref. 18) plus a ^{60}Co spectrum (Ref. 20) (●); 30 and 50 MV calculated spectra for the racetrack microtron (closed ◆); measured spectra from thick targets (110% of CSDA range) of Al (◇), Pb (△), and Be (×) (Refs. 22 and 23); aluminum thick-target spectra filtered by an additional 14 cm of Al (▽); calculated thin-target Schiff spectra for Al (□); and spectra in NRC calibration beams at 10 and 20 MV (■). All the beams for flattened, i.e. practical beams, are shown as closed symbols.

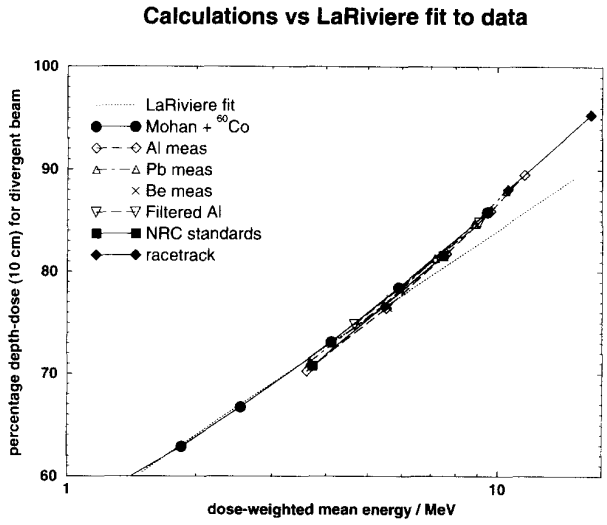

FIG. 3. Values of $\%dd(10)$ versus calculated dose-weighted mean energies of the incident beams. The symbols show the current results for all the thick-target bremsstrahlung and ^{60}Co spectra described in the caption of Fig. 2. The dotted straight line is LaRiviere's fit to his measured $\%dd(10)$ values (Ref. 7).

not on the standard clinical curve. This implies values in the AAPM TG-21 or IAEA dosimetry protocols will not apply to those accelerators since the protocol values are based on the "typical" clinical spectra represented here by the Mohan *et al.* spectra.

A final point worth noting is that the values for the aluminium thick-target bremsstrahlung spectra filtered by 14 cm of aluminum are very close to the "clinical spectra." This confirms the NPL's approach in their calibration facility of using a thick aluminum filter to make their thick-target spectrum more clinic-like.

B. Comparison to LaRiviere

LaRiviere expressed some hope that $\%dd(10)$ would correlate well with TPR^{20}_{10} so that the value of $\%dd(10)$ would be useful for dosimetry purposes. We have done such a comparison and find scatter in the data that is comparable to that found in Fig. 2 for stopping-power ratios versus TPR^{20}_{10}. The reasons for this will be clear below.

We have also compared our calculated dose-weighted mean energies versus $\%dd(10)$ curves to LaRiviere's calculated mean energy versus measured $\%dd(10)$ curves. He found a close linear relationship between $\%dd(10)$ and $\log \bar{E}_d$, which is shown as a dotted straight line in Fig. 3. We also find a very tight relationship between these quantities for all of our thick-target bremsstrahlung spectra. Our values are within 1% of LaRiviere's linear fit for beams with $\%dd(10)$ values less than 78%. For higher-energy beams our results have increasingly higher values of $\%dd(10)$ for a given $\bar{E}_d$, about 2.5% of D_{max} for the 24-MV spectrum of Mohan *et al.* As will be discussed below, this is likely due to electron contamination in the experimental beams. This would increase the dose maximum and thereby decrease the measured value of $\%dd(10)$. On the other hand, there could also be a systematic problem at higher energies for LaRiviere's rather crude evaluation of $\bar{E}_d$ (for which he specified no depth dependence).

Figure 4 shows the calculated values of stopping-power ratios versus $\bar{E}_d$. Rather than the scatter in the stopping-power ratios seen in Fig. 2 for a given TPR^{20}_{10} value, here we see virtually no scatter amongst any of the curves for a

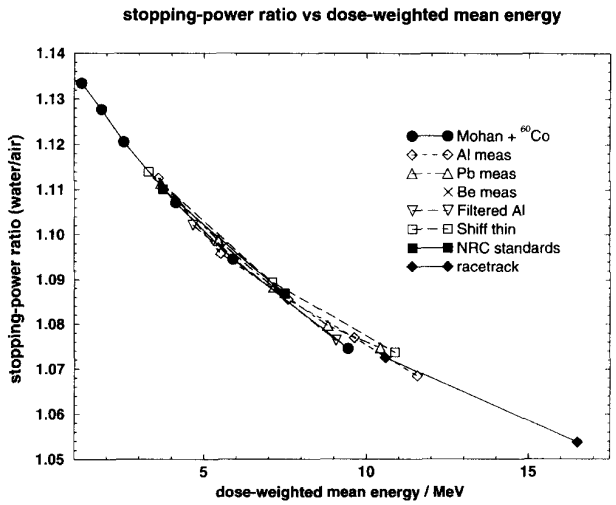

FIG. 4. Calculated Spencer–Attix water to air stopping-power ratios versus dose-weighted mean energies of incident photon spectra for all the bremsstrahlung spectra presented in Fig. 2.

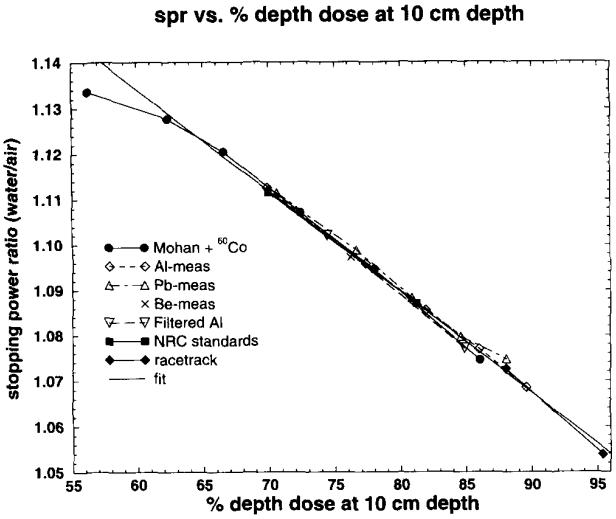

FIG. 5. Calculated Spencer–Attix water to air stopping-power ratios (based on ICRU 37 stopping powers) versus percentage depth dose at 10 cm for the thick-target bremsstrahlung and ^{60}Co spectra described in Fig. 2. The straight line shown is the linear fit to all the bremsstrahlung beams given by Eq. (5) with a rms deviation of 0.0013 and a maximum deviation of 0.003.

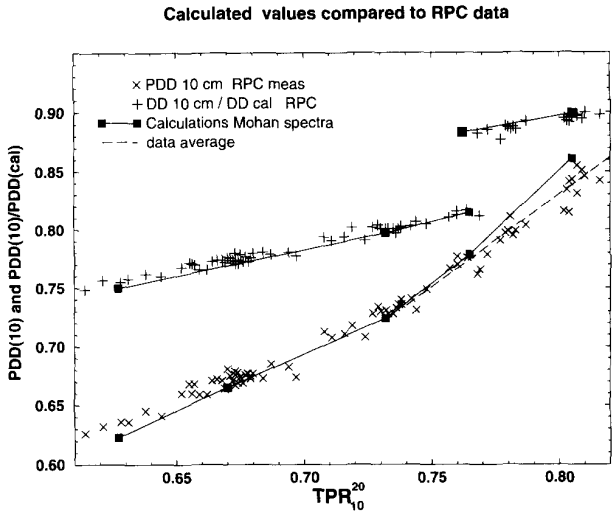

FIG. 6. Comparison as a function of TPR^{20}_{10} of calculated and measured values of $\%dd(10)$ (given as a fraction PPD), and the dose at 10-cm depth divided by the dose at the calibration depth that changes from 5 to 7 cm at a value of TPR^{20}_{10} of ≈ 0.76. The experimental data are a representative sample from the database of the Radiological Physics Center from their site visits at many different clinics. The long-dashed line is the estimate of the mean value of the measured values which is used in Sec. III D to estimate electron contamination effects.

thick-target bremsstrahlung spectrum, no matter how it is created or filtered. This is an interesting result that is consistent with Owen's result at the NPL, which showed that absorbed-dose calibration factors fell on a universal curve when plotted against a calculated $\bar{E}_d$.[6] Since LaRiviere found a unique relationship between measured values of $\%dd(10)$ and $\bar{E}_d$ (see Fig. 3), one can immediately see that measured $\%dd(10)$ can be used to determine $\bar{E}_d$ uniquely which, in turn, can be used to specify stopping-power ratios uniquely. However, it can be even easier.

C. Stopping-power ratios versus %dd(10)

Figure 5 presents the central result of this paper, showing that for all the thick-target bremsstrahlung spectra considered here, there is a very well-defined linear relationship between the calculated $\%dd(10)$ values and the water to air stopping-power ratios for these beams. We find

$$\left(\frac{\bar{L}}{\rho}\right)^{\text{water}}_{\text{air}} = 1.2676 - 0.002\,224[\%dd(10)], \qquad (5)$$

fits all 22 data points with a rms deviation of 0.0012 and a maximum deviation of 0.0028 (and this is for an unfiltered 30-MV thick target of Pb). This unique relationship must be compared to the obvious spread in the families of curves in Fig. 2 (excluding the non-thick-target spectra, which are not included in Fig. 5).

This figure suggests completely bypassing the concept of $\bar{E}_d$ and using $\%dd(10)$ to determine stopping-power ratios.

D. Electron contamination

The above suggestion ignores the discrepancies in Fig. 3 between the calculated $\%dd(10)$ versus $\bar{E}_d$ curves and the straight line fit to the measured $\%dd(10)$ data. To investigate this discrepancy further we have compared our calculated values to a large selection of measured data extracted from the Radiological Physics Center's database. Figure 6 shows two sets of RPC data as a function of TPR^{20}_{10}. The first is $\%dd(10)$ (actually fractional depth dose, PDD, rather than %) and the second is the ratio of dose at 10 cm to that at the calibration depth (5 cm for $TPR^{20}_{10} < 0.76$, 7 cm for higher energy beams). Also shown are our calculated values for "clinical" spectra (the Mohan *et al.* series). In this comparison we assume that the calculated values of TPR^{20}_{10} are accurate based on previous studies that show Monte Carlo calculations can accurately calculate depth-dose curves past the region of electron contamination.[18,4] On this same basis one would expect the Monte Carlo calculations to be accurate when calculating the ratio of doses at 10 cm and the calibration depth. This is confirmed in the upper curves in Fig. 6. Note that the curve is in two parts because the calibration depth changes from 5 to 7 cm for a value of TPR^{20}_{10} of about 0.76. The calculations are in good agreement with experiment, and the experimental results are tightly grouped. The lower part of the figure shows the $\%dd(10)$ versus TPR^{20}_{10} values. Here the experimental data exhibit much more scatter, presumably because the dose maximum must be measured to determine $\%dd(10)$ whereas it cancels out in the other dose ratios. Even with the more highly scattered experimental data it is clear that the calculated values of $\%dd(10)$ are 1% or 2% high for large values of TPR^{20}_{10}

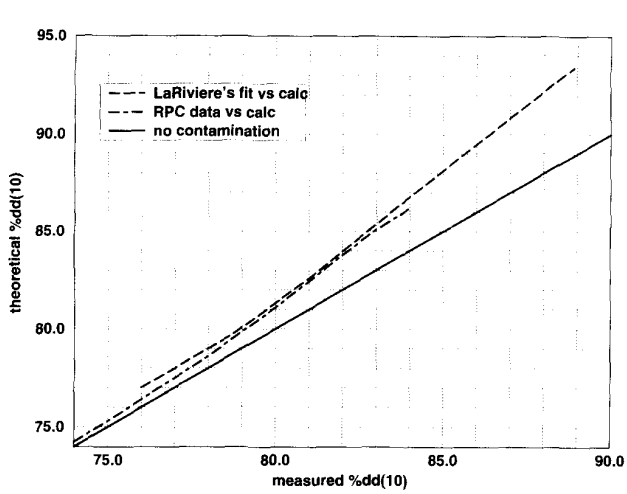

FIG. 7. Two estimates of the relationship between measured values of the percentage depth dose at 10 cm, $\%dd(10)$ and those calculated here. Those measured include electron contamination at the maximum dose, whereas the calculations are for pure photon beams. The curves are based on the differences between the measured and calculated values of $\%dd(10)$ in Figs. 3 and 6 and on the assumption that the TPR_{10}^{20} and $\bar{E}_d$ values are correct. The underlying data show considerable fluctuation and these corrections apply only to typical clinical accelerators.

and possibly 1% low for low values of TPR_{10}^{20}. Calculations with the other beams used in this study would predict even higher values of $\%dd(10)$ for a given TPR_{10}^{20} and thus cannot explain the discrepancies at high energies.

However these results are consistent with the results in Fig. 3, where it is also found that the calculated $\%dd(10)$ values are too high compared to measurements for high-energy beams. As discussed above, this is likely due to electron contamination affecting the measured dose maximum. In Fig. 6 there is no uncertainty in the TPR_{10}^{20} part of the data, and thus the differences can be associated with the calculated $\%dd(10)$ values, unlike the case in Fig. 3, where there is some uncertainty in LaRiviere's values of $\bar{E}_d$.

If it is assumed that the measured and/or calculated values of TPR_{10}^{20} and $\bar{E}_d$ are correct in Figs. 3 and 6, then it is possible to determine the relationship between the calculated and measured $\%dd(10)$. If there were perfect measurements and no electron contamination, the measured and calculated values would be the same, as represented by the solid line in Fig. 7. However, electron contamination implies the measured values of $\%dd(10)$ are less than the calculated values. Figure 7 shows that the data in Figs. 3 and 6 imply the same electron contamination effects on average. This gives some confidence that this is a reasonable approach, but note there is considerable fluctuation in the experimental data. This may represent real machine to machine variations and/or measurements uncertainties. Also note that although there appears to be no correction needed below $\%dd(10)$ values of 74% [11 MV from Eq. (1)], this only means that the electron contamination does not affect the value of the dose maximum in a 10×10-cm^2 beam.

Although electron contamination plays a role in defining $\%dd(10)$, even if we completely ignore a 2% electron contamination, the error introduced in the stopping-power ratio is about 0.4%. However, we can do better than that. Making use of the data presented in Fig. 7, the measured $\%dd(10)$ can be corrected for electron contamination with sufficient accuracy to reduce the uncertainty in the stopping-power ratio to 0.2% or less. Another approach is to utilize thin high-Z scattering foils, which can substantially reduce the electron contamination without significantly affecting the beam quality.[20,21] For example, we find that although lead filters of up to 1 mm thickness reduce the beam intensity by up to 7%, they have no practical effect ($<0.03\%$) on the calculated values of TPR_{10}^{20}, stopping-power ratios, or $\%dd(10)$ in a 24-MV, 100-cm^2 clinical photon beam. At the same time these filters completely eliminate electrons originating in the accelerator head, although there is still some electron contamination from photon interactions in the air and the filter itself. This needs to be investigated further for high-energy beams, but it seems very likely that procedures can be developed that remove electron contamination from the dose maximum, at least while determining beam quality for small area beams.

IV. OTHER ISSUES

Measuring $\%dd(10)$ requires measuring the dose maximum and the dose at 10 cm depth. The replacement correction factors (P_{repl}) will differ in these locations, and thus the photon depth-dose curve is not just given by the depth-ionization curve. At dose maximum the replacement correction should be unity because in photon beams it is just a gradient correction factor. However, past the maximum dose it becomes 0.992 for a farmer-like chamber in a low-energy beam and closer to unity for high-energy beams. If this change is ignored when measuring $\%dd(10)$, then the measured $\%dd(10)$ will be high by 0.7% for low-energy beams and somewhat less so for higher energy beams. There is also a small variation in the stopping-power ratio with depth (0.2% or less[9]) which would have a small effect on measurements of $\%dd(10)$.

Another alternative to $\%dd(10)$ is the depth at which the dose falls to 80% of dose maximum. Figure 8 shows that this parameter also is a good specifier of stopping-power ratios. Although an alternative to $\%dd(10)$, this quantity suffers from the same problems concerning electron contamination.

V. CONCLUSIONS

With care, it appears possible to use $\%dd(10)$ as an accurate indicator of beam quality. LaRiviere has shown it is a good indicator of the nominal accelerating potential in MV and the present results indicate it is a better indicator of stopping-power ratios than the value of TPR_{10}^{20}. The value of $\%dd(10)$ maintains better sensitivity to beam quality changes for high-energy beams than does the value

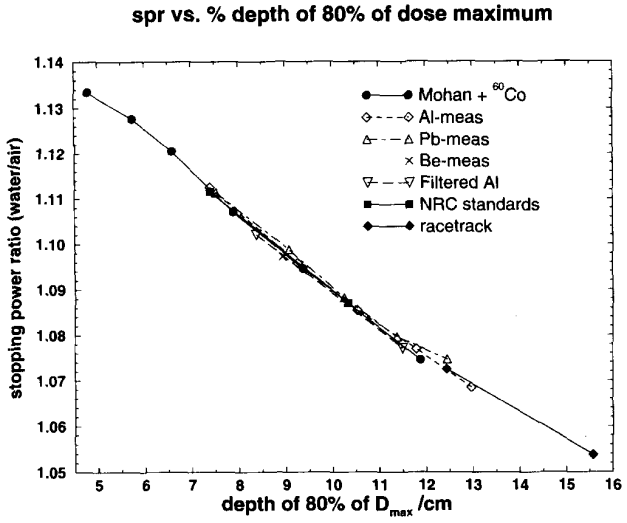

FIG. 8. Spencer–Attix water to air stopping-power ratios versus d_{80}, the depth at which the dose falls to 80% of the maximum dose (in cm), for the thick-target bremsstrahlung and ^{60}Co spectra described in Fig. 2. The value of d_{80} specifies the stopping-power ratios uniquely within a few tenths of a percent.

of TPR_{10}^{20}. The final, perhaps greatest, advantage of $\%dd(10)$ is that the parameter $\%dd(10)$ is a physically meaningful and clinically useful parameter, whereas TPR_{10}^{20} values provide no intuitive insight into the quality of the beam being considered. The only drawback with $\%dd(10)$ is the problem of electron contamination of the beam. Global corrections can be made based on the experimental data presented here, and a proposal for a more definitive approach that reduces electron contamination using scattering foils is being investigated. It would be more satisfying to have a good understanding of why this unique and linear relationship occurs between $\%dd(10)$ and stopping-power ratios, however, such an understanding awaits further insight. Whatever the explanation, it is not very simple since it must take into account that the variation in the stopping-power ratios is primarily related to the difference in the density effect on the stopping powers of water and air, whereas the parameter $\%dd(10)$ is related to photon interactions in water.

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APPENDIX: INCIDENT PHOTON SPECTRA

The calculations require knowledge of families of incident photon spectra which are obtained from a variety of experimental and calculated sources.

For the ^{60}Co beam a calculated spectrum is used which included a 30% fluence contribution from photons scattered from the source capsule and collimators.[20] For typical clinical accelerator spectra, the calculated spectra of Mohan *et al.* are used.[18]

Calculations are also done for two photon beams at 10 and 20 MV which are used for standards work at NRC. They are generated by a thick target of aluminum and flattened by two different conical aluminum flattening filters. Another two beams are 30- and 50-MV beams from the Scanditronix 50-MV racetrack microtron in which bremsstrahlung beams from a thick tungsten/copper target are swept across the field to achieve flatness. For these last four spectra, preliminary EGS4-calculated on-axis photon spectra are obtained from the NRC user code ACCEL which models cylindrically symmetric accelerator heads (this is a minor modification of the NRC user code FLURZ described in detail in Ref. 14). The flattening filters are modeled as a stacked set of cylinders. The details of these calculations are not critical, and the spectra are only used as representative of different spectra.

The calculations also make use of a series of measured spectra generated at 0° by 10-, 15-, 20-, 25-, and 30-MeV beams of electrons incident on thick targets (110% of the CSDA range of the incident electrons) of aluminum, lead, and berylium.[22,23] These measured spectra have been shown to be in good agreement with Monte Carlo calculated spectra for the same configurations.

Another pair of spectra is obtained by starting from the measured 10- and 20-MV thick target aluminum spectra and analytically filtering these by an additional 14 cm of aluminum. This extra filter is thought to have some of the characteristics of flattening filters in clinical accelerators, and also bears some relationship to the filtering used at the NPL.[6]

The final family of spectra are calculated using the standard Schiff 0° formula for 10-, 20-, and 30-MeV electrons incident on a thin target of aluminum.[24]

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