Variations in photon energy spectra of a 6 MV beam and their impact on TLD response

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Purpose: Measurement of the absorbed dose from radiotherapy beams is an essential component of providing safe and reproducible treatment. For an energy-dependent dosimeter such as thermoluminescent dosimeters (TLDs), it is generally assumed that the energy spectrum is constant throughout the treatment field and is unperturbed by field size, depth, field modulation, or heterogeneities. However, this does not reflect reality and introduces error into clinical dose measurements. The purpose of this study was to evaluate the variability in the energy spectrum of a Varian 6 MV beam and to evaluate the impact of these variations in photon energy spectra on the response of a common energy-dependent dosimeter, TLD.

Methods: Using Monte Carlo methods, we calculated variations in the photon energy spectra of a 6 MV beam as a result of variations of treatment parameters, including field size, measurement location, the presence of heterogeneities, and field modulation. The impact of these spectral variations on the response of the TLD is largely based on increased photoelectric effect in the dosimeter, and this impact was calculated using Burlin cavity theory. Measurements of the energy response were also made to determine the additional energy response due to all intrinsic and secondary effects.

Results: For most in-field measurements, regardless of treatment parameter, the dosimeter response was not significantly affected by the spectral variations ($<1\%$ effect). For measurement points outside of the treatment field, where the spectrum is softer, the TLD over-responded by up to 12% due to an increased probability of photoelectric effect in the TLD material as well as inherent ionization density effects that play a role at low photon energies.

Conclusions: It is generally acceptable to ignore the impact of variations in the photon spectrum on the measured dose for locations within the treatment field. However, outside the treatment field, the spectra are much softer, and a correction factor is generally appropriate. The results of this work have determined values for this factor, which range from 0.88 to 0.99 depending on the specific irradiation conditions. © 2011 American Association of Physicists in Medicine.

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I. INTRODUCTION

Recently published data have demonstrated variations in the photon energy spectra of therapeutic radiotherapy beams. These variations are especially pronounced for small field sizes, such as those used in intensity-modulated radiation therapy (IMRT),^{1,2} but variations also exist as a function of field size, depth, and distance away from central axis for large symmetric fields.^{1,3} The presence of heterogeneities, such as lung or bone tissue, may also perturb the photon energy spectra compared to homogeneous tissues or soft tissue. Outside of the treatment field the photon spectra is likely to be very different from the spectrum on the central axis. Out-of-field radiation is known to be composed of leakage radiation from the head of the treatment machine and scatter produced from the collimator system or inside the phantom or patient.⁴ While head leakage has a comparable spectrum to the primary beam, scattered radiation has a much lower (softer) energy spectrum. These variations in the energy spectrum of a therapeutic radiotherapy beam will

impact the response of energy-dependent dosimeters and could thereby result in an incorrect measure of dose unless these variations are accounted for.

Thermoluminescent dosimeters (TLDs) are one such type of energy-dependent dosimeter, and they are widely used for dose measurement because of their availability, small size, and accuracy. TLDs use the principles of thermoluminescence to estimate the absorbed dose to the reference medium at the location of the TLD. While many materials exhibit thermoluminescent properties, the most widely used in medical applications is lithium fluoride (LiF). LiF has an effective atomic number similar to that of soft tissue (8.3 versus 7.2), indicating that there is a close correlation between the energy deposited in LiF and that deposited in soft tissue.⁵⁻⁷ To determine absorbed dose (D), the raw TLD signal (T), is multiplied by the system sensitivity factor (S). A series of correction factors must also be applied to account for the specific application of the TLD Eq. (1); these include linearity (K_L), fading (K_F), and energy response (K_E), which are generally determined during commissioning of the TLD.⁸

$$D = T \times S \times K_L \times K_F \times K_E \quad (1)$$

In this work we investigated the final correction factor: energy response. The standard energy-response factor (K_E) is required to account for differences in the TLD response to the experimental beam energy (6 MV, for example) relative to the reference beam energy (e.g., ^{60}Co). This standard energy correction factor is commonly established by measuring the response of the TLD to a known dose of 6 MV photons (determined on the central axis at a depth of maximum dose for a 10 cm $\times$ 10 cm field) versus the response of the TLD to a known dose of ^{60}Co photons (determined under the same reference conditions). The basic definition of the energy-response factor is given in Eq. (2).⁸

$$K_E = \frac{D^{6\text{MV}} T^{\text{Co}-60}}{D^{\text{Co}-60} T^{6\text{MV}}} \quad (2)$$

In practice, this single energy response/characterization is used for any measurement conducted at a nominal beam energy (e.g., 6 MV). Either a single energy response is used, or if both experimental and reference TLD are irradiated at the same nominal energy, a value of 1 is assumed. This is done regardless of difference in field size, depth, measurement position inside or outside the treatment field, and regardless of any tissue or phantom heterogeneity. Inherent to this use of a single energy correction factor is the assumption that there is no variation in the energy spectra at that given nominal beam energy. This is clearly not a valid assumption, but the impact of this faulty assumption has not been evaluated.

The objectives of this study were, first, to evaluate the variability in the energy spectrum of a Varian 6 MV beam as a function of the treatment field size, measurement location, field modulation, and presence of heterogeneities; and second, to evaluate the impact of these variations in the photon energy spectra on the response of a common energy-dependent dosimeter, TLD.

II. MATERIALS AND METHODS

II.A. Monte Carlo simulations of photon energy spectra

Photon spectra were calculated using a Monte Carlo model of a Varian 6 MV linear accelerator that was developed using the Monte Carlo N-Particle eXtended (MCNPX) transport code⁹ and was benchmarked for in-field dose calculations within 2% and out-of-field dose calculations within 16%.^{10,11} The model geometry incorporated beam-line components including a 120 leaf multileaf collimator (MLC), head-shielding, and structural components and was used to calculate photon energy spectra in water at several locations and for various treatment parameters. A large water phantom (50 cm width $\times$ 70 cm length $\times$ 50 cm depth) was simulated at 100 cm source to surface distance (SSD). Photon flux tally voxels were added throughout the water phantom, including in-field as well as out-of-field locations, to capture the full range of energy spectra that could be clinically encountered. The spectra were calculated from 0.05 keV to 10 MeV, and

were normalized to the total photon fluence for comparisons between simulations.

Three field sizes were investigated for static fields: 5 cm $\times$ 5 cm, 10 cm $\times$ 10 cm and 20 cm $\times$ 20 cm. These three field sizes were selected to provide a range of low-energy scattered photons to the dose in the water tank, and to represent the range of field sizes commonly used in radiation therapy. The 20 cm $\times$ 20 cm field (defined by both jaws and MLC) represents a relatively soft spectrum, with a larger contribution of scattered photons from within the water tank. The smaller field size (5 cm $\times$ 5 cm, defined by jaws with the MLC completely retracted) represents a relatively hard spectrum as the head leakage contribution is more pronounced for small fields. The 10 cm $\times$ 10 cm field size was defined using the jaws with the MLC completely retracted to represent the typical reference conditions used for beam calibration and determination of the standard correction factors. For each field size, photon energy spectra were calculated for more than 30 unique positions at various depths along central axis and various depths and distances off-axis (up to 50 cm from the central axis).

The impact of field modulation through the use of IMRT on the photon spectrum was evaluated by simulating a 10 cm $\times$ 10 cm field as a series of 5 rectangular strips (each strip 2 cm $\times$ 10 cm), giving an IMRT field with a modulation factor of 5. The photon energy spectra were calculated at each of the same positions in the water phantom as for the static fields.

Finally, photon energy spectra were calculated at numerous positions with heterogeneous materials (simulated bone and lung tissue) placed within the simulated water phantom. Simulated bone and lung tissue, density of 1.85 and 0.26 g/cm³, respectively, were each considered in separate analyses as 9 cm of the heterogeneous material replacing the top 9 cm of the water phantom. Thus, the SSD was unchanged, and the positions of the tally voxels also remained constant. The photon spectra were calculated in water, below the heterogeneous material.

II.B. Impact of variations in energy spectrum on the TLD response

The energy response of TLD for non-reference measurement conditions has been divided into two components: the component due to the relative probability of photoelectric interactions (K_{PE}) and the component due to solid-state effects and all other intrinsic and secondary effects (K_{SE}). The overall energy response for measurement under non-reference conditions (K_{NR}) is the product of these two components, as shown in Eq. (3).

$$K_{\text{NR}}(E) = K_{\text{PE}}(E) \times K_{\text{SE}}(E) \quad (3)$$

II.B.1. Dosimeter response based on photoelectric effect

As mentioned previously, LiF has a slightly greater atomic number (Z) than that of tissue, which results in an enhanced probability of photoelectric interactions and energy

deposition. This increased probability of photoelectric interactions has the potential of exaggerating the response of the TLD to low-energy x-rays and gamma rays^{5,6} relative to those that would occur in tissue, resulting in an artificially high reading. To determine the impact of this effect on TLDs, we calculated the energy-response correction factor for the photon spectra resulting from each case described in the previous paragraphs.

LiF TLD capsules represent a cavity that can be placed in a medium (water) as a means to measure the absorbed dose. Thus, cavity theory can be used to theoretically determine the energy-response correction factor, given the energy spectrum of the photon beam incident on the TLD capsule. Burlin cavity theory in particular is well suited for medium-sized cavities in which the dose arises both from electrons crossing the cavity as well as from secondary electrons that are created inside of the cavity.⁵ These conditions are consistent with the interactions in LiF TLD100 (Harshaw Chemical Company, Solon, OH). The general form of Burlin cavity theory is shown in Eq. (4).⁵

$$\frac{D_{\text{cav}}}{D_{\text{med}}} = d \left(\frac{S}{\rho} \right)_{\text{med}}^{\text{cav}} + (1-d) \left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{med}}^{\text{cav}} \quad (4)$$

In this expression, D_{med} is the dose to the medium (water), D_{cav} is the dose to the cavity (TLD), (S/ρ) is the collisional stopping power, and (μ_{en}/ρ) is the mass energy absorption coefficient. Finally, d is a parameter that describes the average value of the relationship of electron fluence generated in the cavity wall (Φ_w) to the initial equilibrium electron fluence (Φ_w^e).⁵ The extreme values of 1 and 0 for d provide the traditional formalisms for small or large cavity theory.⁵ In this work, d was determined as a function of incident photon energy for each discrete energy bin used in the Monte Carlo simulations based on a 2 mm diameter TLD capsule.

$$d \equiv \frac{\bar{\Phi}_w}{\Phi_w^e} = \frac{1 - e^{-\beta L}}{\beta L} \quad (5)$$

The parameter L is equal to the mean chord length, taken as four times the TLD volume divided by its surface area. The parameter β was determined using Eq. (6), where t_{max} is the maximum depth of electron penetration, approximated using the continuous slowing down approximation range for the maximum electron energy in each discrete energy bin. In

this equation, the value of 0.04 has been previously found to be the most consistent with experimental data.⁵

$$e^{-\beta t_{\text{max}}} = 0.04 \quad (6)$$

The right side of Eq. (4) is denoted “MCT” for “medium-sized cavity theory” for the remainder of this work. Thus,

$$MCT = d \left(\frac{S}{\rho} \right)_{\text{med}}^{\text{cav}} + (1-d) \left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{med}}^{\text{cav}}, \quad (7)$$

and the response of the TLD is given by:

$$\begin{aligned} D_{\text{cav}} &= D_{\text{med}} \left[d \left(\frac{S}{\rho} \right)_{\text{med}}^{\text{cav}} + (1-d) \left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{med}}^{\text{cav}} \right] \\ &= D_{\text{med}} MCT_{\text{med}}^{\text{cav}}. \end{aligned} \quad (8)$$

Absorbed dose can be reasonably approximated by kerma for each of the selected positions in the phantom, given the existence of electronic equilibrium in these locations.¹⁰ The absorbed dose to water was therefore determined using Eq. (9) where ψ_E is the energy fluence, which was calculated using the results of the Monte Carlo simulations for each location.

$$D_w = \sum_E \Psi_E \left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{water},E}. \quad (9)$$

The dose to LiF (inside the TLD) can then be determined Eq. (10) by the application of Burlin cavity theory, as described in Eq. (8), and the dose to water, as described in Eq. (9).

$$D_{\text{LiF}} = \sum_E \left(\Psi_E \frac{\mu_{\text{en}}}{\rho_{\text{water},E}} \times (MCT_{\text{water}}^{\text{LiF}})_E \right). \quad (10)$$

An appropriate non-reference energy-response correction factor accounts for the differences in absorbed dose between the point of measurement (at some arbitrary position away from the reference position, or under non-reference conditions) and the point of reference, where the dose calibration has taken place. For dosimetry using TLDs, the point of reference is found at the depth of maximum dose on the central axis for a 10 cm × 10 cm field size. A dose measured at an arbitrary position would be multiplied by the factor defined in Eq. (11) to correct for the response of the detector due to variations in the photon energy spectrum at the arbitrary position, such that it is equivalent to the response expected at the reference position.

$$K_{\text{PE}} = \frac{D_{\text{LiF,reference}}}{D_{\text{LiF,non-reference}}} = \frac{\sum_j \left[\frac{\phi_j E_j \left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{water},j} \times (MCT_{\text{water}}^{\text{LiF}})_j}{\sum_i \phi_i E_i \left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{water},i}} \right]_{\text{reference}}}{\sum_j \left[\frac{\phi_j E_j \left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{water},j} \times (MCT_{\text{water}}^{\text{LiF}})_j}{\sum_i \phi_i E_i \left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{water},i}} \right]_{\text{non-reference}}}. \quad (11)$$

Thus, Eq. (11) is equal to the calculated energy correction factor for non-reference conditions and effectively corrects for the different interaction coefficients arising from differences in the photon energy spectra between the point of measurement and the reference position.

II.C. Determination of secondary effects on TLD response

While the above approach accounts for theoretical differences arising from differences in the interaction coefficients, there is also a known solid-state energy response. Solid-state effects, such as the charge density effect, are those effects which introduce variation in the energy deposition on the scale of the electron track length. These effects are not predicted by interaction coefficients. To isolate the different components and to determine the total energy correction factor for non-reference conditions, a set of measurements were made with both ion chambers and TLDs. The absorbed dose was measured for several positions corresponding to locations where the photon energy spectra were simulated using the Monte Carlo model. The component of the TLD energy response that is due to the solid-state energy response and other secondary effects that do not include differences in photoelectric interactions is described in Eq. (12).

$$K_{SE} = \frac{D_{\text{ion chamber}}}{D_{\text{TLD}}} \times \frac{1}{K_{PE}} \quad (12)$$

All measurements were performed using a Varian Clinac 21EX accelerator (Varian Medical Systems, Palo Alto, CA) using 6 MV photons in a water phantom (Blue Phantom, Scanditronix Wellhofer, Uppsala, Sweden). At each location in the water phantom, dose was measured using both TLDs and a PTW Farmer-type ionization chamber (type 30001, 0.6 cm³ volume). Measurement locations included in-field and out-of field locations for the 10 cm × 10 cm, 5 cm × 5 cm and 20 cm × 20 cm field sizes. The depth in the water phantom was also varied and dose was measured for depths of 1.6, 5, and 20 cm.

The measured charge at the thirteen selected positions in the water phantom were adjusted according to the protocol outlined in AAPM TG-21 (Ref. 12) to determine dose at each location based on the photon spectrum at that location (as calculated by the Monte Carlo simulations). As this protocol extends only to energies as low as 1.25 MeV (Co-60), additional consideration was required to extrapolate correction factors to lower energies when needed. The TG-21 protocol was more appropriate for this purpose than the newer TG-51 protocol because cavity theory is explicitly included in this formalism, whereas the TG-51 formalism is simplified by the combining of several factors. Using the chamber reading, M_{raw} , the dose to the medium (water) was determined using the following expression:¹²

$$D_{\text{med}} = \bar{M}_{\text{raw}} P_{\text{TP}} P_{\text{elec}} P_{\text{pol}} N_{\text{gas}} \left(\frac{L}{\rho} \right)_{\text{gas}}^{\text{med}} P_{\text{ion}} P_{\text{repl}} P_{\text{wall}}. \quad (13)$$

Each measurement of charge was corrected for temperature and pressure (P_{TP}), the electrometer factor (P_{elec}), and polar-

ity effects (P_{pol}). The cavity-gas calibration factor (N_{gas}) was determined using the ratio of the chamber's absorbed dose to water calibration factor ($N_{D,w}$, provided by an ADCL) and data in the literature for N_x for the chamber used in this study (PTW 30001).¹³ As this value is considered constant for all photon energies that expend an average of 33.97 eV to produce one ion pair [which is the case for photon energies as low as 10 keV (Ref. 12)] it was sufficient for all of the measurements collected in this study, regardless of the non-reference conditions. The ratio of the mean restricted stopping power for the phantom material (water) to the chamber gas (air) was determined as a function of average photon energy at each measurement location using published values.^{12,14} The ion-recombination correction factor (P_{ion}) was directly measured. The replacement correction factor (P_{repl}) is dependent on the photon energy and the radius of the chamber cavity and was estimated using the TG-21 protocol (Ref. 12) and data reported by Cunningham and Sontag¹⁵ for each individual measurement position. Extrapolation was necessary for photon spectra with average energy less than 1.25 MeV, but uncertainty in the extrapolated values was expected to be small (<1%) based on the minimal dependence on energy. Finally, the P_{wall} factor was determined using the expression in Eq. (14), where α is the fraction of the total ionization produced by electrons arising in the chamber wall.¹²

$$P_{\text{wall}} = \frac{\left[\alpha \left(\frac{L}{\rho} \right)_{\text{gas}}^{\text{wall}} \left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{wall}}^{\text{med}} + (1 - \alpha) \left(\frac{L}{\rho} \right)_{\text{gas}}^{\text{med}} \right]}{\left(\frac{L}{\rho} \right)_{\text{gas}}^{\text{med}}} \quad (14)$$

For photon energies lower than 1.25 MeV (⁶⁰Co), Lempert *et al.* has shown that α is equal to unity¹⁶ for our chamber type. Values for the stopping power and mass energy absorption coefficients as a function of photon energy were determined from tabulated data in the literature, and P_{wall} was individually determined for each of the measurement locations.

These measurements were repeated using TLDs in a customized holder that simulated the shape and size of an ion

FIG. 1. Customized solid water TLD holder.

FIG. 2. Variations in photon energy spectra as a function of depth along central axis for a 6 MV 10 cm $\times$ 10 cm field.

chamber, shown in Fig. 1. Two TLDs were irradiated for each measurement, and the TLDs were read according to the protocol used by the Radiological Physics Center (RPC, MD Anderson Houston, TX) and were individually corrected for sensitivity, linearity, and fading and reference energy⁸ as previously described.

III. RESULTS

III.A. Spectral variations

Photon energy spectra were calculated using MCNPX. The energy spectra for the most pertinent cases of the variety of sampled locations are shown in Figs. 2–7. The particle flux was tallied for each simulation using a logarithmic energy binning structure from 1 keV to 10 MeV. The top energy bin collected particles in the energy range of 7.36–10 MeV, and therefore has a value of zero for all 6 MV simulations. All curves were normalized to have the same area, and

FIG. 3. Photon energy spectra at depth of 10 cm in water along central axis for three field sizes.

FIG. 4. Variations in out-of-field photon energy spectra as a function of distance from the central axis for 10 cm $\times$ 10 cm field.

error bars are included based on the 1-sigma statistical uncertainty in the MCNPX simulations.

III.A.1. Variations in in-field photon spectra

The variations in photon energy spectra for a 6 MV therapeutic beam along the central axis at depths of 0.1 mm, 1.6 cm (reference spectrum), 10 cm, and 25 cm for the 10 cm $\times$ 10 cm field size are shown in Fig. 2. The spectra show some minor variation with depth; however they appear to be qualitatively similar, peaking at approximately the same energy.

The variations in photon energy spectra for three field sizes (5 cm $\times$ 5 cm, 10 cm $\times$ 10 cm, and 20 cm $\times$ 20 cm) are compared in Fig. 3 at a depth of 10 cm on the central axis in water. As expected, the 5 cm $\times$ 5 cm field resulted in a slightly harder spectrum relative to the spectrum from the 10 cm $\times$ 10 cm field while the 20 cm $\times$ 20 cm field (designed to maximize scatter) resulted in a softer spectrum relative to the spectrum from the 10 cm $\times$ 10 cm field.

FIG. 5. Variations in out-of-field photon energy spectra for three different field sizes.

Fig. 6. Variation in photon energy spectra for both an in-field and out-of-field measurement location for an IMRT field compared to 10 cm $\times$ 10 cm open field.

III.A.2. Variations in out-of-field photon spectra

The photon energy spectra were determined for each of the three field sizes at several distances from the central axis for three different depths in water: 1.6, 5, and 20 cm. As the distance away from the central axis increased, the photon energy spectra became progressively softer. In general, the photon spectrum out-of-field would be composed of a dominant low-energy component due to patient scatter and a high-energy component due to head leakage. While the spectra changed minimally with off-axis distance within the treatment field, the low-energy scatter component became more pronounced at increasing distances outside of the treatment field. Furthermore, for out-of-field locations, changes in depth induced a more pronounced difference in the spectra than increased distance off-axis. This is depicted in Fig. 4 and 5, in Fig. 4 as a function of distance from the central axis for a 10 cm $\times$ 10 cm field size and in Fig. 5 as a function of field size for an out-of-

Fig. 7. Variation in photon energy spectra with depth along central axis due to presence of bone tissue and lung tissue.

field measurement location. For contrast, the photon energy spectrum at $d_{\max}$ on the central axis is also included.

III.A.3. Variations in spectra for IMRT field

A single IMRT field was simulated and the resulting photon energy spectra were calculated at several locations with varying depth and distance from central axis. Two of these spectra are shown in Fig. 6 along with the spectra for the corresponding locations for a 10 cm $\times$ 10 cm open field. At a depth of $d_{\max}$ on central axis, the IMRT photon energy spectrum was harder than the open 10 cm $\times$ 10 cm field energy distribution. This trend is less pronounced for out-of field positions, where the spectra are very similar. The energy spectra for the IMRT field and the 10 cm $\times$ 10 cm open field at a distance 10 cm from central axis were nearly identical. The out-of field spectra show contributions from low-energy phantom scatter and relatively high-energy head leakage, consistent with other out-of-field spectra.

III.A.4. Variations in spectra in presence of heterogeneities

The results for the simulations which included two common heterogeneous materials (bone and lung) are shown in Fig. 7 for a 10 cm $\times$ 10 cm field at varying depths in water. The reference spectrum in homogeneous media (water) is also plotted in Fig. 7 for comparison. There is little difference in the spectra below the heterogeneous materials compared to the reference spectrum in homogeneous material, illustrating that the composition of the tissue has a minimal impact on the photon energy spectra. These results are consistent with the out-of field spectra; the presence of a heterogeneous material does not further perturb the photon spectra for out-of field measurement locations. For a 6 MV photon beam, beam hardening is relatively minimal, even for a large amount of bone tissue. Furthermore, the energy spectrum through 9 cm of lung material has been shown to not be substantially different from the photon energy spectrum in water.

Fig. 8. Variation of parameter d as a function of incident photon energy.

TABLE I. Calculated K_{PE} for in-field spectral variations (Monte Carlo) due to field size and depth.

Condition, field size cm × cm	Depth (cm)	Mean photon energy (MeV)	Energy correction factor (K_{PE})
Reference position: 10 × 10	1.6	1.49	1.00
Reference field size, depth			
10 × 10	5	1.41	0.99
10 × 10	10	1.39	0.99
10 × 10	25	1.53	1.00
In field, hard spectrum: 5 × 5	10	1.70	1.01
In field, soft spectrum: 20 × 20	10	1.06	0.99
Bone heterogeneity, 10 × 10	10	1.35	0.99
Lung heterogeneity, 10 × 10	10	1.50	0.99
IMRT field	1.6	1.68	1.00
IMRT field	5	1.54	1.00

III.B. Impact of spectral variations on energy response according to cavity theory

The non-reference condition energy correction factor based on the impact of the interaction coefficients (K_{PE}) was calculated as described above using Burlin cavity theory for each calculated photon energy spectra. The parameter d was calculated as a function of incident photon energy and is shown in Fig. 8. Using tabulated stopping power data, the calculated parameter d , and the simulated photon energy spectra for each of the various tally positions, K_{PE} was determined for the different measurement positions and treatment parameters.

Table I summarizes the mean photon energy and the calculated values of K_{PE} for selected in-field positions with varying treatment parameters, including the addition of tissue heterogeneities and field modulation. The positions described in the table were all measured along the central axis and, for heterogeneous cases, beneath the heterogeneous material. The energy correction factor, K_{PE} , varied by less than 1% over the range of irradiation conditions and energy spectra investigated for in-field locations.

Relative to the reference position there was little variation in the cavity-theory-based response of the TLD resulting from differences in the photon energy spectra, despite the visible variations in the photon energy spectra (e.g., Fig. 7).

FIG. 9. K_{PE} for various field sizes in water and presence of heterogeneities, as a function of mean photon energy.

At these locations, and all other in-field locations, the data showed similar results; the TLD response to the non-reference spectra was less than 1% different from that of the reference spectra as a result of differences in interaction coefficients.

The energy response of the TLD showed more variation for out-of-field measurement locations than for in-field measurement locations. The mean energy was substantially lower at these locations – in some cases less than a fifth of the mean energy at the reference position. The lower energy is associated with increased photoelectric interactions in the TLD at these positions, and, as a result, increased discrepancy between the dose measured by the TLD and the dose actually absorbed by the tissue. Table II summarizes the non-reference energy correction factor based on interaction coefficients (K_{PE}) for measurement positions that were made outside of the treatment field for varying treatment parameters. As can be seen in the table, the TLD may over-respond by as much as 7% based on cavity theory (corresponding to a K_{PE} of 0.93). These data displayed the greatest degree of variation both in the photon energy spectra and K_{PE} .

The non-reference energy-response correction factor based on cavity theory (K_{PE}) is shown in Fig. 9 as a function of mean photon energy. This figure includes the three field sizes examined as well as the energy correction factors

TABLE II. Calculated K_{PE} for spectral variations (determined via Monte Carlo) occurring at out-of-field locations.

Condition, field size (cm × cm)	Depth	Distance off-axis	Mean photon energy (MeV)	Energy correction factor (K_{PE})
Reference position, 10 × 10	1.6	0	1.49	1.00
Near treatment field, 10 × 10	5	15	0.31	0.96
Far from treatment field, 10 × 10	20	50	0.22	0.94
Bone heterogeneity, 10 × 10	20	50	0.22	0.94
Lung heterogeneity, 10 × 10	20	50	0.24	0.95
Out of field, hard spectrum: 5 × 5	20	50	0.23	0.95
Out of field, soft spectrum: 20 × 20	20	50	0.17	0.93
IMRT field	1.6	10	0.41	0.98
IMRT field	5	15	0.34	0.97

TABLE III. Total energy correction factor (K_{NR}) for selected locations with varying mean photon energy.

Field size (cm $\times$ cm)	Depth (cm)	Distance off-axis (cm)	Mean photon energy (MeV)	K_{PE}	K_{SE}	Total energy correction factor (K_{NR})
20 $\times$ 20	20	30	0.19	0.94	0.95	0.89
20 $\times$ 20	5	20	0.20	0.95	0.94	0.89
20 $\times$ 20	20	20	0.23	0.95	0.97	0.93
10 $\times$ 10	5	30	0.30	0.96	0.98	0.95
10 $\times$ 10	5	15	0.31	0.96	0.97	0.93
10 $\times$ 10	20	15	0.31	0.97	0.95	0.92
5 cm $\times$ 5 cm	5	15	0.54	0.98	1.00	0.98
20 $\times$ 20	25	0	1.08	1.00	1.00	1.00
20 $\times$ 20	1.6	0	1.29	1.00	1.00	1.00
10 $\times$ 10	20	0	1.46	1.00	1.00	1.00
5 cm $\times$ 5 cm	1.6	0	1.65	1.00	0.99	0.99
5 cm $\times$ 5 cm	20	0	1.85	1.00	1.01	1.01

determined when heterogeneous materials were added to the phantom. The response follows the mean photon energy, with larger correction factors required as the photon energy decreases.

III.C. Determination of total energy correction factor

To determine the total energy correction factor, including both interaction coefficient-based effects (cavity theory) (K_{PE}) and solid-state and other secondary effects (K_{SE}), a series of measurements were performed using the TLDs and an ionization chamber. A measured energy correction factor was determined using the ratio of the fully corrected ion chamber measured dose to the TLD measured dose. This measured correction factor for non-reference measurement conditions [K_{NR} , defined previously in Eq. (3)] includes components of the difference in photoelectric interaction probability as well as all other secondary effects, and it is provided in Table III for several selected measurement locations with varying spectra. The value of K_{NR} for the reference position (10 cm $\times$ 10 cm, depth of 1.6 cm, along central axis) is 1.00 by definition.

The results of our data showed that for mean photon energies of 500 keV and greater, K_{SE} was nearly unity, and so for

all in-field locations the non-reference energy correction factor (K_{NR}) was virtually unity. For those measurement positions where the photon energy spectra were lower in energy (out-of-field locations), the values of K_{SE} deviated from unity. For energy spectra with a mean energy of less than 500 keV, measured data showed that the TLDs over-responded by an average of 3.5% more than that predicted by the theoretical calculations using cavity theory. A graphical representation of the agreement between K_{NR} and K_{PE} is provided in Fig. 10. This additional over-response is not due to the increased photoelectric effect due to higher atomic number, but rather due to solid-state effects such as charge density effects or other secondary effects.

In general, about one third of the energy response was due to solid-state effects, while two thirds of the energy response was due to differences in interaction coefficients as described by cavity theory. Overall, the total energy correction is unity within the treatment field, but increased to approximately 12% for the softest spectra outside of the treatment field. The total uncertainty in the ion chamber measurements of absorbed dose, based on the uncertainty in each correction factor is estimated to be less than 2%. The total uncertainty in the TLD measurements based on the RPC TLD protocol is estimated to be less than 3%.

IV. DISCUSSION

Variations in the photon energy spectra as a function of field size, depth in water, the presence of heterogeneity, field modulation, and measurement location were quantified for 6 MV photon fields. The impact of these variations on dose measurement using a common energy-dependent dosimeter (TLD) was determined based on cavity theory, and was also measured to account for solid-state effects.

The non-reference energy correction factor was determined for several measurement positions with varying treatment parameters, representing a clinically useful range of field sizes and treatment techniques including conventional RT and IMRT. This factor, K_{NR} , varied from 0.88 to 1.01 among the situations examined as part of this work. Within the treatment field, over all field sizes and depths examined, the response of the TLD varied by less than 1% relative to the reference point (d_{max} , on the central axis from a

FIG. 10. Comparison of theoretical energy response (K_{PE}) and total measured response (K_{NR}).

Fig. 11. Literature comparison of TLD response as a function of photon energy.

10 cm $\times$ 10 cm field) because of variations in the photon energy spectrum. The presence of heterogeneities in the field did not greatly impact the photon energy distribution along the central axis, nor did field modulation, and corrections to the measured dose would be less than 1% for these situations. It may be possible that more highly-modulated IMRT fields could induce a notable energy-response by the TLD, and further analysis of more complex IMRT plans and field arrangements could be performed to further explore these initial findings. However, based on the extensive range of conditions explored in this work, in-field dose measurement from 6 MV photon fields using TLD do not require an additional non-reference energy correction factor.

Calculated dose to water has been approximated with kerma, which is a valid assumption under conditions of electronic equilibrium; however this condition is not met at the surface of the phantom or in the field penumbra. This may be a relevant concern as TLD are often placed on the patient surface as a method of estimating dose to the patient. Our methodology may have increased uncertainty in these situations. Simulations performed at the surface of the phantom for a 10 cm $\times$ 10 cm field size along central axis show that there is little variation in the photon energy spectrum from the spectrum at the reference position (shown in Fig. 2). Furthermore, there is little variation in the value of the ratio of stopping powers (S/ρ) between water and LiF over the relevant energy range. These two relationships indicate that neither photons nor electrons should affect the response of the dosimeter at the surface relative to the response at $d_{\max}$. Therefore, despite the increased uncertainty in the surface location, we believe K_{NR} to be nearly identical to the value at $d_{\max}$. For surface measurements made out of field, a bath of contaminating electrons results in an elevated dose at the surface relative to the dose at depth, potentially by up to a factor of 6.¹⁰ Without adequate buildup material, dose to the TLD is not likely to be representative of dose to the patient. For these locations, the TLD should be covered with buildup material (such as bolus), to bring the TLD to an effective depth of $d_{\max}$. Measurements

conducted at depth of $d_{\max}$ for locations outside of the treatment field would then use the appropriate non-reference energy correction factor as described in Table III.

For locations outside of the treatment field, energy correction factors varied more extensively. The results showed that TLD over-responded by up to 12% when used to measure dose outside of the treatment field relative to within the treatment field. The majority of this over-response was due to the atomic number of TLD altering the photon and electron interaction coefficients relative to water. The magnitude of this energy response may be relevant to TLD-based measurements of out-of-field dose.

Due to the theoretical basis and assumptions used to derive the cavity theory [Eq. (10)], it is necessary to test the validity of the results obtained with its use. This was done by determining K_{PE} for monoenergetic photons following the procedure outlined above and normalizing relative to ^{60}Co . The calculated energy correction factors were then compared with those in the literature, which had been determined either through measurement or Monte Carlo for monoenergetic photon beams.^{17–19} This comparison is shown in Fig. 11 and shows very good agreement, indicating that the theoretical approach we have taken is sound. The only discrepancy is at very low energies (below those energies relevant to this study), where self-attenuation of the TLD is known to reduce its sensitivity.

The second component (K_{SE}) due to secondary effects, which is not predicted by cavity theory, was demonstrated in the measured data to be relevant at photon energies less than 500 keV. According to the data, the magnitude of this additional over-response can range between 2 and 5% with an average value of 3.5% for photon energies between 10 and 500 keV. The magnitude of this effect is comparable to that described in the literature, which has suggested an over-response of less than 10% at comparable or even lower photon energies.²⁰

This study has resolved the total energy-response factor (K_{NR}) into its two components; the first of which has been calculated using fundamental cavity theory, and the second was examined through the measured data. The total energy-response correction factor (K_{NR}) that was determined in this study accounts for deviations in the photon energy spectra from the reference point of calibration for measurement locations that do not coincide with the point of calibration. This factor does not replace the TLD energy correction factor discussed in the introduction but rather should be applied multiplicatively to it depending on the measurement conditions. A revised expression for determining absorbed dose from TLD measurements is provided in Eq. (15), where K_{NR} is the energy correction factor dependent on the measurement conditions and location.

$$D = T \times S \times K_L \times K_F \times K_E \times K_{\text{NR}} \quad (15)$$

This energy-response correction factor will correct for variations in the photon energy spectra when measurements are taken under noncalibration conditions, such as out-of-field measurements.

V. CONCLUSION

Currently, TLD are used to measure or verify the absorbed dose for many applications in therapeutic physics, including measurement of numerous points within a wide range of treatment fields as well as outside of the treatment field. The energy correction factor determined in this work should be applied multiplicatively to thermoluminescent signal to account for varying detector responses with different photon spectra. In general, it is acceptable to ignore variations in the photon spectrum within the treatment field because there is a negligible impact on the measured dose. However, outside the treatment field, the spectra are much softer, and a correction factor is generally appropriate. The value of this factor has been shown to range from 0.88 to 0.99 depending on the specific irradiation conditions.

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¹H. H. Liu and F. Verhaegen, "An investigation of energy spectrum and lineal energy variations in mega-voltage photon beams used for radiotherapy," *Radiat. Prot. Dosim.* **99**(1–4), 425–427 (2002).

²S. Y. Jang, H. H. Liu, R. Mohan, and J. V. Siebers, "Variations in energy spectra and water-to-material stopping-power ratios in three-dimensional conformal and intensity-modulated photon fields," *Med. Phys.* **34**(4), 1388–1397 (2007).

³G. X. Ding, "Energy spectra, angular spread, fluence profiles and dose distributions of 6 and 18 MV photon beams: Results of Monte Carlo simulations for a Varian 2100EX accelerator," *Phys. Med. Biol.* **47**(7), 1025–1046 (2002).

⁴K. Kase, G. Svensson, A. Wolbarst, and M. Marks, "Measurements of dose from secondary radiation outside a treatment field," *Int. J. Radiat. Oncol., Biol., Phys.* **9**, 7 (1983).

⁵F. H. Attix, *Introduction to Radiological Physics and Radiation Dosimetry* (Wiley-VCH Verlag GmbH & Co., Weinheim, Germany, 2004).

⁶G. F. Knoll, *Radiation Detection and Measurement Third Edition* (John Wiley & Sons, New York, 2000).

⁷G. Shani, *Radiation Dosimetry: Instrumentation and Methods*, 2nd ed. (CRC, Boca Raton, Florida, 2001).

⁸T. H. Kirby, W. F. Hanson, and D. A. Johnston, "Uncertainty analysis of absorbed dose calculations from thermoluminescence dosimeters," *Med. Phys.* **19**(6), 1427–1433 (1992).

⁹J. S. Hendricks, MCNPX, Version 2.5. e, LA-UR-04-0569, 2004.

¹⁰S. F. Kry, U. Titt, F. Ponisch, D. Followill, O. N. Vassiliev, R. A. White, R. Mohan, and M. Salehpour, "A Monte Carlo model for calculating out-of-field dose from a Varian 6 MV beam," *Med. Phys.* **33**(11), 4405–4413 (2006).

¹¹F. Ponisch, U. Titt, S. F. Kry, O. N. Vassiliev, and R. Mohan, "MCNPX simulation of a multileaf collimator," *Med. Phys.* **33**(2), 402–404 (2006).

¹²RTC Task Group 21, AAPM, "A protocol for the determination of absorbed dose from high-energy photon and electron beams," *Med. Phys.* **10**(6), 741 (1983).

¹³R. C. Tailor, W. F. Hanson, N. Wells, and G. S. Ibbott, "Consistency of absorbed dose to water measurements using 21 ion-chamber models following the AAPM TG51 and TG21 calibration protocols," *Med. Phys.* **33**(6), 1818 (2006).

¹⁴H. E. Johns and J. R. Cunningham, *The Physics of Radiology*, 4th ed. (Charles C. Thomas, Springfield, IL, 1983).

¹⁵J. R. Cunningham and M. R. Sontag, "Displacement corrections used in absorbed dose determination," *Med. Phys.* **7**(6), 672 (1980).

¹⁶G. D. Lempert, R. Nath, and R. J. Schulz, "Fraction of ionization from electrons arising in the wall of an ionization chamber," *Med. Phys.* **10**(1), 1 (1983).

¹⁷A. A. Nunn, S. D. Davis, J. A. Micka, and L. A. Dewerd, "LiF:Mg, Ti TLD response as a function of photon energy for moderately filtered x-ray spectra in the range of 20–250 kVp relative to Co-60," *Med. Phys.* **35**(5), 1859–1869 (2008).

¹⁸S. D. Davis, C. K. Ross, P. N. Mobit, L. Van Der Zwan, W. J. Chase, and K. R. Shortt, "The response of LiF thermoluminescence dosimeters to photon beams in the energy range from 30 kV x rays to 60Co gamma rays," *Radiat. Prot. Dosim.* **106**(1), 33–43 (2003).

¹⁹A. Konnai, N. Nariyama, S. Ohnishi, and N. Odano, "Energy response of LiF and Mg2SiO4 TLDs to 10–150 keV monoenergetic photons," *Radiat. Prot. Dosim.* **115**(1–4), 334–336 (2005).

²⁰Y. S. Horowitz, *Thermoluminescence and Thermoluminescent Dosimetry*, 1st ed. (CRC Boca Raton, FL, 1984), Vol. 2.