Monte Carlo calculations of the absorbed dose and energy dependence of plastic scintillators

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Detector systems using plastic scintillators can provide instantaneous measurements with high spatial resolution in many applications including small field and high dose gradient field applications. Energy independence and water equivalence are important dosimetric properties that determine whether a detector will be useful in a clinical setting. Using Monte Carlo simulations, we calculated the energy dependence of plastic scintillators when exposed to photon beams in the radiotherapeutic range. These calculations were performed for a detector comprised of a BC-400 plastic scintillator surrounded by a polystyrene wall. Our results showed the plastic scintillation detector to be nearly energy independent over a range of energies from 0.5 to 20 MeV. The ratio of the dose absorbed by the scintillator to that absorbed by water was nearly a constant, approximately equal to 0.98 over the entire energy range of interest. These results confirm the water equivalence of the plastic scintillation detector and are in very good agreement with earlier results obtained using Burlin cavity theory.
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Key words: Monte Carlo, Burlin cavity theory, plastic scintillation dosimetry, scintillator, water equivalence

I. INTRODUCTION

Detector systems using plastic scintillators can provide instantaneous measurements with high spatial resolution in small field and high dose gradient field applications. These small-volume detectors offer reproducibility, linear response, resistance to radiation damage, temperature independence, and superior spatial resolution. Applications using plastic scintillation detectors include dosimetry of high-energy photon and electron beams,[1,2] quality assurance of ^{60}Co and high-energy therapy machines,[3] ophthalmic plaque dosimetry,[4,5] and stereotactic radiosurgery dosimetry.[6,7] Plastic scintillators have even been used for *in vivo* β-particle detection in the brains of small animals[8,9] and intravascular brachytherapy for in stent restenosis.[10]

Energy independence and water equivalence are important dosimetric properties that determine whether a detector can be widely used in a clinical setting or languish on a laboratory shelf. The material that comprises a plastic scintillator has previously been shown to be a good match to water in the mega electron volt energy range.[1] The mean mass-energy absorption coefficients of the plastic scintillator are quite close to those of water above 100 keV, as are the mass collision stopping powers and the mass angular scattering powers. The dose absorbed by plastic scintillators has previously been calculated using Burlin cavity theory,[1] and the results showed near energy independence for photons within the megavoltage energy range. Kirov *et al.*[11] and Williamson *et al.*[12] studied the same scintillator material for photon energies under 1 MeV. The results of both Monte Carlo calculations and experiment showed a steep drop in the absorbed dose with decreasing energy, making the unadulterated plastic scintillator less than ideal for brachytherapy and other low-energy applications.

Monte Carlo simulations have not previously been used to calculate the dose absorbed by plastic scintillators for energies above 1 MeV. Our purpose, therefore, was to calculate, using Monte Carlo simulations, the absorbed dose to a plastic scintillator embedded in a polystyrene probe. The calculations were performed for a range of energies typically used in accelerator beam radiation therapy, and both monoenergetic and polyenergetic beams were used. Finally, we compared the results of our Monte Carlo calculations with those determined using Burlin cavity theory.

II. METHODS AND MATERIALS

A. Monte Carlo calculations with monoenergetic beams

Dose to the scintillator and water was calculated using Monte Carlo simulations with the Monte Carlo N-particle transport code (MCNPX, version 2.4.k, Los Alamos National Laboratory, Los Alamos, NM). The most important features of MCNPX relevant to electron-photon transport are the upgrades to the Integrated Tiger Series (ITS) version 3.0 electron physics. The electron-physics enhancements include changes to the density-effect calculation for collision (electronic) stopping power, radiative stopping power, bremsstrahlung production (spectra and angular distribution), and electron-impact ionization.[13] The plastic scintillation detector system was modeled on a system designed by Beddar *et al.*[1,2]

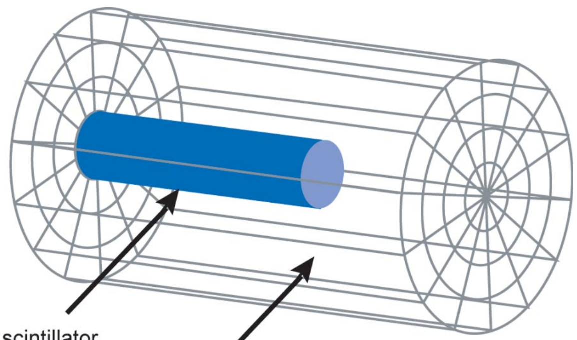

FIG. 1. Diagram of the polystyrene probe containing the scintillator.

and consisted of a plastic scintillator embedded in a polystyrene probe (Fig. 1). The chemical composition of the scintillator was based on the BC-400 plastic scintillator (Bicron Corporation, Newbury, OH), which is made from polyvinyltoluene doped with organic fluors. The scintillator detecting volume was 1 mm in diameter and 4 mm in length, and the polystyrene probe was 4 mm in diameter and 7 mm in length. The probe was surrounded by a $20\times20\times20$ cm^3 water bath to simulate immersion in a water phantom. Table I lists the density and composition by weight of the materials used in these simulations. Monoenergetic photon beam simulations with energies of 0.2, 0.5, 1, 2, 4, 6, 8, 10, 15, and 20 MeV were placed at the topmost surface of the water phantom, and the center of the scintillator was placed 5 cm below. Both the scintillator and the polystyrene probe were replaced by water, and the absorbed dose was calculated for the same volume as the plastic scintillator.

All coupled photon–electron simulations were performed on a Pentium 42.0 GHz laptop computer operating WINDOWS 2000 PROFESSIONAL. Each calculation took, on average, 14 h to complete. The electron energy cutoff was set to 100 keV, while the photon energy cutoff was set to 10 keV. This was thought to be a valid cutoff energy because the detector diameter of 1 mm is larger than the range of a 100 keV electron in the media investigated. The default MCNP4C and MCNPX 2.4 energy-indexing algorithm was changed to the ITS-style algorithm, as discussed in the literature.[14,15] The MCNPX electron step sizes were chosen as a default to result in an average energy loss of 8.3% per step, as recommended

TABLE I. Physical properties of the scintillator, polystyrene, and water.

Parameter	Scintillator	Polystyrene	Water
Density (g/cm^3)	1.032	1.060	1.000
Ratio of the number of e^- in the compound to the molecular weight $\langle Z/A\rangle$	0.5414	0.5377	0.5551
Electron density (10^{23} e^-/g)	3.272	3.238	3.343
Composition (wt %)	H: 8.47 C: 91.53	H: 7.74 C: 92.26	H: 11.19 O: 88.81

by Berger.[16] These steps were further broken into a number of equal-length substeps, the default electron substeps per energy step of 3 for water, polystyrene, and polyvinyltoluene were used.[15] The absorbed dose was obtained by tallying the energy within the scintillator or equivalent water volume and dividing the result by the mass of the scintillator or water to convert to dose. For each energy simulation, 200 million histories were transported to obtain a statistical relative error of 2%–4% (1 s.d.), with higher percentages of errors occurring for the lower photon energies. The statistical error in the calculation of absorbed dose to the scintillator was about the same as that for the calculation of absorbed dose to water at the same energy. All ten statistical checks provided by the MCNPX output indicated convergence at the completion of the simulation.

B. Monte Carlo calculations with a polyenergetic beam

For the Monte Carlo calculations using polyenergetic photon beams, radiation transport in the accelerator was modeled using the BEAMnrc Monte Carlo code.[17–20] The accelerator geometry and materials were obtained from the manufacturer's data for the Varian Clinac 2100 series accelerator (Varian Medical Systems, Palo Alto, CA). The study was limited to the 6 and 18 MV photon modes. The accelerator model was built using the standard BEAMnrc component modules, which include a bremsstrahlung target, a primary collimator, a flattening filter, and two movable collimators (jaws). Properties of the electron beam incident on the target are not known with sufficient accuracy; therefore, on the basis of previous studies,[21,22] the beam was assumed to be parallel, and the energy spectrum and lateral spread of the electron fluence were assumed to be Gaussian. The three parameters describing the electron beam, the center and width of the energy distribution and the width of the lateral spread, were determined by comparing simulation results with dosimetry data for a water phantom. The experimental data included depth-dose data, dose profiles, and output factors. Best agreement between simulation results and dosimetry data for the 6 MV beam was achieved for an electron beam with an energy spectrum centered at 6.2 MeV, a full width at half maximum (FWHM) of 3%, and a lateral spread of FWHM $=1$ mm. In the 18 MV mode, the spectrum was centered at 18.0 MeV and had the same FWHM values, 3% and 1 mm. The bulk of the data agree to within 2%, or 2 mm. Larger discrepancies were found mainly in low-dose regions (outside the penumbra) and in data for a 40×40 cm^2 field. Fortunately, these cases are of little relevance to the present study.

Calculations were performed for a 10×10 cm^2 field. The number of simulated histories was set to 1.5×10^8 (6 MV) and 4×10^7 (18 MV). To improve efficiency, selective bremsstrahlung splitting was used, with respective minimum and maximum splitting numbers of 20 and 200 and an effective field size of 20×20 cm^2. Phase space coordinates of all particles reaching a plane located at a source-to-surface distance (SSD) of 100 cm were recorded in a file. The data were

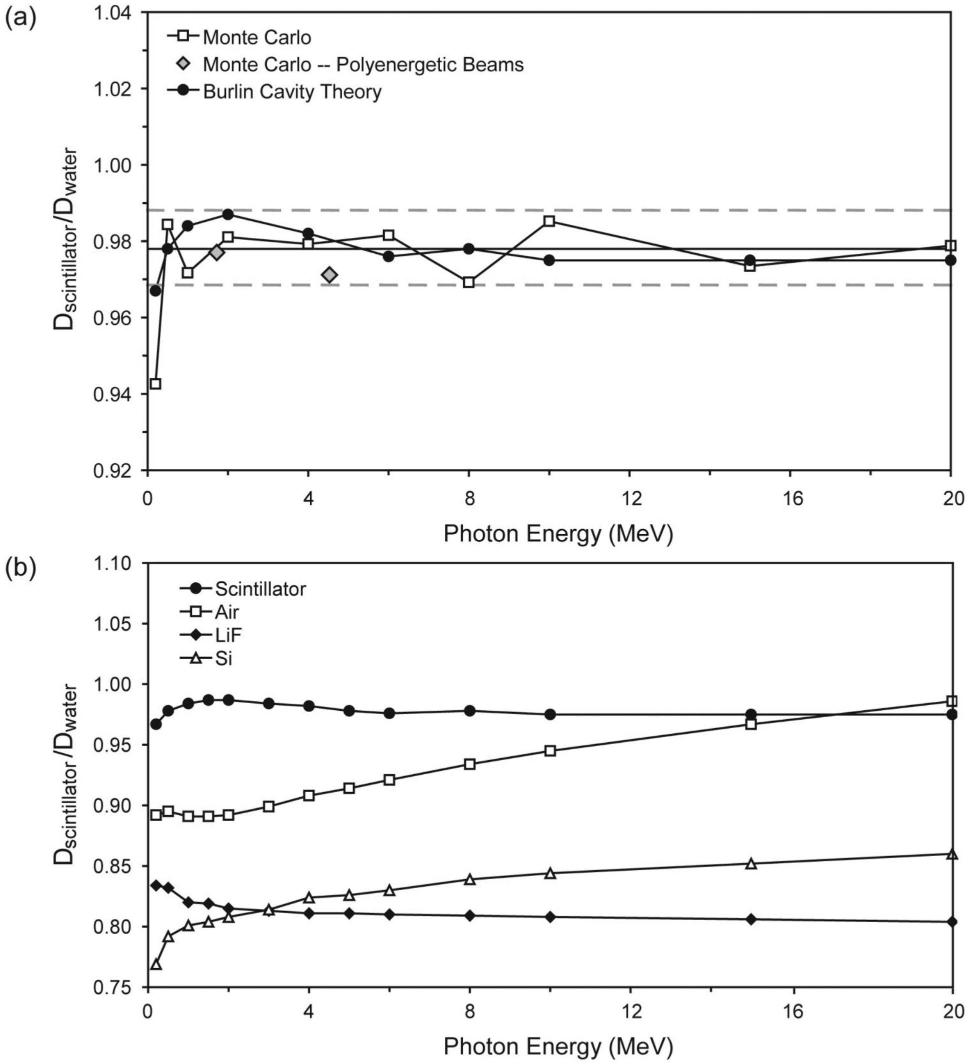

FIG. 2. (a) Ratio of absorbed dose in the plastic scintillator to that in water. Results for Monte Carlo calculations with a range of monoenergetic photon beams and 6 and 18 MV polyenergetic beams are shown. Calculations using Burlin cavity theory assumed a zero-thickness polystyrene wall. The solid line represents the average value of Monte Carlo calculations for monoenergetic beams, while the dashed lines are at ±1% of the average. (b) Results of Burlin cavity theory calculations for the plastic scintillator and equivalent volumes of air (for ion chambers) and LiF (for TLDs). Si (for diodes) was modeled using Bragg–Gray cavity theory.

then processed by the BEAMDP program[23] to obtain the energy and angular distributions of the photon fluence. These distributions generally showed planar variation and were therefore averaged over a $10 \times 10\ \mathrm{cm}^2$ square located at the SSD and centered on the beam central axis. This approach reduces statistical errors and simplifies further calculations. The distributions were then used as a source for the MCNPX calculations, using the same method as described earlier.

C. Burlin cavity theory

Burlin cavity theory was previously applied to compare the dose absorbed by the plastic scintillator to that absorbed by water,[1] and we briefly review the method here. The theory is appropriate for cavities of intermediate size and bridges the gap between small cavities, for which the Bragg–Gray theory applies, and large cavities, for which the influence of the cavity walls is negligible. Burlin cavity theory may be expressed as[1,24,25]

$$\frac{\overline{D}_{\mathrm{scin}}}{D_{\mathrm{water}}} = d_m(\overline{S})^{\mathrm{scin}}_{\mathrm{water}} + (1-d)\left(\frac{\overline{\mu}_{\mathrm{en}}}{\rho}\right)^{\mathrm{scin}}_{\mathrm{water}}, \tag{1}$$

where $\overline{D}_{\mathrm{scin}}$ is the average absorbed dose in the scintillator, D_{water} is the dose in charged-particle equilibrium at the same location in water, ${}_m(\overline{S})^{\mathrm{scin}}_{\mathrm{water}}$ is the mean ratio of mass collision stopping powers for the plastic scintillator and water, $(\overline{\mu}_{\mathrm{en}}/\rho)^{\mathrm{scin}}_{\mathrm{water}}$ is the mean ratio of the mass energy absorption coefficients, and d accounts for the attenuation of electrons entering the cavity and the build-up of electrons generated inside the cavity. For diffuse fields of electrons whose fluence is approximately the same in both materials, d is given by

$$d = (1 - e^{-\beta L})/\beta L, \tag{2}$$

where β is the effective absorption coefficient of the electrons in the cavity and L is the mean path length of the electrons across the cavity. For small cavities, the value of d approaches unity, whereas for large cavities, it approaches zero. The value of d for the plastic scintillator was found to

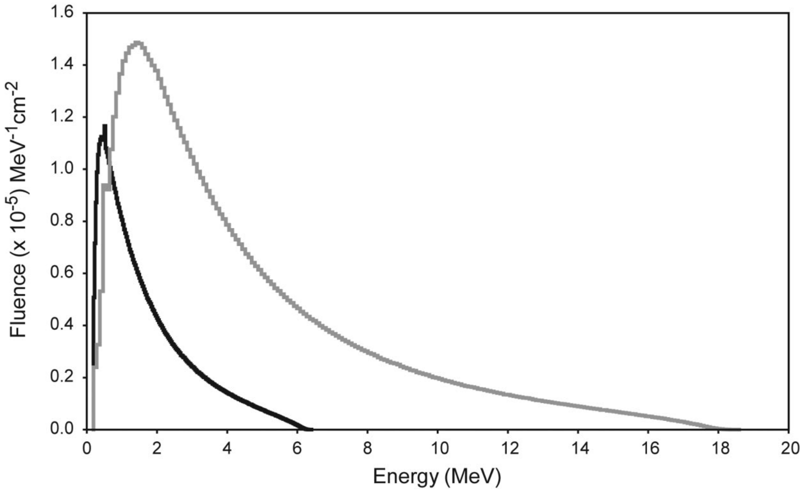

FIG. 3. Beam fluence as a function of energy for the 6 MV (black) and 18 MV (gray) polyenergetic photon beams.

be nearly zero for low photon energies and close to one for energies above 10 MeV. Thus, the large cavity approximation is justified for orthovoltage energies, while the small cavity approximation is applicable to high energies. In the range of most radiotherapy beams, d was found to lie in the intermediate region, justifying the use of Burlin cavity theory.

III. RESULTS AND DISCUSSION

The ratio of absorbed dose in the scintillator to that in water is shown in Fig. 2(a). Looking first at the results for the monoenergetic photon beams, we can see that for energies of 0.5 MeV and above, the ratio is nearly a constant, and the value is close to 0.98 throughout the megavoltage radiotherapeutic energy range for experimental conditions close to that selected in this work. At 0.2 MeV, the ratio drops to about 0.94, and as shown previously,[11,12] the plastic scintillator begins to show strong energy dependence at lower energies.

Beam fluence as a function of photon energy for the polyenergetic 6 and 18 MV photon beams is shown in Fig. 3, where the average photon energies were 1.7 and 4.5 MeV, respectively. Looking back at Fig. 2(a), we can see that the results of Monte Carlo calculations for the polyenergetic beams are similar to those obtained for the monoenergetic beams. It is clear that the plastic scintillator is both water equivalent and energy independent in the megavoltage energy range. Thus, a plastic scintillation detector could be calibrated in a ^{60}Co teletherapy beam, for instance, and then used without any correction factors to measure a polyenergetic beam produced by a high-energy accelerator for experimental conditions close to those selected in this work.

The results of Burlin cavity theory are in close agreement with those obtained from the Monte Carlo calculations [Fig. 2(a)]. This demonstrates that the assumption of an intermediate-sized cavity is appropriate. Furthermore, because the polystyrene probe has radiological properties similar to those of both water and the plastic scintillator, the zero-thickness wall approximation is valid. The results also suggest that Burlin cavity theory could be used to validate new detectors of a similar size to the one studied, unless conditions necessitate the use of Monte Carlo calculations (for example, use of a detector made from high-Z material). Considering the consistency in energy response, the scintillation detector can be expected to outperform many commonly used detectors, including ion chambers, thermoluminescent devices (TLDs), and silicon diodes. This is demonstrated in Fig. 2(b), where Burlin cavity theory was used for the detector materials air (ion chambers), LiF (TLDs), and Si (diodes).[1] Typical Si diodes have effective thicknesses of the order of 50 μm, and therefore the Burlin parameter d was taken as unity, which reduces to Bragg–Gray theory. Whereas the plastic scintillator is energy independent in the megavoltage energy range, the other detectors show somewhat stronger energy dependence.

Aside from energy independence, the plastic scintillation detector has many attractive dosimetric characteristics, including reproducibility, linear response to dose, temperature independence, and resistance to radiation damage.[1–4,6,26] The detector can accurately measure the depth dose profiles for either x rays or electron beams in the megavoltage energy range. Its spatial resolution is superior to many traditional detectors, including ionization chambers, radiographic film, and silicon diodes.[2,7] One important issue still facing the detector is its signal-to-noise ratio, particularly when exposed to electron beams, where Cerenkov radiation cannot be ignored. It is hoped that improvement in signal collection efficiency and reduction in background will make these detectors more widely utilized for clinical applications.[27–30]

IV. CONCLUSIONS

Monte Carlo calculations show the plastic scintillator to be energy independent and water equivalent in the energy range of megavoltage radiotherapeutic photon beams. A plastic scintillation detector could thus be calibrated at an energy above 1 MeV and then be used over the megavoltage energy range without any energy correction. The good agreement between Monte Carlo and Burlin cavity theory calculations validates the approximations used in this application to Burlin cavity theory. Monte Carlo calculations can be a useful tool to determine the energy dependence of new materials that could potentially be used as detecting media for radiation detectors.

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