Dosimetric verification of the anisotropic analytical algorithm in lung equivalent heterogeneities with and without bone equivalent heterogeneities

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Purpose: In this study, the authors evaluated the accuracy of dose calculations performed by the convolution/superposition based anisotropic analytical algorithm (AAA) in lung equivalent heterogeneities with and without bone equivalent heterogeneities.

Methods: Calculations of PDDs using the AAA and Monte Carlo simulations (MCNP4C) were compared to ionization chamber measurements with a heterogeneous phantom consisting of lung equivalent and bone equivalent materials. Both 6 and 10 MV photon beams of 4×4 and 10×10 cm² field sizes were used for the simulations. Furthermore, changes of energy spectrum with depth for the heterogeneous phantom using MCNP were calculated.

Results: The ionization chamber measurements and MCNP calculations in a lung equivalent phantom were in good agreement, having an average deviation of only $0.64 \pm 0.45\%$. For both 6 and 10 MV beams, the average deviation was less than 2% for the 4×4 and 10×10 cm² fields in the water-lung equivalent phantom and the 4×4 cm² field in the water-lung-bone equivalent phantom. Maximum deviations for the 10×10 cm² field in the lung equivalent phantom before and after the bone slab were 5.0% and 4.1%, respectively. The Monte Carlo simulation demonstrated an increase of the low-energy photon component in these regions, more for the 10×10 cm² field compared to the 4×4 cm² field.

Conclusions: The low-energy photon by Monte Carlo simulation component increases sharply in larger fields when there is a significant presence of bone equivalent heterogeneities. This leads to great changes in the build-up and build-down at the interfaces of different density materials. The AAA calculation modeling of the effect is not deemed to be sufficiently accurate. © 2010 American Association of Physicists in Medicine. [DOI: [10.1118/1.3464748](https://doi.org/10.1118/1.3464748)]

Key words: dosimetric verification, AAA dose calculation algorithm, Monte Carlo, MCNP, convolution-superposition algorithm, heterogeneity corrections

I. INTRODUCTION

Intensity-modulated radiation therapy requires a high degree of dose calculation accuracy for radiation treatment planning systems, especially in the presence of heterogeneities.¹⁻³ However, predictions of correction-based dose calculation algorithms may deviate greatly from the measurement in heterogeneous media or their proximity.⁴⁻⁹ In these algorithms, dose distributions are typically calculated by using density

scaling to correct for dose changes in the presence of heterogeneities. Patient CT images supply the tissue density for the material required for this scaling.¹⁰ On the other hand, dose contribution due to scattered photons dose information is supplied by the dose kernel derived by Monte Carlo (MC) simulation photon interactions in water. One source of large calculation uncertainty in the presence of heterogeneities is due to the invariance of the dose kernel. The dose kernel of

correction-based algorithms does not account for spectral changes depending on the distance from the interaction point to the energy absorption point.

It is not practical to utilize the MC algorithms in clinically realistic situations due to long computation times consuming calculation.¹¹ For this reason, convolution/superposition algorithms have been preferred in many commercial radiation treatment planning systems. In these convolution/superposition algorithms, dose calculations are performed in heterogeneous media by transforming the energy deposition kernel using the density scaling derived from the electron density distributions obtained from CT images.

Many studies have evaluated the accuracy of dose calculations using convolution/superposition algorithms in low-density media similar to lung. One report states a dose calculation difference to measured value less than 2% in the presence of low-density heterogeneities.⁶ However, convolution/superposition algorithm dose predictions may be less accurate in high-density regions composed of high atomic number materials.⁸

The anisotropic analytical algorithm (AAA), a convolution/superposition algorithm, is the most recent photon dose calculation algorithm implemented in the commercial Eclipse treatment planning system (Varian Medical Systems, Palo Alto, CA). Several studies have evaluated the accuracy of dose calculations using the AAA in the presence of heterogeneities.^{12–16} In this study, we evaluated the accuracy of the AAA dose calculations in lung equivalent materials with and without the presence of bone equivalent heterogeneities. Percentage depth dose (PDD) curves predicted by the AAA and MC methods were compared to ionization chamber measurements. The relationship between energy spectrum changes in depth and dose calculation uncertainties were examined for the heterogeneous phantom using MC calculation.

II. MATERIAL AND METHODS

II.A. AAA

The AAA is a novel photon dose calculation algorithm implemented in the Varian Eclipse TPS. It is a 3D pencil beam convolution-superposition algorithm based on a triple-Gaussian photon kernel model.^{17–20} In this model, the radiation source is classified into three subsources: Primary photons, scattered extrafocal photons, and scattered electrons from the beam limiting devices, which are used to calculate the dose by EGSnrc MC.²¹ A Varian Clinac 2300 C/D (Varian Medical Systems, Palo Alto, CA) is used as a reference accelerator. The AAA accounts for heterogeneities by correction of the photon scatter kernel according to the three-dimensional anisotropic changes in electron density in the neighborhood of an interaction point. The broad clinical beam was divided into narrow beamlets along a fan line. Convolution is first performed on one beamlet for each of the three subsources and then the final dose distribution is obtained via superposition of the dose contribution of all the individual beamlets. Near interfaces between different media, build-up and build-down corrections are calculated using

TABLE I. Energies and FWHMs simulated for the initial electron-beam parameters.

Nominal photon energy (MeV)	Mean energy (MeV)	Energy FWHM (%)	Radial FWHM (mm)
6	6.0	3.0	1.3
10	10.25	3.0	1.5

a separate kernel that is a dual exponential function describing changes in the electronic equilibrium across the interfaces of different heterogeneities and water. In this study, dose calculations were performed with the AAA using a 2.5 mm calculation grid in the presence of heterogeneities.

II.B. MC simulation

MCNP code (Monte Carlo N-particle transport code system, version 4C, Los Alamos National Laboratory, Los Alamos, NM) was used for simulations in this study. It is able to simulate physical interactions of photons, electrons, and neutrons with material.^{22–24} A typical configuration of a Clinac 2100C accelerator head (Varian Medical Systems, Palo Alto, CA) was modeled faithfully based on a detailed configuration diagram provided by the vendor. The calculation matrix for heterogeneous phantoms are divided into voxels of size $2 \times 2 \times 0.25$ (the depth of the build-up region is 0–2 cm for 6 MV and 0–3 cm for 10 MV, respectively) or $2 \times 2 \times 0.5$ cm³ (outside the build-up region) along the central beam axis when calculating PDDs.

Initial incident electron-beam parameters were determined by performing commissioning prior to this study. Table I lists the initial incident electron-beam parameters used in MCNP calculations. Differences compared to ionization chamber measurements was less than 2% for PDDs along the central beam axis and beam profiles in broad beams of various field sizes (5×5 , 10×10 , 20×20 , and 40×40 cm²) in water. PDDs and beam profiles are influenced by the mean energy, energy spread, and radial intensity of incident electrons. It has been reported that calculation accuracy is improved by modeling energy spread and radial intensity as a Gaussian distribution.^{25–30} The full width at half maximum (FWHM) of the Gaussian distribution was adjusted such that PDDs and profiles by MCNP calculations corresponded to ionization chamber measurements in water for this study. To reduce the calculation times, variance reduction techniques were employed. The cutoff energy for electrons and photons were set to 0.7 and 0.01 MeV, respectively, so that the weights of particle numbers in a phantom would be maximized for photons as well as electrons (weight window). The number of incident electrons was set to 3×10^8 and 2×10^8 for 4×4 and 10×10 cm² fields at 6 MV, respectively, while the number was set to 2×10^8 and 1.5×10^8 for 4×4 and 10×10 cm² fields at 10 MV, respectively. These incident numbers of electrons reduced the statistical uncertainty of the calculation results for our area of interest under 1%. An Intel Core™ 2 Duo 2 GHz machine was used for calculation and computation speed

FIG. 1. Diagram of the heterogeneous layer phantoms in the PDD study: Water equivalent slabs with (a) 5 cm lung equivalent slab with 15 cm inserted water-equivalent slabs and (b) with 5 and 6 cm lung equivalent slabs inserted, with a 4 cm bone equivalent slab inserted.

allowed processing of particle histories at a rate of 8.06×10^4 particles/min.

II.C. PDD curves

As shown in Fig. 1, two phantoms were constructed from slabs of water-equivalent (Solid Water, Gammex Inc., Middleton, WI), lung equivalent (Tough Lung, Kyoto Kagaku, Co., Ltd., Kyoto, Japan), and bone equivalent material (Tough Bone, Kyoto Kagaku, Co., Ltd., Kyoto, Japan). Electron density ratios (ρ_{ew}) to water for these media were 1.046, 0.32, and 1.73, respectively. PDDs were calculated using the Varian Eclipse TPS and AAA with a source-surface distance of 90 cm for two field sizes (4×4 and 10×10 cm²) and two photon beam energies (6 and 10 MV). Comparison of these PDD predictions with MCNP simulation was the basis for evaluation of the AAA predictive accuracy. Measurements of PDD made with a 0.6 cm³ PTW Farmer ionization chamber (PTW-Freiburg, Freiburg, Germany) were used in turn to evaluate the accuracy of the MCNP calculations. The AAPM TG-51 and IAEA TRS-398 protocols for dosimetry of absorbed dose only provide conversion from ionization to absorbed dose at a reference depth in water.^{31,32} Therefore, correction of stopping power ratio for water-to-heterogeneous medium is necessary for conversion to absorbed dose in heterogeneities. Siebers *et al.*³³ reported that the stopping power ratios $s_{w,med}$ of a lung equivalent phantom to water obtained by MC calculation were 0.999 and 0.992 at 6 and 10 MV, respectively, with a very small measurement uncertainty. However, a measured difference of more than 10% was observed for a bone equivalent phantom.³³ For this reason, this study restricted measurements with the ionization chamber to water and a lung equivalent media.

II.D. Energy spectrum

MCNP simulations of the energy spectrum in the beam's vertical plane were performed for the same energies and irradiation fields used in PDD calculations. These simulations enabled the evaluation of changes in energy spectrum with and without bone equivalent media present in the composite

lung equivalent and bone equivalent phantoms (Fig. 1). Energy bin size was 0.05 MeV and calculations were performed in the depth on planes at 2.5 and 5 mm intervals that corresponded to those used for the PDD calculations. The low-energy component below 1.0 MeV was integrated to evaluate the changes of the low-energy component in the depth of the heterogeneous slab phantoms.

III. RESULTS

III.A. PDD curves

Figures 2 and 3 compare PDD curves predicted by the AAA and MCNP simulation to ionization chamber measurement in the two heterogeneous slab phantoms depicted in Figs. 1(a) and 1(b), respectively. The comparison of PDDs are for the field sizes 4×4 [Figs. 2(a) and 2(c)] and 10×10 cm² [Figs. 2(b) and 2(d)] for beam energies of 6 [Figs. 2(a) and 2(b)] and 10 MV [Figs. 2(c) and 2(d)]. Figures 2 and 3 show MCNP simulations demonstrated that build-down was evident on the downstream interfaces of water-lung and bone-lung slabs for the 4×4 cm² fields, but not evident for the 10×10 cm² fields. Build-up was evident at the interface between a lung-bone slab and the interface between a lung-water slab for the 4×4 cm² field; it is not evident for the 10×10 cm² field.

PDDs calculated by the AAA and MCNP mostly showed good agreements with deviations of less than 3% for the 4×4 and 10×10 cm² field sizes for both 6 and 10 MV beams in phantom (a) of Fig. 1. A similar agreement was also apparent for the 4×4 cm² field in phantom (b) of Fig. 1. However, discrepancies in the 10×10 cm² field for phantom (b) of Fig. 1 were clear. Table II lists average dose deviations and standard deviations of the AAA calculations compared to MCNP calculations in the lung equivalent slab of phantom (a) in Fig. 1 and the bone equivalent and lung equivalent phantom of before and after the bone insert in phantom (b) in Fig. 1. For all energies and fields in phantom (a) of Fig. 1 and for the 4×4 cm² field in phantom (b) of Fig. 1, agreement was good with an average deviation of less than 2%, while for a 10×10 cm² field, the average deviation was more than 3% in the lung equivalent material before and after a bone equivalent slab. Maximum deviations for a 10×10 cm² field were 5.0% and 4.1% for 6 and 10 MV photon beams, respectively. The ionization chamber measurements and MCNP calculations were in good agreement in every case and the average deviation was $0.64 \pm 0.45\%$.

III.B. Energy spectrum

Energy spectra were obtained by MCNP simulation in the lung equivalent and bone equivalent slab phantoms of Fig. 1. Figure 4 shows the energy spectra in the low-energy region at depth of 5 mm on the source proximal the lung/bone interfaces and 5 mm distal the bone/lung interfaces of this phantom for 6 [Figs. 4(a) and 4(c)] and 10 MV photon beams [Figs. 4(b) and 4(d)]. The energy spectra were compared to the bone slab present and without at equivalent depth in 4×4 and 10×10 cm² fields. Any differences represent

FIG. 2. PDDs comparison in the heterogeneous layer phantom with water-equivalent and lung equivalent materials in Fig. 1(a), predicted by AAA (solid curve), calculated by MCNP (open circles), and measured with cylindrical ionization chamber (closed circles). (a) 6 MV in 4×4 cm² field size, (b) 6 MV in 10×10 cm² field size, (c) 10 MV in 4×4 cm² field size, and (d) 10 MV photon beams in 10×10 cm² field size.

backscatter radiation from the bone slab. Energy spectra did not change regardless of the presence or absence of the bone slab for the energy region above 0.5 MeV, but the lower energy component increased significantly in the presence of a bone slab. This difference was greater for a 10×10 cm² field.

Figure 5 shows the variation in an integral low-energy component below 1.0 MeV of the spectrum in depth for the heterogeneous slab phantoms described in Fig. 1 for the 6 (a) and 10 MV (b) photon beams. Under all conditions, the low-energy component showed an overall moderate depth-dependent decrease. The low-energy component increased more when a bone equivalent slab was present, and this increase was greater in a 10×10 cm² field than in a 4×4 cm² field.

IV. DISCUSSION

MC calculation of dose distributions require accurate modeling of the linac head to set physical phase space pa-

rameters for the electrons incident upon the target to guarantee the accuracy. The ionization chamber measurements and MCNP calculations performed in the lung equivalent phantoms used in this study agreed very well (average deviation of $0.64 \pm 0.45\%$), supporting the premise that the MCNP calculation is an appropriate tool for validation of the AAA calculations on the PDDs.

As shown in Figs. 2 and 3, build-down was seen on the source distal interfaces of the water-lung and bone-lung slab phantoms calculated by MCNP in a 4×4 cm² field for 6 and 10 MV photon beams. Build-down was more significant at 10 than 6 MV, but not apparent in a 10×10 cm² field. In other studies, prominent build-down was seen for higher energies and smaller fields.⁴ One of the causes for this phenomenon may be the lack of lateral electron equilibrium in a lung equivalent phantom. In other words, we speculate that build-down is a function of both the decreased cross section of photon interaction and the transmission factor for primary radiation in a lung equivalent phantom at high energies and

FIG. 3. PDDs comparison in the heterogeneous layer phantom with water-equivalent, lung equivalent, and bone equivalent materials in Fig. 1(b), predicted by AAA (solid curve), calculated by MCNP (open circles), and measured with cylindrical ionization chamber (closed circles). (a) 6 MV in 4×4 cm² field size, (b) 6 MV in 10×10 cm² field size, (c) 10 MV in 4×4 cm² field size, and (d) 10 MV photon beams in 10×10 cm² field size.

in small fields.⁴ A small build-up was found on the source distal region on the interface of the lung equivalent and bone equivalent phantoms at both photon beam energies, but was more prominent in a 10×10 cm² field than a 4×4 cm² field. Since the cross section for the Compton effect increases with the density of the medium, and the scattered angle of the generated electrons is larger in bone than

in lung material, the formation of rebuild-up is observed. According to Table II, the AAA calculations of 4×4 and 10×10 cm² fields for the 6 and 10 MV in the lung equivalent phantoms without a bone slab and a 4×4 cm² field with a bone slab were in good agreement (average deviation of less than 2%) compared to MCNP calculations. In contrast, a discrepancy was observed in a 10×10 cm² field for the lung

TABLE II. Results of the average dose deviations and standard deviations (1SD) between AAA calculations and MCNP calculations for each particular combination of energy, field size, and heterogeneous phantom configuration in Fig. 1.

$\Delta D(\%)$ (AAA/MCNP)	Field size (cm ²)	Phantom (a): Lung	Lung (before)	Phantom (b): Bone	Lung (after)
6 MV	4×4	0.4±0.5	0.1±0.7	2.0±0.5	-1.6±0.4
	10×10	-1.8±0.6	-2.3±0.6	2.4±0.5	-3.5±1.6
	4×4	1.5±0.7	1.6±0.8	-0.1±0.4	-0.3±0.3
10 MV	10×10	-0.7±0.5	-2.3±0.6	0.3±0.5	-3.2±0.9

FIG. 4. Energy spectrum at depth of 5 mm [(a) and (b)] proximal and [(b) and (d)] distal; the interface of lung and bone material phantom for low energy. [(a) and (c)] 6 MV and [(b) and (d)] 10 MV photon beams for 10×10 (bold curve) and 4×4 cm² field sizes (thin curve), with bone equivalent materials in the heterogeneous phantom layer (solid curve) and without bone equivalent materials (dotted curve).

equivalent phantom before and after a bone slab. Maximum deviations were 5.0% and 4.1% in the 10×10 cm² field for 6 and 10 MV photon beams, respectively.

MCNP calculation of the energy spectrum was performed to examine the underlying cause for these differences. Figure 4 shows the changes in energy spectrum at depth of 5 mm on the source proximal and distal the interface of the lung-bone phantom. The energy region lower than 0.5 MeV was increased by the presence bone. In other data, Fig. 5 confirmed these changes in the low-energy component along depth. Low-energy scattered radiation significantly increased in the bone slab for a 10×10 cm² field compared to a 4×4 cm² field.

In contrast to the convolution/superposition method, MC dose calculation in the presence of a heterogeneity is highly accurate. For example, Spiridovich *et al.*⁹ reported that dose calculated with the convolution/superposition method differed by more than 10% as compared to MC calculations in

a region downstream of a high-density tooth and this difference was even larger in presence of metal fillings with high atomic number.

Corrections can be made for build-up and build-down between different media using the AAA. This correction method is unique to AAA and does not work for other convolution/superposition algorithms. In AAA, kernel corrections are applied between different media using build-up kernel corrections between different heterogeneities and water. Tillikainen *et al.*²¹ reported that an overestimation of the dose calculation by the AAA near the interface between lung equivalent and water-equivalent media resulted by an overestimation of build-up kernel corrections.

In this study, a bone slab was inserted into the lung equivalent phantoms, increasing density difference. Low-energy scattered radiation increased in the bone slab especially in a 10×10 cm² field, and the energy spectra in the bone slab varied from the lung. It is argued that an uncertainty is introduced to the kernel adjustment for build-up or

FIG. 5. Relationship between the low-energy scattered photons below 1.0 MeV and several depth for two field sizes (4×4 and 10×10 cm²) by MCNP in the heterogeneous phantom layer. (a) 6 and (b) 10 MV photon beams with bone equivalent materials in the heterogeneous phantom layer (closed circles) and without bone materials (open circles).

build-down corrections between different media when using the AAA to calculate dose in the presence of heterogeneity. This may result in an underestimate of dose in a lung phantom before and after a bone slab.

V. CONCLUSIONS

The accuracy of PDDs calculated using the AAA in a heterogeneous region including lung and bone was verified by MCNP MC calculation. The AAA predictions of PDD dose varied by more than 4% from the MCNP results before and after a bone slab in a 10×10 cm² field. Energy spectrum analysis indicated that a source of the discrepancy was the low-energy scatter increase. This spectral change introduces an uncertainty in the corrections for build-up and build-down for the AAA resulting in a systematic error. This systematic error is greater in larger fields and when there is a significant difference in densities.

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