Modification and validation of an analytical source model for external beam radiotherapy Monte Carlo dose calculations

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Purpose: A dose calculation tool, which combines the accuracy of the dose planning method (DPM) Monte Carlo code and the versatility of a practical analytical multisource model, which was previously reported has been improved and validated for the Varian 6 and 10 MV linear accelerators (linacs). The calculation tool can be used to calculate doses in advanced clinical application studies. One shortcoming of current clinical trials that report dose from patient plans is the lack of a standardized dose calculation methodology. Because commercial treatment planning systems (TPSs) have their own dose calculation algorithms and the clinical trial participant who uses these systems is responsible for commissioning the beam model, variation exists in the reported calculated dose distributions. Today's modern linac is manufactured to tight specifications so that variability within a linac model is quite low. The expectation is that a single dose calculation tool for a specific linac model can be used to accurately recalculate dose from patient plans that have been submitted to the clinical trial community from any institution. The calculation tool would provide for a more meaningful outcome analysis.

Methods: The analytical source model was described by a primary point source, a secondary extra-focal source, and a contaminant electron source. Off-axis energy softening and fluence effects were also included. The additions of hyperbolic functions have been incorporated into the model to correct for the changes in output and in electron contamination with field size. A multileaf collimator (MLC) model is included to facilitate phantom and patient dose calculations. An offset to the MLC leaf positions was used to correct for the rudimentary assumed primary point source.

Results: Dose calculations of the depth dose and profiles for field sizes 4×4 to 40×40 cm agree with measurement within 2% of the maximum dose or 2 mm distance to agreement (DTA) for 95% of the data points tested. The model was capable of predicting the depth of the maximum dose within 1 mm. Anthropomorphic phantom benchmark testing of modulated and patterned MLCs treatment plans showed agreement to measurement within 3% in target regions using thermoluminescent dosimeters (TLD). Using radiochromic film normalized to TLD, a gamma criteria of 3% of maximum dose and 2 mm DTA was applied with a pass rate of least 85% in the high dose, high gradient, and low dose regions. Finally, recalculations of patient plans using DPM showed good agreement relative to a commercial TPS when comparing dose volume histograms and 2D dose distributions.

Conclusions: A unique analytical source model coupled to the dose planning method Monte Carlo dose calculation code has been modified and validated using basic beam data and anthropomorphic phantom measurement. While this tool can be applied in general use for a particular linac model, specifically it was developed to provide a singular methodology to independently assess treatment

plan dose distributions from those clinical institutions participating in National Cancer Institute trials.

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Key words: Monte Carlo, source model, clinical trial

1. INTRODUCTION

The Imaging and Radiation Oncology Core Houston Quality Assurance (QA) Center (IROC Houston) is a clinical trial quality assurance (QA) organization in support of the National Cancer Institute's National Clinical Trials Network. The mission of IROC Houston is to ensure that the data from radiotherapy clinical trials are of high quality. To this end, IROC Houston has several programs to efficiently provide dosimetric and QA services including the evaluation of advanced technology treatments with anthropomorphic QA phantoms. This audit enables comparison between measurement of the actual delivered dose distribution with film and thermoluminescent detectors (TLDs) and the institutions' planned treatment.

Although measurement-based comparisons remain the gold standard in determining an institution's performance in terms of correct dose delivery, there has been growing concern that advances in treatment delivery techniques have pushed the limits of measurement uncertainty.¹ In part, this is due to the common occurrence of steep dose gradients. In addition to the uncertainties associated with TLDs and radiochromic film, IROC Houston has published the results from their remote anthropomorphic phantom audit program detailing the ability of institutions to deliver the intended dose. The results show a varying degree of compliance even for treatment plans from the same treatment planning system (TPS).²⁻⁶ The TPS dose calculation algorithm, the beam model commissioning process, and delivery may individually or all contribute to these discrepancies. These differences have caused concern over an institution's ability to consistently deliver IMRT treatments and/or perform consistent heterogeneous dose calculations for patients who entered clinical trials. While end-to-end phantom measurements are robust at identifying errors, the nature of an end-to-end test makes it difficult to identify specific causes for discrepancies. Therefore, there is a quality assurance benefit to calculating patient dose distributions directly and independently without inferring through measurement-based means.

The use of a trusted independent dose calculation tool, such as the Monte Carlo (MC) technique, can help to improve the understanding of the dose delivered by bridging the gap between the actual dosimetry and TPS's predicted dose distributions. Such a tool that is capable of being applied to a majority of patient plans across many institutions and linear accelerator (linac) platforms can aid in the reduction of dose uncertainty by providing a centralized dose calculation method for treatment verification and retrospective dose analysis, thereby improving the quality assurance of clinical trial reporting. A Monte Carlo method that is similar to that found in conventional dose calculation algorithms and described in detail in Medical Physics Task Group Report

105 (Ref. 7) relies on fluence and energy distributions derived from analytical models that are subsequently used as source inputs to the MC calculations.⁸⁻¹⁰ The analytical models are measurement based and the model parameters are optimized to minimize differences between a standard set of beam measurement data and the MC calculation. This approach does not rely on phase space data or linac head geometry definitions found in traditional methods, and may simplify its use while maintaining an acceptable level of accuracy.

In this work, we describe the modification, implementation, and initial results of a reference MC recalculation system for evaluation of IROC Houston anthropomorphic phantom calculations. Specifically, a parameterized multisource model was derived from a standard set of measurements for the Varian 2100 series linac (Varian Medical Systems, Inc., Palo Alto, CA) 6 and 10 MV photon beams.^{11,16,17} The source model coupled to the dose planning method (DPM) Monte Carlo method¹² provided a unique and accurate solution for addressing the problem of multiple institutions reporting dose distributions using various TPSs. Further, the analytical model approach provides a generic way to extend the model to other linacs regardless of manufacturer or model type.

2. MATERIAL AND METHODS

2.A. Source model

The analytical model was developed and doses were calculated using DPM. The DPM Monte Carlo algorithm was largely chosen because of its open source code allowing for flexibility in coupling and modifying the dose calculation tool. DPM uses the standard condensed history model for electron transport. This model handles large and small energy transfer collisions by direct sampling of the distance to the next collision using the total electron interaction cross section and the continuous slowing down approximation (CSDA), respectively. Photon transport for photoelectric absorption, Compton scatter, and pair production is handled interaction-by-interaction. The transport mechanics were reformulated to increase the computing speed by enabling large electron transport steps even across heterogeneous boundaries while adhering to the applied multiple scattering distribution theory.¹²

Two distinct models were created: a 6 and a 10 MV Varian linear accelerator; each was separately commissioned and validated. Each simulation of open fields, phantom plans, or patient plans, used 10×10^6 particles per square centimeter. The calculations for the 6 and 10 MV beams applied low energy electron and photon cutoffs of 200 and 50 keV, respectively. The voxel resolution of the calculated data was $0.2 \times 0.2 \times 0.3$ cm.

The measurement-driven parameterized source model consists of three sources expressed by analytical functions. The

general form of these three analytic functions was taken from the available literature: a primary isotropic point source; an extra-focal disk source;¹³ and an electron contamination uniform disk source.¹⁴ The extra-focal disk was spatially uniform and the energy distribution was scaled down from the primary source. The electron contamination disk was also spatially uniform. The off-axis energy¹⁵ effects due to the flattening filter were accounted for by modifying the initial simulation of the energy and direction of each primary photon.

The horn-effect, which describes the fluence increase of the primary photons as the off-axis angle increases due to the flattening filter, was modeled by a piecewise linear function. The multileaf collimator (MLC) leaf positions for modulated fields were imported from the TPS. A plan specific composite fluence map from the segmented fields was generated. Leaf transmission, interleaf leakage, and rounded leaf ends were modeled as part of the fluence map.^{11,16,17}

2.B. Commissioning

The initial step of assigning appropriate parameters to the three analytic sources is the commissioning process. All fitting parameters were determined at once using an automated optimization process for the specific linac model and linac energy based on the 10×10 cm field size by setting parameter boundary conditions and minimizing the differences between the measured and calculated percent depth dose (PDD) curve and dose profiles.^{11,16,17} Three parameters formed the shape of the primary energy spectrum function.¹¹ One parameter reduced the scale of the primary spectrum to construct the extra-focal energy spectrum. Ratio parameters were fitted to express the fluence contributions from the extra focal photons and the contaminant electrons relative to the primary photons. Since the penumbra of the dose profiles measured with an ion-chamber is influenced by partial volume effects related to the chamber diameter, a seventh parameter describing the chamber diameter was used in a convolution function to modify the calculated dose profiles so that comparisons with dose profiles produced from ion-chamber measurements could be made. Prior to determining model parameters via the optimization process, precalculated dose data sets were provided with each set computed from a narrow energy bin width of 0.25 MeV over an appropriate energy range for primary photons and extra focal photons. As part of the commissioning process, the energy bins were optimized to produce a weighted energy spectrum. A dose data set from electron contamination was also computed. The model parameters were determined from optimally weighted energy bins that produced reasonable spectra and fluence contributions to satisfy the agreement between the calculation and measurement of the basic beam data for a single field size of 10×10 cm. With the model parameters known, a piecewise linear function was used to model the increase in fluence of the primary photons as a function of off-axis angle. The increase in the fluence effect is commonly referred to as the “horn-effect” and is due to the preferential attenuation of the conical-like shaped flattening filter. In this step, 1600 beamlet calculations of 1×1 cm

size that comprised a 40×40 cm field size were calculated on the virtual 3D water phantom. The beamlet weightings were adjusted by adjusting coefficients of the piecewise linear function until an accurate dose profile at a depth of the maximum dose ($d_{\max}$) was obtained. A final conversion factor was applied to convert the Monte Carlo simulation output into absolute dose per monitor unit. To avoid the statistical uncertainties of a single normalization point, the conversion factor was determined by normalizing to an integral dose defined by the area under the curve from the percent depth dose (PDD) between the depths of 5–15 cm for the 10×10 cm open field.¹⁸ The conversion factor was applied to every dose calculation performed for the specific commissioned beam model.

Cui *et al.* has shown that repeating the commissioning process by tightening the parameter boundary conditions in an iterative fashion may improve the accuracy of the dose calculation.^{16,17} However this was found to be inefficient and did not yield satisfactory results. Instead, the initial commissioned model used in this work was improved by applying a hyperbolic function to correct for the difference between the calculated and measured output based on field size.¹⁹ A similar correction was also used for electron contamination in the Varian 10 MV model because the model assumed a fixed contribution from contaminant electrons. The Varian 6 MV model did not include electron contamination because its contribution was only 0.2% for the 10×10 cm field size.

An assumption of the source model is that the primary source is a point source. However, it has been reported that the source diameter size is generally on the order of 2 mm or less.^{20–24} Because the primary source has a finite dimension, penumbra broadening will occur at the MLC leaf edge due to the varying location of the initial primary photons within the finite-dimensioned source. To account for this, the MLCs were offset by a fixed amount.

2.C. Validation testing

The validation of the Varian 6 and 10 MV photon beams was performed by comparing calculated doses to measured doses for the basic beam dosimetry data in water. These data consisted of PDD and dose profiles at several depths for field sizes ranging from 4×4 to 40×40 cm². The profile depths included $d_{\max}$, 6, 12.5, and 22 cm for the 6 MV data and $d_{\max}$, 5, 10, and 20 cm for the 10 MV data. Because of the consistency in output performance of today's modern linac, the specific source models produced from this data are expected to be broadly applicable to Varian 2100 series linacs.^{23,25–35} Measurements were made using a small volume (0.04 cm³, 4 mm diameter) ion chamber. To account for the volume effects of the ion chamber, which tend to smear the dose and exaggerate the dose penumbra, a Gaussian convolution was applied to the calculated dose data for all profile comparisons.³⁶ The standard deviation of the Gaussian kernel was one of the seven parameters determined in the model optimization commissioning process.^{16,17}

The agreement of the calculated data with the measured data (PDDs and dose profiles) for field sizes from 4×4 to 40×40 cm² was tested using a gamma criterion³⁷ of $\pm 2\%$ of the maximum dose along the central axis and ± 2 mm distance to agreement (DTA). In addition, local dose differences were studied for PDD and dose profiles. The dose profile local dose differences were evaluated in the high dose regions ($>80\%$ of the central axis dose). The depth of dose maximum was

evaluated by quantifying the depth difference between the calculated $d_{\max}$ and measured $d_{\max}$.

2.D. Benchmark testing

The validated linac source models were benchmarked under a variety of conditions using IROC Houston anthropomorphic head and neck, and thorax, phantoms. The head and neck

FIG. 1. (a) The IROC Houston head and neck phantom, (b) The IROC Houston thorax phantom, (c) CT images of head and neck phantom with targets and critical structure contoured, and (d) CT images of thorax phantom with target and critical structures contoured.

phantom of Figs. 1(a) and 1(c), as described previously^{38,39} was filled with water and made of an acrylic shell that included an insert made of polystyrene. It includes a solid water primary and secondary target, and an acrylic spinal cord. The heterogeneous thorax phantom³⁸ [Figs. 1(b) and 1(d)] consisted of a high density polyvinyl chloride (PVC) housing, a nylon heart, a polybutylene terephthalate polyester (PBT) spinal cord, CIRS exhale lung material (Computerized Imaging Reference Systems, Inc., Norfolk, VA), a nylon target located in the medial-anterior section of the left lung, and water used to fill the remaining volume.

Presently, the DPM code allows up to five material definitions based on user input of the elemental composition, atomic number, and density of a specific material. Air, polystyrene, water, solid water, and acrylic were defined for the head and neck phantom while air, CIRS lung, water, nylon, and PVC were defined for the thorax phantom. PVC material properties were applied to the PBT spinal cord because they share a similar composition.

The head and neck phantom and thorax phantom housed TLD capsules (Radiation Detection Company, Gilroy, CA) and radiochromic film (Ashland Inc., Wilmington, DE). For the head and neck phantom, TLD capsules were located at the centers of the primary and secondary targets and within the critical structure to collect near-point absolute dose information. For the thorax phantom, TLD capsules were located in the center of the target and critical structures (heart and spinal cord). The films were located in close proximity to the primary target TLD capsules so that the films' intersection coincided with the center of the primary target and were normalized to the TLD dose. The films were located in all three major planes for the thorax phantom while the films for the head and neck phantom were positioned in the axial and sagittal planes.

Designed to be progressively more difficult, treatment plans were evaluated for homogeneity, heterogeneity, small fields, and highly modulated fields. The IROC's anthropomorphic IMRT homogeneous head and neck phantom was used to test a highly modulated delivery of nine coplanar beams [Figs. 1(a) and 1(c)]. The IROC's heterogeneous thorax phantom was used to test a nine beam SBRT lung plan and a five beam noncoplanar IMRT lung plan [Figs. 1(b) and 1(d)]. The IMRT head and neck plan and SBRT lung plan were designed using the credentialing guidelines and irradiation instructions employed by the Radiation Therapy Oncology Group (RTOG) to credential institutions for participation in specific advance technology protocols. Each plan was independently designed for each treatment energy and each plan was delivered three times to evaluate reproducibility.

The resolution of the DPM calculation was governed by the size of the CT voxel. For this work, the voxel size was $0.195 \times 0.195 \times 0.25$ cm for all treatment plans studied. The accuracy of the DPM MC calculation was determined by comparing point doses, dose profiles, and two-dimensional (2D) gamma maps to the measured data. Point dose comparisons were made between the calculated values and measured TLD values of the target and critical structure locations [the calculated point doses were reported from the small region of interest (ROI) of the contoured TLD powder from the CT scan]. The 2D

dose distributions between the calculation and measurement were evaluated in CERR (Computational Environment for Radiotherapy Research; Washington University, St. Louis, MO)⁴⁰ using a $\pm 3\%$ 2 mm gamma criterion.

Finally, proof of principle studies were performed by recalculating patient plans with DPM and comparing to the original plans computed with the Pinnacle TPS superposition convolution algorithm. The purpose was to compare a commercial treatment planning system's dose calculation algorithm against the validated and benchmarked independent dose calculation tool in an effort to establish a baseline of what might be expected from a retrospective calculation study as part of the clinical trial outcomes analysis. The cases were selected from past patient treatments that were planned with Pinnacle using IMRT and SBRT. The test cases included treatments of the prostate, abdomen, and lung. Specifically, the recalculated treatments using the source model DPM calculation were: IMRT abdomen, 6 MV, 9 beams, 134 segments; IMRT prostate, 10 MV, 9 beams, 93 segments; SBRT lung, 6 MV, 7 beams, equivalent field size from 5.2 to 5.8 cm²; and IMRT lung, 10 MV, 6 beams, 82 segments. Comparisons for each

Fig. 2. (a) Energy spectrum comparison between spectrum calculated with the BEAM code and the spectrum computed from the analytical model for the 6 MV photon beam. (b) Energy spectrum comparison between spectrum calculated with the BEAM code and the spectrum computed from the analytical model for the 10 MV photon beam.

FIG. 3. The 6-MV model: Output factor at $d_{\max}$ versus field size for the measured, calculated, and corrected output. A hyperbola curve was determined to correct the calculated values.

plan included quantifying the dose in the regions of interest, dose volume histograms, and 2D dose distributions that were evaluated using gamma analysis using 5% of the maximum dose and 3 mm DTA to match the criteria applied by IROC in their QA protocols. In the proof of principle studies, the gamma analysis was used to qualitatively show where the regions of agreement and disagreement existed.

3. RESULTS AND DISCUSSION

3.A. Commissioned source model

The energy spectrums produced as a result of the optimized parameters of the commissioned analytical source model were compared to those computed using the BEAM code.⁴¹ The spectrum for the 6 and 10 MV source models are shown in Figs. 2(a) and 2(b), respectively. The spectrum of the 10 MV source model more closely represented the BEAM

FIG. 4. 10 MV percent depth dose of 40×40 cm field size showing build-up region for the measured and calculated data sets. The plot includes the change in the electron contamination with field size (corrected) versus a constant contribution, regardless of field size (uncorrected).

FIG. 5. Calculated and measured percent depth dose curves for 6 MV, 4×4 cm field and 10 MV, 40×40 cm field.

spectrum than the spectrum of the 6 MV source model. The commissioned Varian 6 MV source model spectrum peak energy differed slightly from that produced with the BEAM code, but provided a good representation of the Varian 6 MV photon beam based on the validation and benchmark results.

The extra-focal energy and fluence for the 6 and 10 MV models were reduced relative to their primary. For the 6 MV the extra-focal source energy was reduced by a factor of 1.7 with a relative photon fluence of 13% and an electron contamination of 0.2%, while the 10 MV model source energy was reduced by a factor of 2.8 with a relative fluence of 23% and an electron contribution of 0.5%. The values determined in our model are consistent with other published reports that modeled the entire accelerator head.^{13,42,43} The dose contribution from electron contamination of the 6 MV model was only 0.2% of the primary dose and was not included. However, the

TABLE I. PDD comparison of the Varian 10 MV 40×40 cm field and the Varian 6 MV 4×4 cm field for a portion of the PDD from 0.8 to 6.0 cm depth showing the local percent difference. The values for the calculated and measured data are percent depth doses relative to 100% at $d_{\max}$ of the 10×10 cm field size.

Depth (cm)	6 MV			10 MV		
	Calculated	Measured	Local difference (%)	Calculated	Measured	Local difference (%)
0.8	86.2	85.7	0.6	100.7	100.8	0.0
1.2	92.8	91.9	1.1	107.0	107.9	-0.9
1.6	92.7	92.6	0.1	109.4	110.4	-1.0
2.0	91.7	91.5	0.2	110.6	111.2	-0.6
2.4	89.5	89.8	-0.4	110.2	110.9	-0.7
2.8	88.9	88.2	0.7	109.3	110.0	-0.6
3.2	86.6	86.4	0.2	108.6	108.7	0.0
3.6	85.0	84.5	0.5	107.2	107.3	-0.1
4.0	83.3	82.6	0.9	105.6	105.9	-0.3
4.4	80.7	80.8	-0.1	103.9	104.4	-0.5
4.8	79.8	78.9	1.1	102.4	102.9	-0.5
5.2	77.5	77.2	0.5	101.2	101.7	-0.5
5.6	75.7	75.4	0.4	100.4	100.2	0.2
6.0	74.2	73.6	0.9	98.4	98.8	-0.4

FIG. 6. Calculated and measured dose profiles at 6 MV from a 4×4 cm field at depths of 1.5 cm ($d_{\max}$), 6, 12.5, and 22 cm. The profiles are shown in order with depth where the profile at the depth of 2.4 cm is shown at the top.

electron contamination source for the 10 MV model (0.5%) was included. The horn-effect coefficients of the piecewise linear function using the 40×40 cm field size showed fluence increased roughly 13% at 10 cm from central axis and about 25% approaching the field edge for both models.

FIG. 7. Calculated and measured dose profiles at 10 MV from a 40×40 cm field at depths of 2.4 cm ($d_{\max}$), 5, 10, and 20 cm. The profiles are shown in order with depth where the profile at the depth of 2.4 cm is shown at the top.

Commissioning was performed based on the 10×10 cm field; the fluence and distribution of photons from the extra focal source, which affect output, did not adequately model the field sizes beyond this. Hyperbolic correction functions were necessary to maintain the proper predicted machine output

a) Gamma map: 3%/2 mm (no MLC offset)

b) Gamma map: 3%/2 mm (with MLC offset)

c) Lateral profile (no MLC offset)

d) Lateral profile (with MLC offset)

FIG. 8. 6 MV IMRT head and neck: Gamma maps (3%/2 mm) [(a) and (b)] and lateral dose profiles [(c) and (d)] from the same irradiation, but before [(a) and (c)] and after [(b) and (d)] changes to the model; implementation of a 0.4 mm MLC leaf bank offset to improve the penumbra. The primary and secondary targets are outlined in white while the critical structure is outlined in red. Note: The lower left corner the film has been clipped and is not part of the gamma analysis.

with varying field size. The hyperbolic function for the 6 MV model is shown in Fig. 3 below (output correction), along with the original and corrected calculated output factors. The original output calculation for the 4×4 cm field size of the 6 MV model overpredicted by about 2% and underestimated by about 7% for the 40×40 cm field. Similarly, the trend was the same for the 10 MV model except the output was over-

estimated by nearly 5% for the 4×4 cm field. The corrected output showed excellent agreement with measurement.

A hyperbolic equation was also developed for the electron contamination of the 10 MV model, $y = 0.025 - [1.008 / (40 + x)]$, where y represents the electron contamination and x is the field size. This function caused the relative fluence to vary from 0.5% for the 10×10 cm field to 1.26% for the

FIG. 9. 6 MV SBRT lung delivery: (a) gamma (b and c) profiles 10 MV IMRT lung delivery: (d) gamma (e and f) profiles.

TABLE II. Varian 6 and 10 MV : The average and range of the percentage of data meeting criteria: gamma index of 3%/2 mm for repeated irradiations from each treatment plan.

		IMRT H&N	SBRT lung	IMRT lung
6 MV	Average % passing	93	94	87
	Range % passing	90–98	90–97	81–92
10 MV	Average % passing	94	96	85
	Range % passing	90–97	91–98	79–90

40×40 cm field. Figure 4 shows the improvement in the build-up region for the 40×40 cm field size. The field size dependent electron contamination also resulted in the correct prediction of the shift in $d_{\max}$ as the field size increased.

3.B. Validation testing results

The uncertainty of the dose determined from the measurements using an ion chamber was estimated to be 1.5% at one standard deviation.⁴⁵ Comparisons of the measured PDD to the source model calculated PDD data for the Varian 6 and 10 MV photon beams for field sizes ranging from 4×4 to 40×40 cm² showed excellent agreement. 100% of data passed the 2%/2 mm criterion up to a field size of 15×15 cm for both 6 and 10 MV. Agreement was slightly poorer for larger field sizes, reducing to 95% and 98% of pixels passing for the 40×40 cm field at 6 and 10 MV, respectively. Average local percent differences were almost always <1% for all cases.

For 6 MV, the few data points that fell outside the stated criteria for the larger fields tended to occur in the build-up region where the calculation reported dose was slightly less than the measured dose. Since the electron contamination was small, it was ignored in the 6 MV model. Including the electron contamination model, which affects shallower depths, and refining its contribution within sound limits may have improved agreement in the build-up region of the larger fields. The location of $d_{\max}$ was typically within 1 mm, but was worst for the 40×40 cm field where the calculated $d_{\max}$ was 1.6 cm versus the measured 1.2 cm. Figure 5 shows the PDD comparison of the 6 MV 4×4 cm field size. In Table I, the 6 MV PDD data from Fig. 5 are compared showing the local percent difference between the depths of 0.8 and 6.0 cm.

Similar good agreement existed between the Varian 10 MV calculated data set and the measured data set along the central axis for all of the field sizes tested between 4×4 and 40×40 cm (Fig. 5). In Table I, the 10 MV PDD data from Fig. 5 are

compared showing the local percent difference between the depths of 0.8 and 6.0 cm. The calculation did predict the shift of $d_{\max}$ toward shallower depths as the field size increased due to the increase in collimator scatter and correctly calculated $d_{\max}$ within 1 mm for all field sizes.⁴⁴

Comparisons of the measured and calculated dose profile data for the Varian 6 MV (e.g., Fig. 6) and Varian 10 MV (e.g., Fig. 7) photon beams were made for the field sizes between 4×4 cm and 40×40 cm at several depths. The worst case agreement for the 6 and 10 MV was the largest field size, but generally good agreement existed throughout the profile at each depth for both models. Disagreement tended to occur for the larger field sizes in the off-axis and penumbra regions where scatter conditions provide their greatest contribution. The penumbra regions benefited from the temporary application of a Gaussian smearing function (standard deviation of 1.2 mm) optimized during commissioning and used for profile comparisons. The smearing function simulated the volume effects of the ion chamber measurement, but indiscriminately accounted for the small effect of the primary finite source size, which had been assumed to be a point. The dose profiles for the field sizes from 4×4 to 40×40 cm field size met the criteria of 2%/2 mm at a minimum of 95% for the 6 MV model and 98% for the 10 MV model. The average local percent difference in the high dose region defined by the measured dose values greater than or equal to 80% of the dose at the central axis was between 0.5% and 1.3%.

3.C. Benchmark testing

Three benchmark plans were created for both energies and delivered to the phantoms three times. The uncertainty of the dose distribution as determined by the EBT film and TLD at one standard deviation was between 2.6% and 3.5%. This is consistent with a previously published report using the same the methodology.⁴⁶ The estimate included the uncertainty of the TLD dose,⁴⁷ the film uniformity, the film-to-film variation, and the fit of the sensitometric curve.⁴⁸ The TLD uncertainty was included because the film was normalized to the adjacent target TLD housed within the phantom.

The importance of good modeling of the MLCs and the resultant fluence map is shown by way of an example in Fig. 8. The 6 MV IMRT head and neck gamma map (3%/2 mm) of the axial plane through the center of the target and an associated lateral dose profile shows the difference in calculated dose from the introduction of a 0.4 mm offset that compensates for

TABLE III. Point dose comparisons of all plan types for 6 and 10 MV photon beams. The measured dose is the average dose from three repeated irradiations for each plan type and energy. Calculated results are from DPM. The Pinnacle calculation is provided for reference.

Plan type	Point dose location	6 MV		10 MV	
		DPM/TLD	Pinnacle/TLD	DPM/TLD	Pinnacle/TLD
Head and neck	Primary	0.983	1.043	0.987	0.986
	Secondary	0.991	1.031	0.981	1.014
SBRT thorax	Primary	1.013	1.014	1.011	0.992
IMRT thorax	Primary	1.025	1.019	1.022	1.020

TABLE IV. Ratio of TPS calculation to DPM calculation for mean dose to GTV, PTV, and critical structure for patient and phantom plans. The uncertainty of the TLD was estimated to be 3.5% at one standard deviation.

Plan	Treatment	Energy	TPS/DPM	
			GTV	TPS/DPM PTV
Patient	IMRT abdomen	6	1.016	1.008
	SBRT lung		1.000	0.984
	IMRT prostate	10	1.000	0.996
	IMRT lung		N/A	1.008
Phantom	IMRT H&N	6	1.049	N/A
	SBRT lung		1.002	0.987
	IMRT lung		1.015	1.010
	IMRT H&N	10	1.019	N/A
	SBRT lung		0.990	0.974
	IMRT lung		1.008	0.984

the finite-dimensioned primary source that was assumed to be a point source. There is a drastic improvement in the dose agreement with a relatively small positioning change in the MLC. The offset amount was fixed for the Varian accelerator as it has been reported that the source size was nearly the same regardless of energy.^{20–24} Based on the relationship of the source-to-MLC distance and the source-to-axis distance, the source diameter, was 0.84 mm FWHM. Note that in addition to an increase in the calculated dose to the primary and secondary targets in the phantom, there was an improvement in the widening and shape of the penumbra.

To illustrate the ability of the DPM dose calculation to predict dose accurately, representative results for the six phantom plans are shown in Fig. 9 (6 MV SBRT and 10 MV IMRT lung plans). In both plans, gamma agreement existed in the gross tumor volume at the acceptance criteria of $\pm 3\%/2$ mm. Some small difference existed in the low dose fall-off regions, but overall good agreement existed throughout for the SBRT. For the IMRT lung plan, the gamma map showed small regions of overestimation in the penumbra region. The penumbra region represented not only the change in the high dose region to low dose region but also the change from the target tissue-like density to the low-density region of the lung where electronic disequilibrium occurred. Disagreement may also be more likely to occur near the lung and the chest wall for the same reason as the target is located within about 1 cm from this region [Fig. 1(b)].

FIG. 11. Gamma map (5%, 3 mm) of the 10 MV IMRT lung patient plan and associated axial CT image with DPM calculated dose distribution. The black arrow shows the dose line profile in the CT image for Fig. 12.

FIG. 10. DVH of the 10 MV IMRT lung patient plan for the PTV, carina, esophagus, and the lungs.

The overall gamma results are given in Table II. The results are the average percentage of data meeting the gamma criteria at 3%/2 mm for all three repeat irradiations on all planes of the 6 plans.

Table III shows the results of the point dose comparisons for average TLD measurement locations for all of the plans.

The agreement between the DPM calculation and measurement for all of the targets on average was within 2.5%. The dose results from the Pinnacle calculations have been provided as representative of the comparison one might find between the source model calculation and treatment planning system calculations.

Table IV shows mean dose ratios between the TPS and DPM calculations from several regions of interest for the patient treatment plans studied (prostate, abdomen, and lung), as well as for the treatment plans delivered (head and neck and thorax) to the anthropomorphic phantoms. In all cases, the difference in the mean dose to the GTV and to planning target volume (PTV) between the calculations was within 5% and typically within 3%.

Dose volume histograms (DVHs), gamma maps (criteria, 5%/3 mm), and dose profiles were generated for the patient treatment plans studied (prostate, abdomen, and lung), as well as the for the treatment plans delivered (head and neck and lung) to the anthropomorphic phantoms. All of the results showed the same tendencies in terms of accuracy, and a select case is presented (Fig. 10). DVHs from the TPS and DPM calculations of a 10 MV IMRT lung patient case are shown for the PTV,

FIG. 12. Dose (cGy) line profile comparing TPS (gold) and DPM (green) calculations. Red line denotes relative CT number. Location of profile is shown in the CT image of Fig. 10. (See color online version.)

carina, esophagus, and lungs. In general, the TPS calculation tended to estimate the dose higher than the DPM calculation. However, at D_{95} the DPM calculation was 3.0% higher and at D_{90} they were negligibly different. The largest difference in the DVH comparison of the PTV was between at D_{30} and D_{80} , and ranged from 1.1% to 1.3%. The D_{30} to D_{80} dose difference for the carina ranged from 3.6% to 7.0%. For the esophagus, the largest dose differences ranged from 7.0% to 9.8% for the dose volumes between D_{10} and D_{40} . The dose to the lungs was consistently the same throughout the lung volume.

Figure 11 shows the gamma map of the axial CT image in the center of the PTV for the lung patient case between the TPS and DPM calculations. Similar to the results of all of the treatment plans, disagreement occurred at the skin surface as well in the beam penumbra regions. The shape of the tumor was irregular so that a small volume of lung tissue was nearly surrounded by the tumor. In this region, differences between the two calculations are clear. The dose profile in Fig. 12 showed that the DPM calculation predicted a build-down of dose within the small volume of lung that was circumscribed by the tumor. The CT value in this region was defined as air and not lung in the MC code. The MC code used the stopping power values of dry air in computing the dose, which may explain the difference in this region.

4. CONCLUSION

A custom-developed method for dose recalculation of patient plans that entered clinical trials was found to be accurate under a variety of conditions. The measurement-driven source model with analytical function parameters derived from a standard set of basic beam measurement data exceeded the objectives specified for agreement between the calculation and measurement for the common range of square fields at energies

of 6 and 10 MV. The validated models also met or exceeded the objectives specified for agreement between the calculation and measurement for a variety of benchmark conditions using the IROC Houston anthropomorphic phantoms and the common SBRT and IMRT treatment techniques. Application of these results was demonstrated with patient cases where a commercial treatment planning system's dose calculation algorithm was evaluated against the DPM method. The DPM method generated predictions similar to the TPS system for IMRT and SBRT treatment plans. Differences were consistent with the superior physics modeling inherent in the Monte Carlo code. As a result, we believe the method will be useful for recalculating dose distributions for patients who entered clinical trials.

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CONFLICT OF INTEREST DISCLOSURE

The authors have no COI to report.

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