A flexible Monte Carlo tool for patient or phantom specific calculations: comparison with preliminary validation measurements

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Abstract. The Dose Planning Method (DPM) is one of several “fast” Monte Carlo (MC) computer codes designed to produce an accurate dose calculation for advanced clinical applications. We have developed a flexible machine modeling process and validation tests for open-field and IMRT calculations. To complement the DPM code, a practical and versatile source model has been developed, whose parameters are derived from a standard set of planning system commissioning measurements. The primary photon spectrum and the spectrum resulting from the flattening filter are modeled by a Fatigue function, cut-off by a multiplying Fermi function, which effectively regularizes the difficult energy spectrum determination process. Commonly-used functions are applied to represent the off-axis softening, increasing primary fluence with increasing angle (‘the horn effect’), and electron contamination. The patient dependent aspect of the MC dose calculation utilizes the multi-leaf collimator (MLC) leaf sequence file exported from the treatment planning system DICOM output, coupled with the source model, to derive the particle transport. This model has been commissioned for Varian 2100C 6 MV and 18 MV photon beams using percent depth dose, dose profiles, and output factors. A 3-D conformal plan and an IMRT plan delivered to an anthropomorphic thorax phantom were used to benchmark the model. The calculated results were compared to Pinnacle v7.6c results and measurements made using radiochromic film and thermoluminescent detectors (TLD).

1. Introduction

The use of measurement devices such as ion chambers, thermoluminescent detectors (TLD), and radiochromic film to determine absorbed dose is well understood, and therefore trusted. The Radiological Physics Center (RPC) has traditionally relied on radiation measurements using TLD and radiochromic film registered within anthropomorphic phantoms to ensure that institutions participating in National Cancer Institute (NCI) sponsored clinical trials administer consistent levels of radiation dose. Meanwhile, advances in radiation therapy delivery, such as Intensity Modulated Radiation Therapy (IMRT) and Stereotactic Body Radiation Therapy (SBRT), have improved dose conformity to the target while limiting dose to critical structures, giving way to the generation of more tightly conforming dose distributions. As a result, the ability to measure dose accurately in high dose gradient regions, especially in regions of anatomical heterogeneity, has come into question.[1] Clinical trial working groups and the RPC have recognized the challenge associated with verifying
treatment delivery of IMRT and SBRT, especially in heterogeneous tissues, and have discussed ways to implement a quality assurance (QA) tool that could be used to audit all treatment planning system (TPS) dose calculations for clinical trial QA purposes. One way to address the difficult dosimetry challenge is through the use of an independent dose calculation algorithm. Monte Carlo dose calculations have been perceived as a complementary method to both measurement and analytical based numerical calculations and may provide the best solution for an independent dose calculation that the RPC can use.[2] In this work, we present the development of a versatile source model calibrated for a Varian linear accelerator head as the input to the Dose Planning Method (DPM) Monte Carlo source code[3] and the preliminary measurement benchmark testing of the code’s ability to calculate radiation doses accurately. The tool is further integrated with the Computational Environment for Radiotherapy Research (CERR)[4], which enables the comparisons to be made with imported commercial treatment planning system calculations.

2. Material and Methods

2.1. Source Model

The source model of the primary photon source energy spectrum is described by the product of the Fatigue-Life[5] and the Fermi functions [equation 1]. We refer to this combined distribution as the Fatigue-Fermi distribution where $E$ is the photon energy, $E_F$ is the cut-off energy, $\mu$, $\gamma$, and $\beta$ shape the photon spectra, and $\phi$ is the probability density function of the standard normal distribution. This fitting function was selected from among an array of possible statistical distribution functions for its ability to fit the nine different photon spectra from various linear accelerator manufacturers based on the BEAM simulations performed by Sheikh-Bagheri and Rogers[6,7].

$$f(E) = \left( \frac{E - \mu}{\beta} + \frac{\beta}{2\gamma(E - \mu)} \right) \left( \frac{E - E_F}{E - \mu} - \frac{\beta}{E - \mu} \right) \frac{1}{\phi(\frac{E - E_F}{kT})} \phi \left( \frac{E - \mu}{\beta} - \frac{\beta}{E - \mu} \right) \left( \frac{E - E_F}{kT} \right) \frac{1}{E > \mu; \gamma, \beta > 0} \quad (1)$$

The source of primary photons (those that reach the patient/phantom without scattering) is modeled as an isotropic point source with the ‘horn effect’ to account for the increase of the fluence as the off-axis angle increases. Additionally, contributions from the flattening filter were included. The flattening filter photon distribution is modeled as a disk source consistent with an exponential intensity as a function of the radius.[8] The photon spectra resulting from the flattening filter, about which little can be inferred from measured data, was also modeled using the same Fatigue-Fermi function, but is scaled down in energy by a single parameter to account for the reduction in the mean energy.

Commonly used functions described below are applied to represent the off-axis softening, horn effect, and electron contamination. The presence of the flattening filter enhances the off-axis softening due to preferential scattering of low-energy photons near the central-axis resulting in lower mean energy and greater beam intensity off-axis. Known as the ‘horn effect’, this can be seen in the dose profiles at shallow depths, becoming less pronounced and less flat with increased depth as the lower mean energy photons quickly attenuate as a function of depth. In our model, the off-axis softening is considered by relating the off-axis half-value layer (HVL) as a function of the off-axis angle, based on the formalism of Tailor, et al.[9] Related to off-axis energy softening is the horn effect, which is modeled by assuming the fluence increases linearly with the off-axis angle. Our most recent model extends this to a piecewise linear function. An electron contamination source (i.e. electrons produced outside the patient/phantom) is assumed to be uniformly distributed on a disk, located at the flattening filter plane, with an energy spectrum which is modeled using an exponential function proposed by Fippel, et al.[10] As we know, ionization chambers (IC) measure an artificially wider penumbra, as described by Arnfield et al.[11] We apply a Gaussian convolution to model this effect, the convolution width is determined during the commissioning process. Finally, the fluence map is generated based on the leaf sequence file exported from TPS, by summing all of the segments.
Based on the fluence map, rectangular beamlets and their weights are generated. For each beamlet, the dose engine, coupled with source model, are called. Then, the doses are scaled by their weights and summed to get the composite dose. The computed tomography scan is imported from the TPS for the particle transport. Figure 1 describes the commissioning process of the source model broken into two sections describing the determination of the fitting parameters and the horn-effect coefficients.

![Diagram](image)

**Figure 1.** Commissioning the head model for linac photon beam.

Built around the source model and the DPM dose calculation is a graphical user interface (GUI). The GUI was designed in the software application Matlab (MathWorks Inc., Natick, MA) and runs off the Computational Environment for Radiotherapy Research (CERR) software platform developed at Washington University.[4] CERR was developed so that researchers in the radiotherapy community could share results in treatment planning. The platform provides a common data structure for the creation of multi-institutional treatment plan databases for various types of research studies, including dose-volume-outcome analysis and IMRT treatment planning comparisons.

### 2.2. Measurement

Preliminary benchmark testing of an anthropomorphic phantom planned using conventional 3-D planning and IMRT planning was performed. The 3-D plan was artificial in that it consisted of five coplanar beams each separated by 72 degrees and relied on the primary and secondary jaw positions to define the fields without using the MLCs as they were kept in their retracted position. This approach was important to isolate the effect on the dose calculation with respect to the MLC transmission contributions. In addition, an IMRT lung plan was also planned using the entire treatment delivery system including the MLC sequence file. The IMRT plan consisted of a step-and-shoot delivery of four coplanar beams and one non coplanar beam comprising 63 MLC static segments.

The anthropomorphic thorax phantom used was that described by Followill et al.[12] and is shown in figure 2a and 2b. Briefly, this phantom is used by the RPC as a QA tool to monitor planned dose deliveries from institutions participating in NCI sponsored clinical trials for lung tumors. The lung phantom was designed to represent the heterogeneous thoracic anatomy. The size, shape, and content of the phantom were based on patient studies submitted to the RTOG 93-11 lung protocol by The University of Texas M. D. Anderson Cancer Center (MDACC). The dosimeters housed within the anthropomorphic thorax phantom include thermoluminescent detectors (TLDs) and radiochromic film, which provide point dose comparisons of the tumor, heart, and spinal cord, as well as dose profile comparisons through the planning target volume (PTV) and low dose lung regions. The phantom was irradiated three times to evaluate the reproducibility of the measured values. For comparison, the Pinnacle superposition convolution calculated values are included in the analysis.
3. Results
As will be reported, we found that the Fatigue-Fermi function is able to fit nine different photon spectra from various linear accelerator manufacturers based on the BEAM simulations performed by Sheikh-Bagheri and Rogers.[6] Thus, we believe it is flexible enough to handle the range of commercial linear accelerators while also not being over-parameterized. All comparisons yielded correlation coefficients greater than 0.998 indicating excellent agreement. For example, figure 3 shows the fitting results for a Varian 10 MV photon spectrum. The root-mean-square (rms) difference is within 1%. To test the accuracy of the physics in DPM, simulations were performed for an isotropic point source, using DPM, BEAM[13], and VMC++[14]. The PDD curves of the three calculations shows good agreement for a 6 MV photon beam, 0.5 cm by 0.5 cm field size, to a water phantom at 100 cm source to surface distance (SSD) (figure 4). (BEAM calculations were performed by Jan Seuntjens and Frank Verhaegen.) For both comparisons, the rms difference is within 1%.

Figure 5 shows the photon spectra of the primary source and the flattening filter. Note that the two spectra have the same shape, but the mean energy is scaled down for the flattening filter. The basic commissioning data of the DPM source model tool was compared to the standard IC measurement describing the performance of a Varian linear accelerator. Figure 6 shows the calculated and measured PDD for 6 MV photon beam, 10x10 cm² field size, in a water phantom at 100 cm source to SSD, while figure 7 compares the calculated and measured profiles at several depths, where the calculated doses are within 1% RMS of measurements.

The point dose comparison of the 3-D and IMRT treatment plans between calculations and measurement from the TLD locations in the tumor, heart, and cord from each treatment plan are shown in table 1 and table 2. The DPM calculation has an uncertainty within 1%. For both plans the tumor dose determined from DPM and the TPS is within 2.5% of TLD measurement.
In the low dose regions of the critical structures, the 3-D plan had more inconsistencies in determining the dose than in the IMRT plan where the fluence maps provide more complexity to the calculations. For instance, in the heart where the dose gradient was high, DPM and the TPS agree within 5% of each other, while both are lower than TLD by about 25%. The setup uncertainty of the phantom may, in part, contribute to discrepancies within this high gradient region. Also, it should be pointed out that comparisons made in low dose regions are typically based on a percentage of a normalization point, such as a point within the PTV, not locally as presented in tables 1 and 2 for the heart and cord.[15]

Table 1. 3-D plan point dose comparisons of calculations to TLD measurements.

| Calculation | Tumor | Heart | Cord |
|-------------|-------|-------|------|
| DPM         | 1.008 | 0.736 | 0.948|
| Pinnacle    | 1.025 | 0.774 | 1.083|

Table 2. IMRT plan point dose comparisons of calculations to TLD measurements.

| Calculation | Tumor | Heart | Cord |
|-------------|-------|-------|------|
| DPM         | 1.018 | 1.075 | 0.993|
| Pinnacle    | 1.019 | 1.053 | 0.983|

Figure 8. 3-D plan: DPM profile shows good agreement to film measurements and is consistent with TPS. Error bars show range of dose values measured from three repeated irradiations.

Figure 9. IMRT plan: DPM profile shows good agreement to film measurements and consistent with TPS. Error bars show range of dose values measured from three repeated irradiations.
Profile comparisons were made through the center of the tumor, extending through the penumbra and into the low dose lung region. As shown in figure 8 for the 3-D plan and in figure 9 for the IMRT plan, good agreement exists between the DPM calculation and measurement. For both plans, the lateral dose profile of calculation is within 3% agreement with the film, by rms difference measure. The error bars in the film data show the range of dose measured from the three repeated irradiations. The TPS superposition convolution calculation is also shown for reference. For the calculations, DPM uses air, lung, tissue, and bones material specifications and not those of the phantom, except for water. The phantom was CT scanned without the film or TLD present. Therefore, the densities in these regions were over-ridden with the densities assigned in the associated ROIs.

4. Conclusion
A versatile source model used for Monte Carlo dose calculation of radiation therapy has been developed and tested. This tool has been coupled with DPM Monte Carlo dose calculations in predicting dose distributions in IMRT clinical applications such as the quality assurance of treatment planning dose calculations. The model is built based on a primary point source supplemented by the flattening filter and electron contamination sources. We also proposed the so called Fatigue-Fermi curve which is able to fit various photon spectra for various linear accelerators [7].

Preliminary tests against data collected for a Varian linear accelerator were successful. In addition to basic beam data comparisons between calculation and measurement, using the source model with the DPM dose calculation, we benchmarked the performance of a 3-D conventional plan (MLCs retracted) and IMRT plan designed for an anthropomorphic phantom. In each case, point dose comparisons with in the phantom tumor were within 2% of the TLD measurements. Profile comparisons revealed the DPM dose calculation was capable of predicting the dose in the penumbra of low density regions, with 3% rms agreement with the film measurement. Although the current machine model appears adequate for radiotherapy quality assurance uses based on these initial tests, more work is necessary to validate and improve the model.

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