AAA Photon Dose Calculation Model in Eclipse™

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Abstract

A new photon dose calculation model, the Analytical Anisotropic Algorithm (AAA), has been implemented in Eclipse™ Integrated Treatment Planning. The AAA model provides a fast and accurate dose calculation for clinical photon beams even in regions of complex tissue heterogeneities.

The AAA dose calculation model is a 3D pencil beam convolution-superposition algorithm that has separate modeling for primary photons, scattered extra-focal photons, and electrons scattered from the beam limiting devices. Functional forms for the fundamental physical expressions in AAA allow analytical convolution, thus reducing significantly the computation times usually required by these types of algorithms. Tissue heterogeneities are accounted for anisotropically in the full 3D neighborhood by the use of 13 lateral photon scatter kernels. The final dose distribution is obtained by superposition of the doses from the photon and electron convolutions.

Configuration of the AAA model is based on Monte-Carlo-determined basic physical parameters that are adapted to measured clinical beam data. These in turn are used to construct a phase space defining the fluence and energy spectrum of the clinical beam specific to each treatment unit. Beam modifying accessories including blocks, hard wedges, dynamic wedges, compensators, MLCs (static and dynamic) are fully supported in the dose calculation.

Introduction

Modern treatment techniques in clinical external beam radiotherapy are posing increasing demands on the accuracy and speed of the dose calculation algorithms. Recent introduction of the highly modulated beams used in IMRT techniques also necessitates accurate dose calculation especially in regions of complex tissue heterogeneities. At the same time, resource constraints in a real clinical environment dictate that any arbitrarily high accuracy cannot be afforded at the expense of lengthy computation times. The AAA photon dose calculation model has been developed to meet both of these clinical expectations, providing a fast Monte-Carlo-based 3D convolution/superposition algorithm for accurate heterogeneity-corrected photon dose calculation for all types of external beam treatments.

Originally developed by Drs. Waldemar Ulmer[2,3] and Wolfgang Kaissl[3], AAA has a long history culminating in the publication of the triple Gaussian photon kernel model in 1995 [2, 3, 4]. The principal ideas of AAA have already been applied to stereotactic radiation therapy planning [5] where the application of the triple Gaussian kernel formalism with appropriate correction mechanisms for tissue heterogeneities has yielded excellent results. The AAA dose calculation model presented in this article is a continuation of the work presented in these previous publications.

The AAA dose calculation model is comprised of two main components, the configuration algorithm and the actual dose calculation algorithm. The configuration algorithm is used to determine the basic physical parameters used to characterize the fluence and energy spectra of the photons and electrons present in the clinical beam and their fundamental scattering properties in water equivalent medium. Although some of the parameters used in the dose calculation algorithm could be deduced with reasonable accuracy from simple measurements of depth dose and lateral dose profiles in a water-equivalent phantom, an experimental determination of all parameters is practically impossible. This is resolved in the AAA model by pre-computing all the parameters using Monte Carlo simulations and then modifying these parameters to match with the actual measured clinical beam data during the beam data configuration phase. This approach ensures a quick and highly accurate determination of all the important basic physical parameters required for the AAA dose calculation. After the treatment-unit-specific fitting procedures in the

beam configuration phase are completed, all the parameters are stored and later retrieved for the actual dose calculation.

The dose calculation is based on separate convolution models for primary photons, scattered extra-focal photons, and electrons scattered from the beam-limiting devices. The clinical broad beam is divided into small, finite-sized beamlets to which the convolutions are applied. The final dose distribution is obtained by the superposition of the dose calculated with photon and electron convolutions for the individual beamlets.

Functional forms of the fundamental physical expressions in the AAA enable analytical convolution, which significantly reduces the computational time required in the dose calculation. Attenuations of the photons and electrons present in the clinical beam are modeled with the energy deposition density functions I, and the dose deposition characteristics with scatter kernels K that are composed of Gaussian functions.

AAA accounts for tissue heterogeneity anisotropically in the entire three-dimensional neighborhood of an interaction site. This is performed by the use of radiological scaling of the dose deposition functions and the electron density based scaling of the photon scatter kernels independently in four lateral directions.

This article concentrates on the description of the dose calculation algorithm in the AAA model. Descriptions of the algorithms developed for the beam data configuration to determine the basic physical parameters required in the dose calculation are largely omitted and will be presented in detail in a separate article.

Methods

Physical Parameters in Dose Calculation

AAA uses fundamental physical parameters that have been pre-computed with Monte Carlo simulations. These parameters are modified during beam data configuration so that the resulting calculated beam characteristics match the measured clinical beam data for each treatment unit. This is a fast and accurate method of determining all the important physical parameters for the

dose calculation. The parameters specific to the clinical beam are stored in the database and retrieved for the patient dose distribution calculation.

The fundamental physical parameters used to model clinical beams are pre-computed for a set of average beam energies from 6 to 23 MV. A Varian Clinac© 2300 C/D was used as the reference accelerator for the determination of the beam fluence and energy phase space. All model parameters for AAA are computed in a water-equivalent medium. During the dose distribution calculation, these parameters are scaled according to the densities of actual patient tissues.

Photon Energy Spectrum

AAA derives the scatter kernels K needed for the dose calculation from the energy spectrum, which is determined during the configuration process. The initial photon spectrum is determined from Monte Carlo simulations of the Bremsstrahlung spectrum of the electrons impinging on the target. Figure 1 shows an example of an initial photon spectrum for a 6 MV beam.

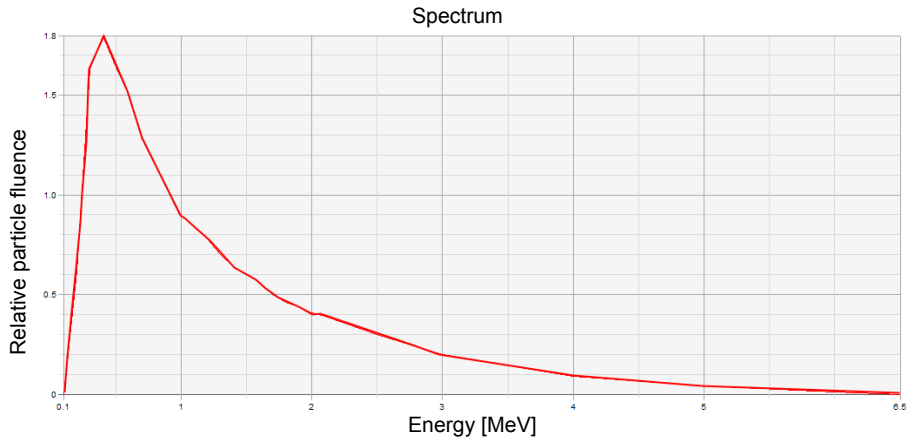

Figure 1: Example of a 6 MV photon spectrum.

Another important parameter that affects the energy spectrum used by AAA is the mean energy as a function of the radius from the beam central axis. An example of the mean radial energy for a 6 MV beam is given in Figure 2. This curve is used by AAA to determine the beam hardening effect of the flattening filter on the photon spectrum. Based on the mean energy curve and the

user-specified flattening filter material, AAA determines the energy spectrum of the beam at any radius from the beam central axis.

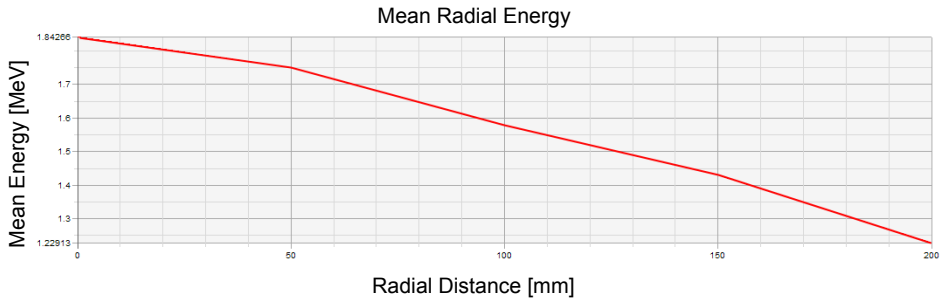

Figure 2: Example of the mean energy as a function of distance from the beam central axis of a 6 MV photon beam.

Intensity Profile

The flattening filter also causes the intensity of the photon beam to vary across the clinical treatment field. The varying photon fluence is modeled with the help of a parameter called the intensity profile curve. The intensity profile is computed as the photon energy fluence (number × energy of photons) as a function of the radial distance from the beam central axis. An example of the intensity profile for a 6 MV beam is shown in Figure 3.

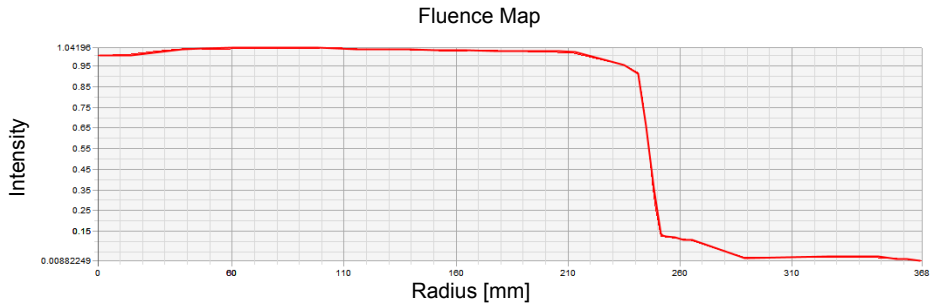

Figure 3: Example of intensity profile of an 18 MV photon beam.

Scatter Kernels

In addition to the phase space parameters, the fundamental physical parameters also include the photon and electron scatter kernels and their depth dependence in a water-equivalent medium. These scatter kernels describe the phantom-scatter effects for different beam qualities. The

EGSnrc Monte Carlo code [6] was used to compute scatter kernels for monoenergetic pencil beams in various media. A polyenergetic scatter kernel is constructed as a weighted sum of the monoenergetic scatter kernels. During the 3D dose calculation these parameters are scaled according to the densities of the actual patient tissues determined from the CT images.

Clinical Beam Modeling

The clinical beam model has three main components: primary photon energy fluence, extra-focal photon energy fluence, and contaminating electron fluence. These are characterized by a number of parameters.

Monte Carlo Simulations for the Clinical Beam Model

The Monte Carlo simulations used to create the initial phase-space model for the clinical beam include the geometrical structure and exact material composition of the accelerator head (including source, target, primary collimator, flattening filter, monitor ionization chamber, and jaws), and the beam modifiers. The parameters derived by Monte Carlo simulation are modified during beam data configuration so that the resulting calculated beam characteristics match the clinical beam data for each treatment unit [1]. The simulations were performed mainly using the EGSnrc program [6], which is well-suited for modeling physical interactions in clinical external beam radiotherapy, such as Compton scattering with outer and inner-shell electrons, Bremsstrahlung, photoelectric effect, and electron-positron pair production and annihilation.

The broad clinical beam is divided into finite-sized beamlets β, as illustrated in Figure 4. Additionally, the clinical beam is divided into separate photon and electron components, each with a beamlet intensity Φ_β. The photons are divided into:

- Primary photons, originating from the target. Separate energy spectra for every fanline of the broad beam are derived from the mean energy curve (see Figure 2).
- Extra-focal photons, scattered in the flattening filter or the beam limiting devices. Extra-focal photons are assumed to be uniformly distributed across the broad beam, and they are modeled with a secondary source having a configurable intensity and location below the primary source collimated by the field aperture [7].

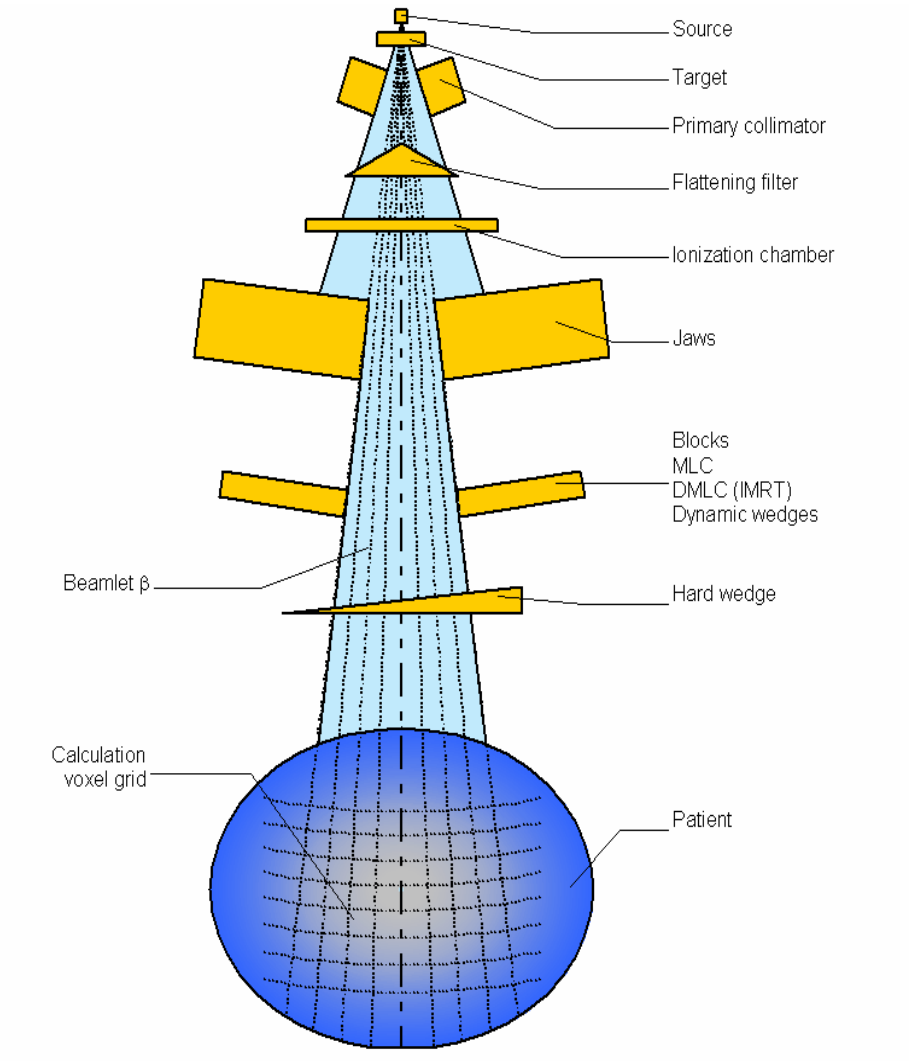

Figure 4: Treatment unit components, broad beam division.

Beam Modifiers in the Clinical Beam Model

Most beam-modifying accessories affect only the beam fluence used in the AAA dose calculation. Blocks and MLCs employ a user-defined transmission factor to model the change in the fluence when a beam is shaped by the accessory. Compensators, Dynamic Wedges and IMRT fields also modify the fluence of the beam. The head scatter effects are taken into account by convolving the fluence distribution with the Gaussian shape of the secondary source to

produce varying amounts of extra-focal radiation, depending on the field shaping. The contribution of the contaminating electrons also depends on the photon beam fluence shape.

Hard wedges modify the fluence and the spectral characteristics of the beam. The configuration program determines the beam hardening effect from the appropriate lateral profile measured. The user-defined wedge material is used to determine the effect of beam hardening on the configured open field energy spectrum.

Volumetric Dose Calculation

For volumetric dose distribution calculation, the patient body volume is divided into a matrix of 3D calculation voxels based on the selected calculation grid (see Figure 4.) The geometry of the calculation voxel grid is divergent, aligning the coordinate system with the beam fanlines. Every calculation voxel is associated with the mean electron density ρ that is computed from the patient CT images according to a user-specified calibration curve.

The 3D dose distribution is calculated from separate convolutions for each of the primary photons, extra-focal photons and contaminating electrons. The convolutions are performed for all finite-sized beamlets that comprise the clinical broad beam. The final dose distribution is obtained by a simple superposition of the individual beamlet contributions.

Beamlets

Figure 5 shows the geometrical definitions of the coordinates referring to a single beamlet β on the X–Z plane, with the Y-axis pointing outwards from the paper. The coordinates are defined in two coordinate systems: patient and beamlet. The coordinates of the calculation point (P) in the figure are $(\tilde{x},\tilde{y},\tilde{z})$ in the patient coordinate system, and (x, y, z) in the beamlet coordinate system. The depth coordinate z is measured from the intersection point of the central fanline and the skin in the beamlet coordinate system.

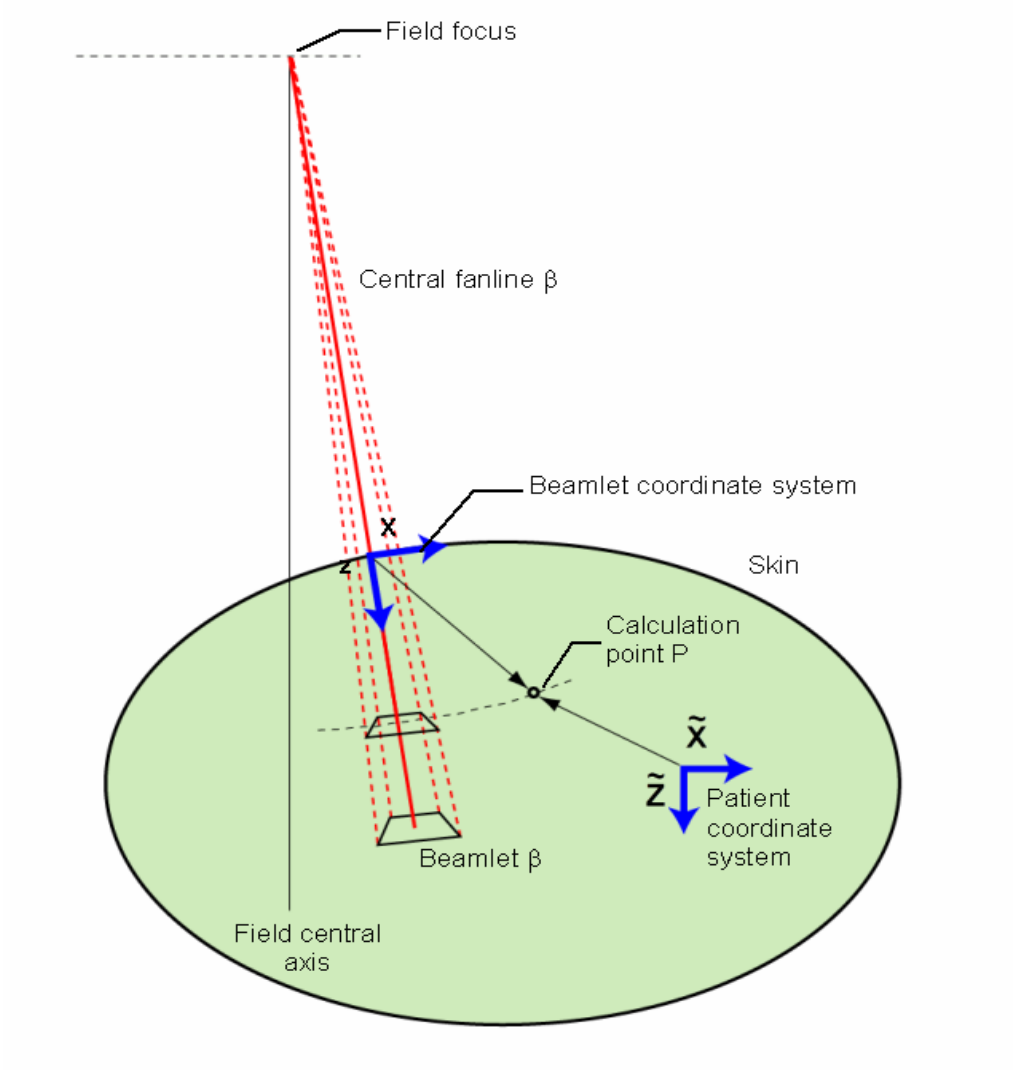

Figure 5: Coordinates in patient coordinate system and beamlet coordinate system on X–Z plane.

The broad clinical beam is divided into finite-size beamlets β. The cross-sectional area of a beamlet corresponds to the resolution of the calculation voxel.

The dose calculation is based on the convolutions over the beamlet cross-sections separately for the primary photons, extra-focal photons (second source), and for electrons contaminating the primary beam. The dose is convolved by using the basic physical parameters defined for every beamlet β.

All depth-dependent functions used in the beamlet convolutions are computed along the central fanline of the beamlet using the depth coordinate z that defines the actual distance traveled from the surface of the patient body (Figure 5.) Lateral dose scattering due to photons and electrons is defined on the level perpendicular to the central fanline of the beamlet.

The dose to an arbitrary calculation point $(\tilde{x},\tilde{y},\tilde{z})$ in the patient is obtained by summing up the dose contributions of all individual beamlets β of the broad beam in the final global superposition.

Photon Dose Calculation

The photon beam attenuation is modeled with an energy deposition density function $I_\beta(z,\rho)$. The photon scatter is modeled with a scatter kernel $K_\beta(x,y,z,\rho)$ that defines the lateral dose scattering. Both functions I and K are defined individually for each beamlet β. The primary and extra-focal photons are calculated in the same way, with the exception of their spectral composition and the position and size of the focal spot.

The dose distribution resulting from an arbitrary beamlet β due to photons in a sufficiently large homogenous neighborhood is calculated by the following convolution:

$$D_{\beta,ph}(\tilde{X},\tilde{Y},\tilde{Z}) = \Phi_\beta \times I_\beta(z,\rho) \times \iint_{(u,v)\in Area(\beta)} K_\beta(u-x,v-y,z;\rho)\,du\,dv \qquad (1)$$

In the convolution, the calculation point $(\tilde{x},\tilde{y},\tilde{z})$ is represented by (x,y,z) relative to the origin of the beamlet coordinate system. The photon fluence Φ_β is assumed to be uniform over the small cross-section of beamlet β.

The energy deposition density function $I_\beta(z,\rho)$ denotes the area integral of the dose over the transverse plane of the pencil beam at depth z, normalized to a single incident photon. The poly-energetic function $I_\beta(z,\rho)$, based on the photon beam spectrum, is constructed from the superposition of pre-calculated monoenergetic energy deposition density functions.

The energy deposition density function I(z,ρ) accounts for tissue heterogeneity by employing the concept of radiological scaling. This is performed by setting I(z,ρ) = I(z'), where the radiological depth z' is defined as

$$z' = \int_0^z \frac{\rho(t)}{\rho_{water}} dt$$

where ρ is the electron density.

The photon scatter kernel $K_\beta(x,y,z,\rho)$ is composed of the weighted sum of four Gaussian functions as shown in the following equation:

$$K_\beta(x,y,z,\rho) = \sum_{k=0}^{3} c_k(z) \frac{1}{\pi\sigma_k^2(z)} \exp\left[-\frac{x^2+y^2}{\sigma_k^2(z)}\right] \quad (2)$$

The Gaussian kernels are characterized with the standard deviations σ_k (Figure 6.) The factors c_k define the weights for the four Gaussian kernels and ensure the unity normalization of the total kernel energy. The above parameters of the polyenergetic scatter kernel $K_\beta(x,y,z,\rho)$ are determined using the Monte Carlo calculated monoenergetic scatter kernels and the spectrum of the photon beam.

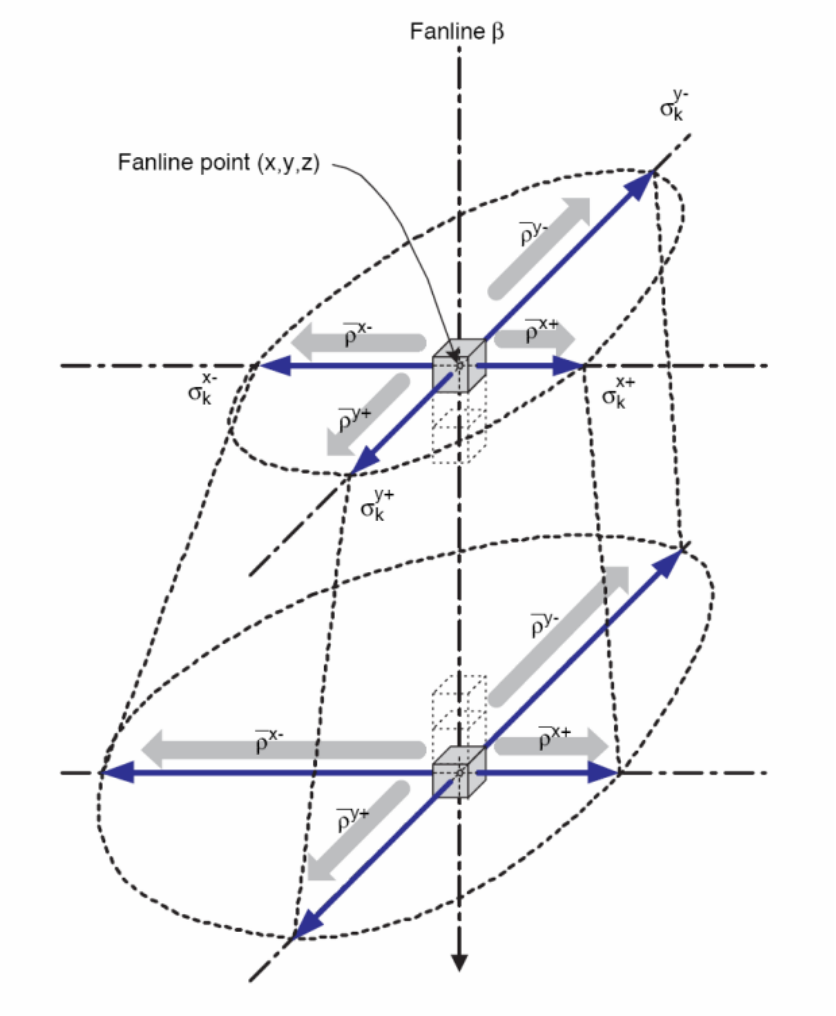

Figure 6: Density scaling for photon scatter sigmas.

Density scaling for the photon scatter kernels is performed separately along the four main lateral directions and according to the average density in these directions.

Scatter kernels σ_k and their depth dependencies are determined at the time of configuration for a water-equivalent medium. In heterogeneous media, the values σ_k in Equation 2 are replaced by density scaled values $\tilde{\sigma}_k^d$ $(k \neq 0)$. During the dose distribution calculation, σ_k when traveling along the fanline from depth z_0 to z_n, is evaluated in a manner that allows the density change to gradually affect the scatter parameters. The short range effect σ_0 is not scaled.

The density scaling also accounts for the heterogeneity in the neighboring voxels. Therefore, the σ_k are individually scaled for the density in four main lateral directions. This is denoted by the labels $d \in \{x+,x-,y+,y-\}$ separately (Figure 6.) Scatter kernels σ_k are scaled according to the average density $\bar{\rho}^d$ computed over the distance of the effective range of the σ_k along the main lateral directions d as in the following equation:

$$\tilde{\sigma}_k^d(z_n) = \sigma_k(z_n) \times \left(\frac{\rho_{water}}{\bar{\rho}^d(z_n)} \right) \text{for } d \in \{x+,x-,y+,y-\} \tag{3}$$

Hence, as a whole, the photon scatter in heterogeneous surroundings is anisotropically modeled by scaling the scatter kernels in all directions.

An essential feature of the Gaussian kernel is that its definitive integral form is conveniently expressed as a sum of error functions [2, 5]. This allows convolution in Equation 1 to be performed analytically, which significantly decreases the computation time required for dose distribution calculation.

Contaminating Electrons

The primary photon beam is contaminated with electrons originating mainly in the flattening filter, ion chamber, collimating jaws and air. If beam modifiers are used, the modifiers may absorb most of the electrons in the open beam, but the modifier itself becomes a secondary source of contaminating electrons. In general, electron contamination depends strongly on the beam energy and the field size.

The dose distribution resulting from an arbitrary beamlet β due to the contaminating electrons is calculated by the following convolution:

$$D_{cont,\beta}(\tilde{X},\tilde{Y},\tilde{Z}) = \Phi_{cont,\beta} \times I_{cont,\beta}(z,\rho) \times \iint_{(u,v) \in Area(\beta)} K_{cont,\beta}(u-x,v-y,z,\rho)\,du\,dv \tag{4}$$

In the convolution, the calculation point $(\tilde{x},\tilde{y},\tilde{z})$ is represented by (x,y,z) relative to the origin of the beamlet coordinate system. The electron fluence $\Phi_{cont,\beta}$ and the energy deposition density $I_{cont,\beta}$ are assumed to be uniform over the cross-section of beamlet β.

The scatter kernel for contaminating electrons is modeled in a conventional way by a Gaussian distribution function:

$$K_{cont,\beta}(x,y,z,\rho) = \frac{1}{\pi\sigma_E^2}\exp\left[-\frac{x^2+y^2}{\sigma_E^2}\right] \quad (5)$$

where factor σ_E is a constant derived from measured data.

The fluence of the contaminating electrons is determined by convolving photon fluence with a Gaussian kernel with $\sigma = \sigma_{cont}$. The energy deposition density function $I_{cont,\beta}(z,\rho)$ for the contaminating electrons is determined from the measured data and tabulated as a function of the depth z.

Superposition

The final dose $D(\tilde{x},\tilde{y},\tilde{z})$ at an arbitrary calculation point in the patient is obtained by a superposition of the separate dose contributions from the primary photons (ph1) (Equation 1), extra-focal photons (ph2) (Equation 1), and contaminating electrons (Equation 4) from all individual beamlets denoted by index β:

$$D(\tilde{X},\tilde{Y},\tilde{Z}) = \sum_{\beta} \left(D_{ph1,\beta}(\tilde{X},\tilde{Y},\tilde{Z}) + D_{ph2,\beta}(\tilde{X},\tilde{Y},\tilde{Z}) + D_{cont,\beta}(\tilde{X},\tilde{Y},\tilde{Z})\right) \quad (6)$$

Essentially, the majority of the convolutions appearing in the superposition can be performed analytically, because the scatter kernels have Gaussian shapes, and because the photon and

electron fluences, as well as the energy deposition functions, can be treated as uniform across the cross-sections of the beamlets without introducing an significant error in the final dose distribution.

MU Calculation

Calculation of monitor units is based on the calibration measurements made with the smallest and largest field sizes and a selection of rectangular field sizes in between. The head scatter effects are Monte Carlo simulated for Varian accelerators using an extra-focal photon source. Phantom scatter factors are calculated by photon transport. Backscatter effects to the MU chamber are determined from the output factor table. Accelerators from other vendors are also supported because the Monte Carlo simulated data merely serves as a starting point for the beam configuration procedures that fit the beam parameters to the measured beam data.

Implementation in Eclipse

The new AAA algorithm is implemented as a dose calculation server in Eclipse. The AAA algorithm is configured in the Eclipse Beam Configuration task. The resolution of the dose calculation grid can be selected in the range of 2–10 mm during treatment planning in the Eclipse External Beam Planning task.

A fundamental model of the radiation generated by the medical linear accelerator is first derived using Monte Carlo simulations of the treatment head. Then for each clinical beam, the Monte Carlo phase space parameters are modified to construct a customized phase space specific to the clinical beam to be modeled. The customized phase space defines the particle fluence and energy spectrum characteristic of the clinical beam. The dose calculation with the AAA supports the use of beam modifiers, such as blocks, compensators, hard wedges, dynamic wedges, MLCs and Intensity Modulated Radiation Therapy (IMRT) with Dynamic MLC (DMLC).

Required Beam Measurements

The basic physical parameters are predefined by Monte Carlo simulations and adapted to the available beam data measured in a water equivalent medium. The minimum set of beam measurements for AAA beam data configuration includes:

- Open field depth doses for three small square field sizes between 3×3 cm^2 and 6×6 cm^2 for 10×10 cm^2 field and for the largest field size.
- Open field lateral dose profiles for five of the field sizes mentioned above. For the largest field size, the diagonal dose profiles at the five depths are also required.
- Wedged field depth dose for the largest wedged square field size.
- Wedged field dose profile at the depth of dose maximum for the largest wedged square field size.
- MU-to-dose calibration table for the user selected calibration depth and Source-to-Phantom Distance (SPD). Measurements are required for the smallest and the largest square fields and for the two extreme rectangular fields (for example, 3×40 cm^2 and 40×3 cm^2).

In general, the beam data required to configure the Pencil Beam Convolution model in Eclipse (or the Single Pencil Beam model in CadPlan™) is sufficient for the configuration of the AAA model. Therefore, there are usually no additional measurements required when implementing AAA for treatment units that are already commissioned in Eclipse.

Results

A large number of experiments have been carried out to verify the accuracy of the implementation of the AAA dose calculation model in Eclipse. Testing has included a full range of field sizes from 3×3 to 40×40 cm^2, both in homogeneous and inhomogeneous phantoms. The measurements presented in this paper are part of the Golden Beam Data for Varian Clinacs for 6 and 18 MV photon energies. Additionally Monte Carlo calculations were conducted in test phantoms in a slab geometry and compared to calculations with the AAA dose calculation model.

The open field dose calculation in a homogeneous water-equivalent phantom is in excellent agreement with measurement and is shown for 6 MV (Figures 7 – 9) and for 18 MV (Figures 10 – 12). Deviations are well within ± 1 % of the dose maximum for all the field sizes tested from 4×4 to 40×40 cm^2 with 6 and 18 MV energies.

Heterogeneity corrections were studied with small field sizes and low to medium densities, as the influence of strong scattering effects on the dose distribution is most prominent under those conditions. Figures 13 to 16 show depth-dose curves for various field sizes in water with (1) a 8 cm thick Styrofoam layer embedded at 2 to 10 cm depth to simulate lung interfaces and (2) a 2 cm thick bone slab embedded at 5 to 7 cm depth. The data are for 6 MV and 18 MV photon energies and are compared to data generated by Monte Carlo calculations. The AAA model accurately predicts both the dose build-down effect at the water-lung interface and the dose build-up effect at the lung-water interface. Table 1 summarizes the results to a 95 % confidence level.

Table 1: Summary of results

Phantom Type	**Criteria**		
	Dose	**Distance**	**Region**
Homogeneous	< 1 %	< 1 mm	Beyond d_{max}
	< 2 %	< 2 mm	Below d_{max}
Heterogeneous	< 3 %	< 2 mm	Inside heterogeneity
	< 1 %	< 1 mm	Outside heterogeneity

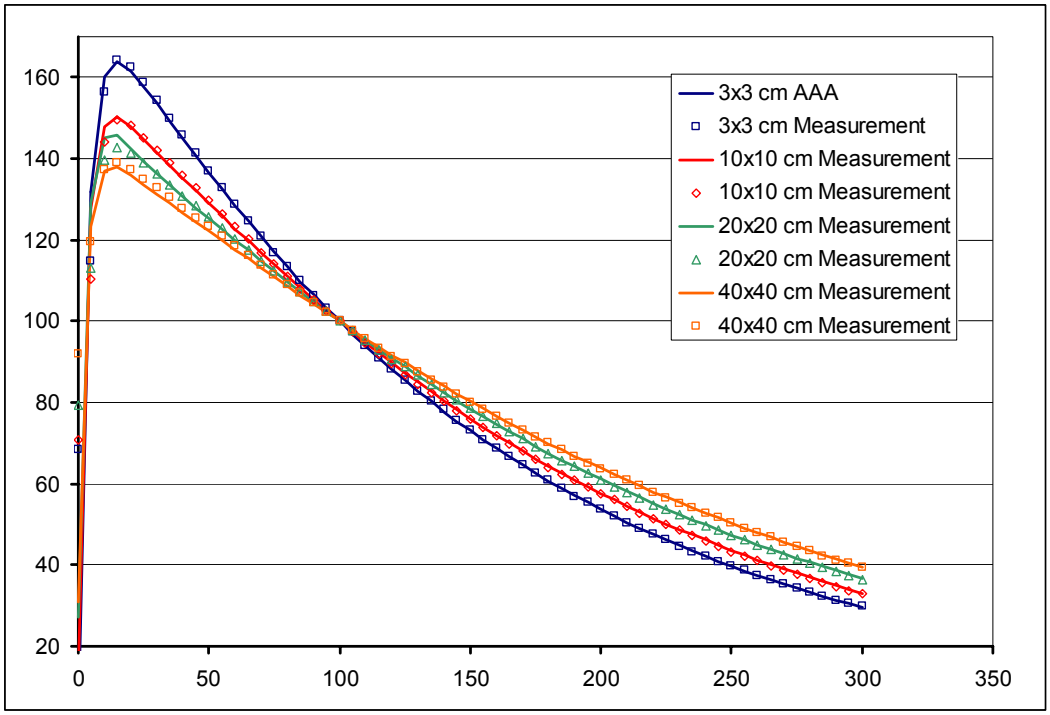

Figure 7: Percentage Depth Doses for 6 MV

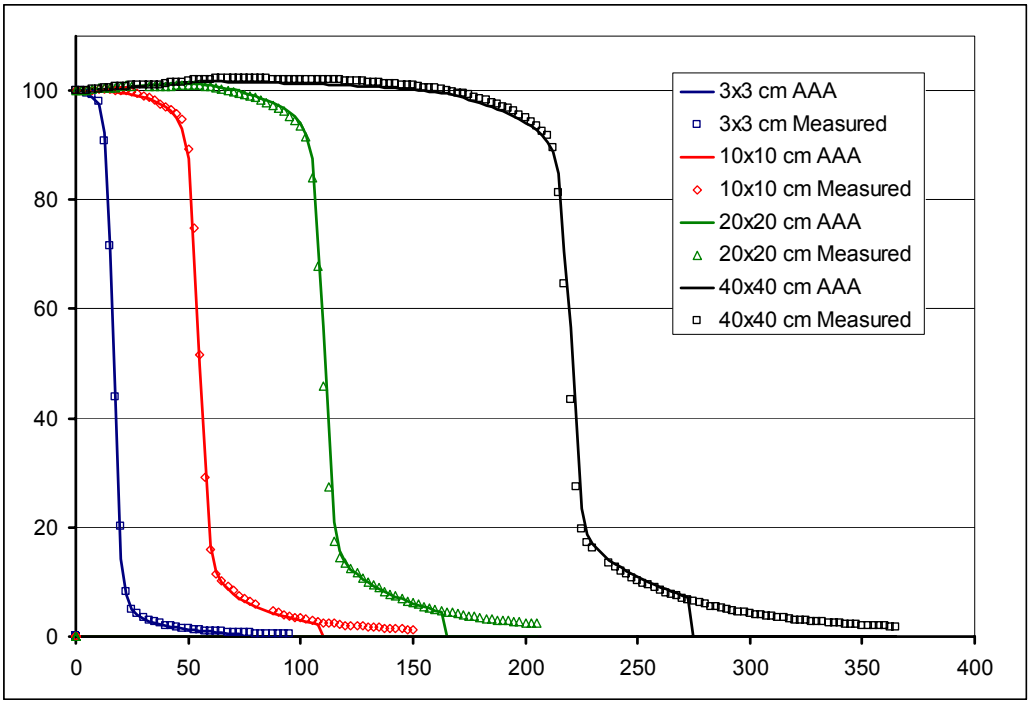

Figure 8: Profiles for depth 10 cm for 6 MV

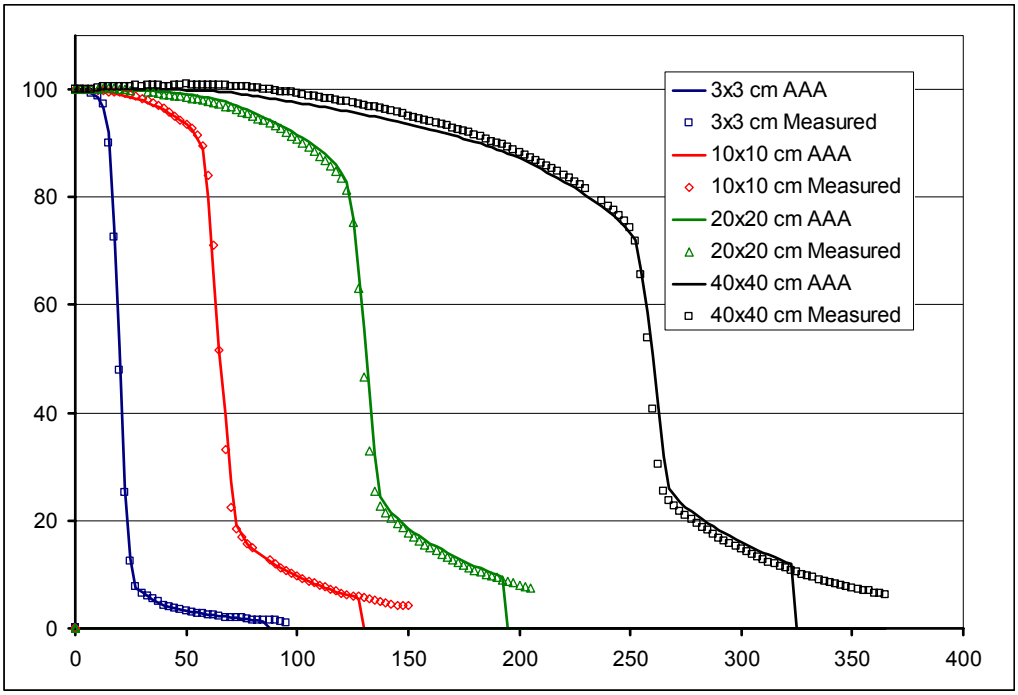

Figure 9: Profiles for depth 30 cm for 6 MV

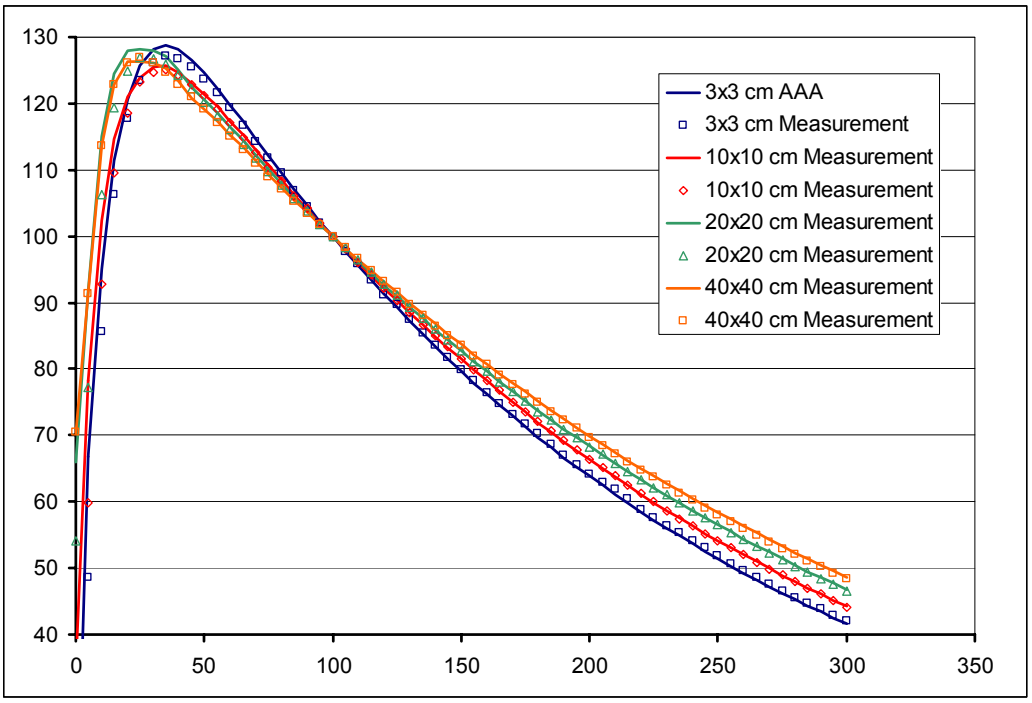

Figure 10: Percentage Depth Doses for 18 MV

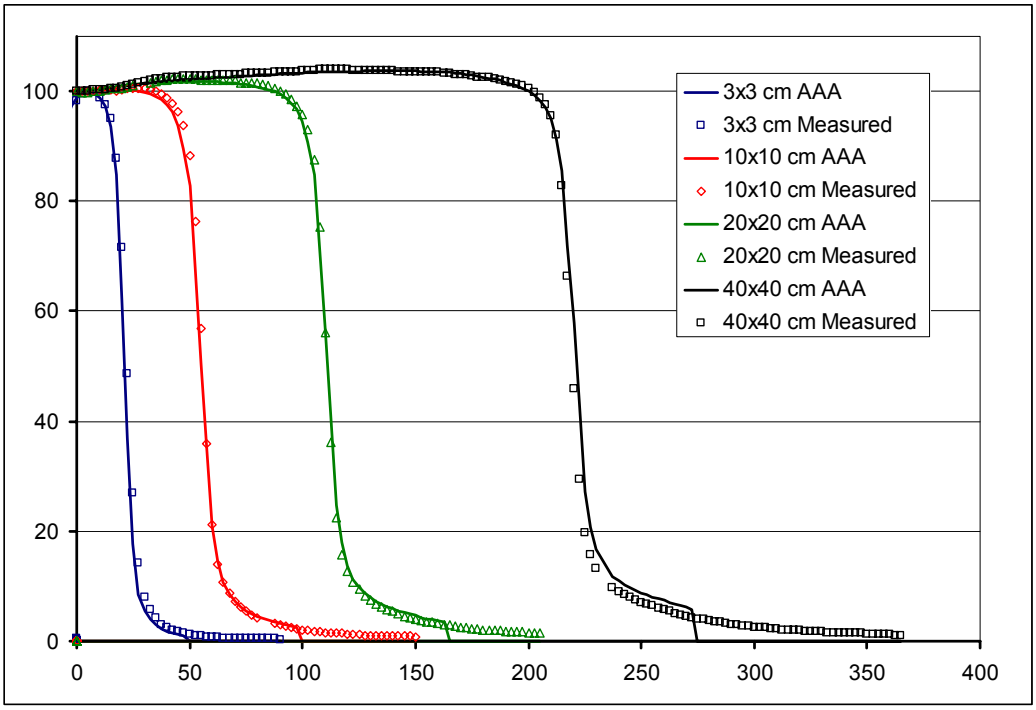

Figure 11: Profiles for depth 10 cm for 18 MV

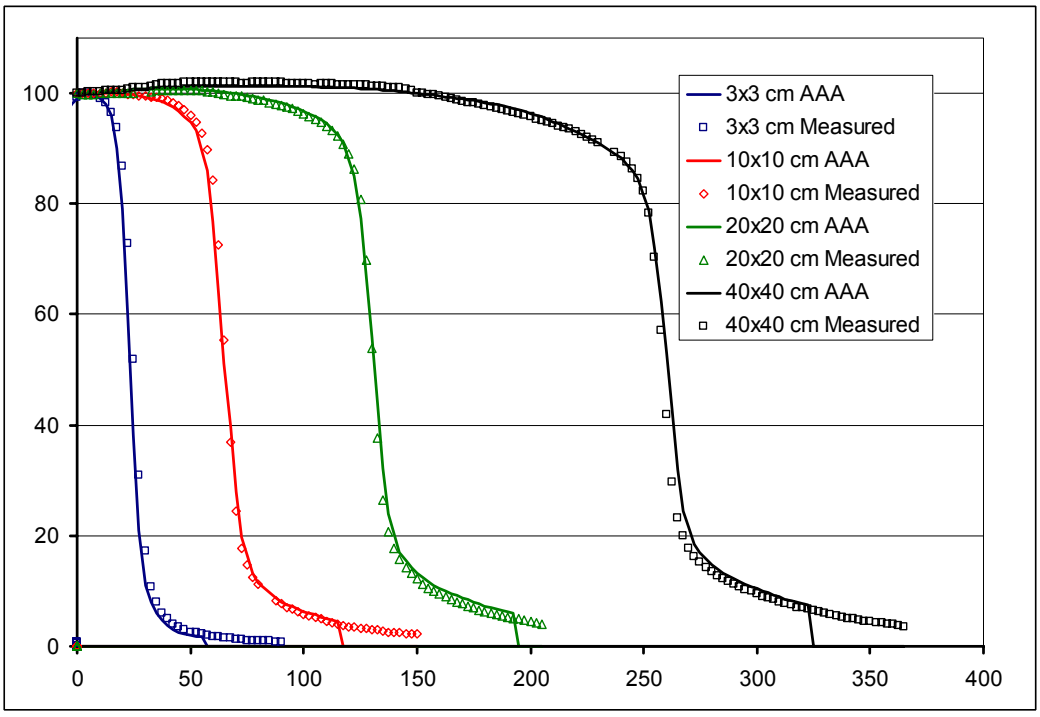

Figure 12: Profiles for depth 30 cm for 18 MV

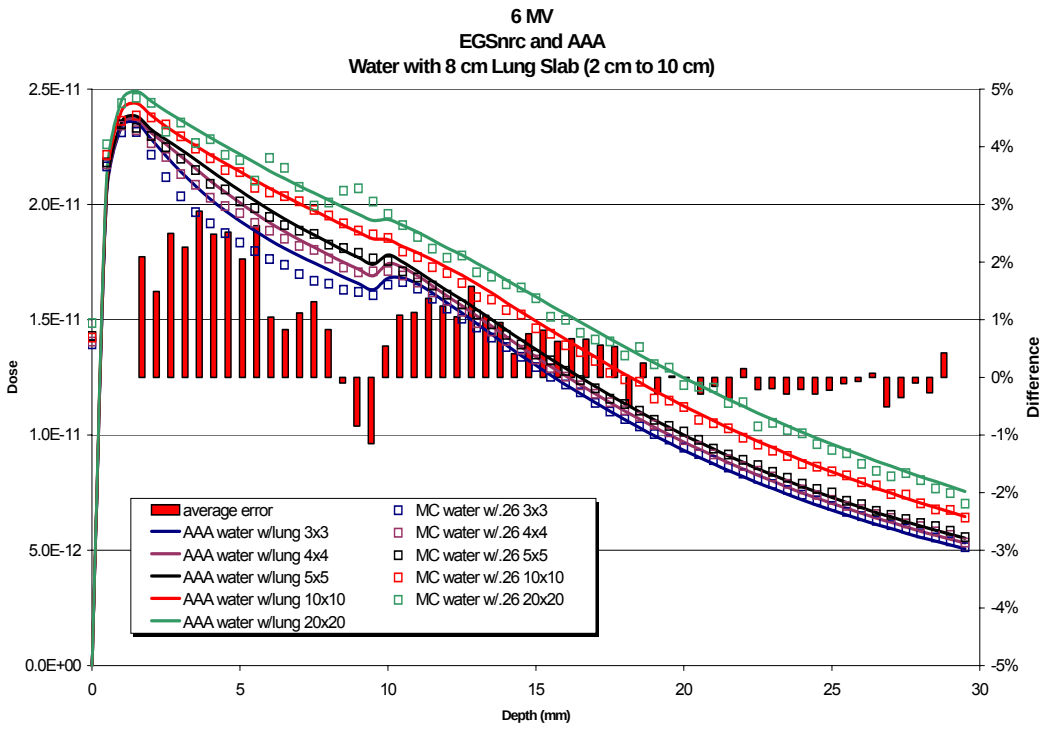

Figure 13: Percentage Depth Doses for 6 MV with Lung

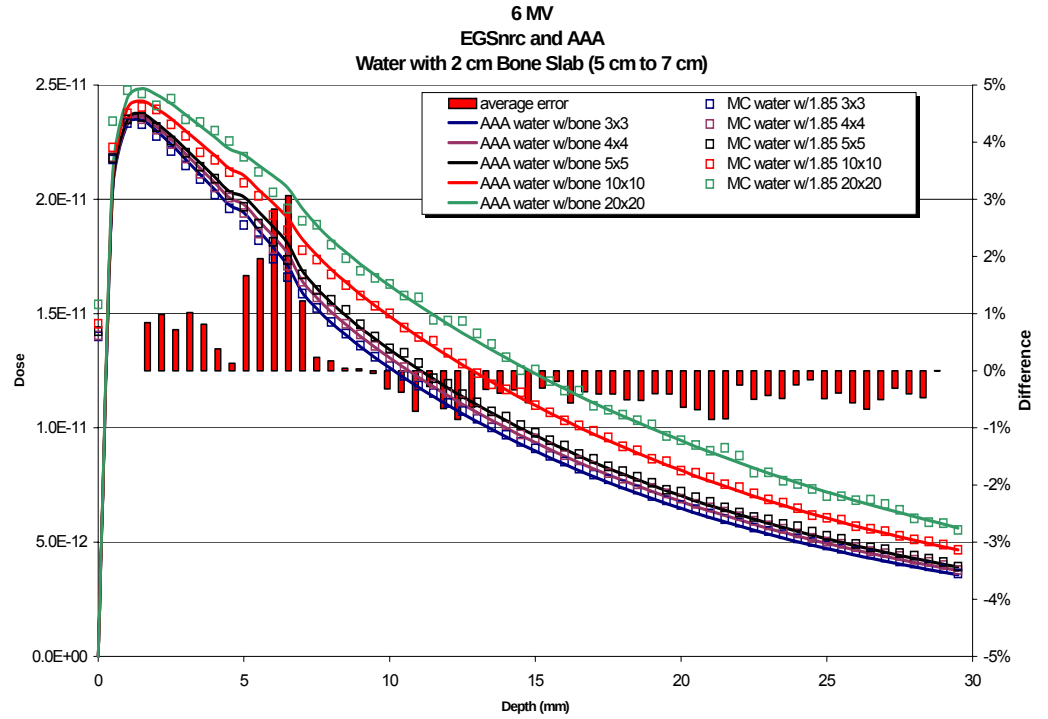

Figure 14: Percentage Depth Doses for 6 MV with Bone

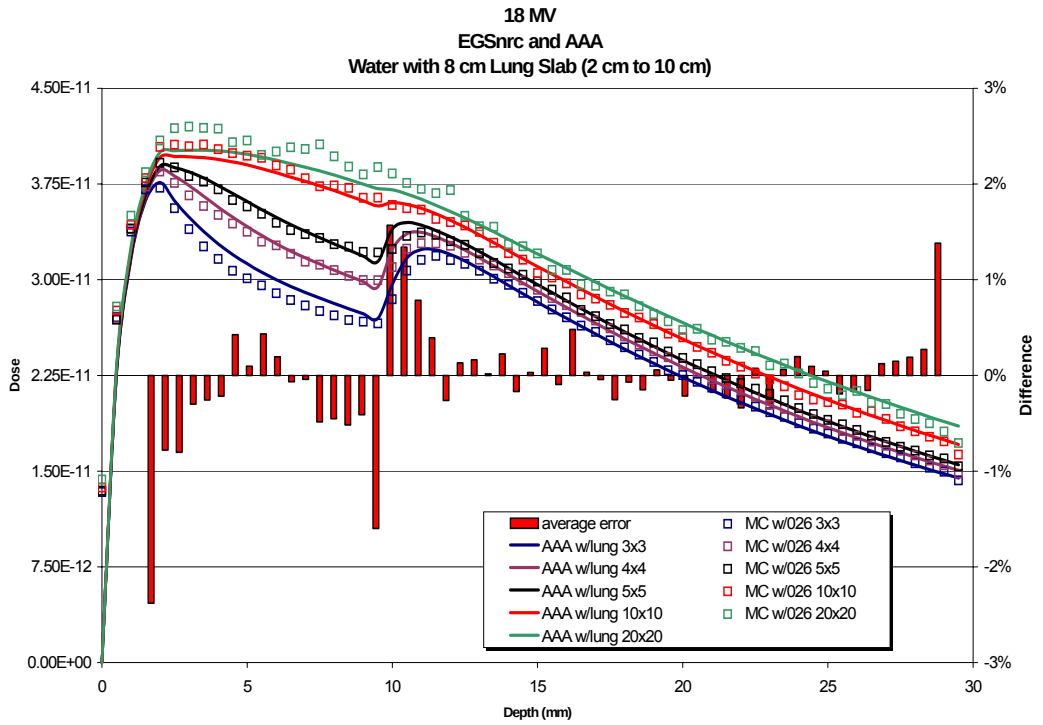

Figure 15: Percentage Depth Doses for 18 MV with Lung

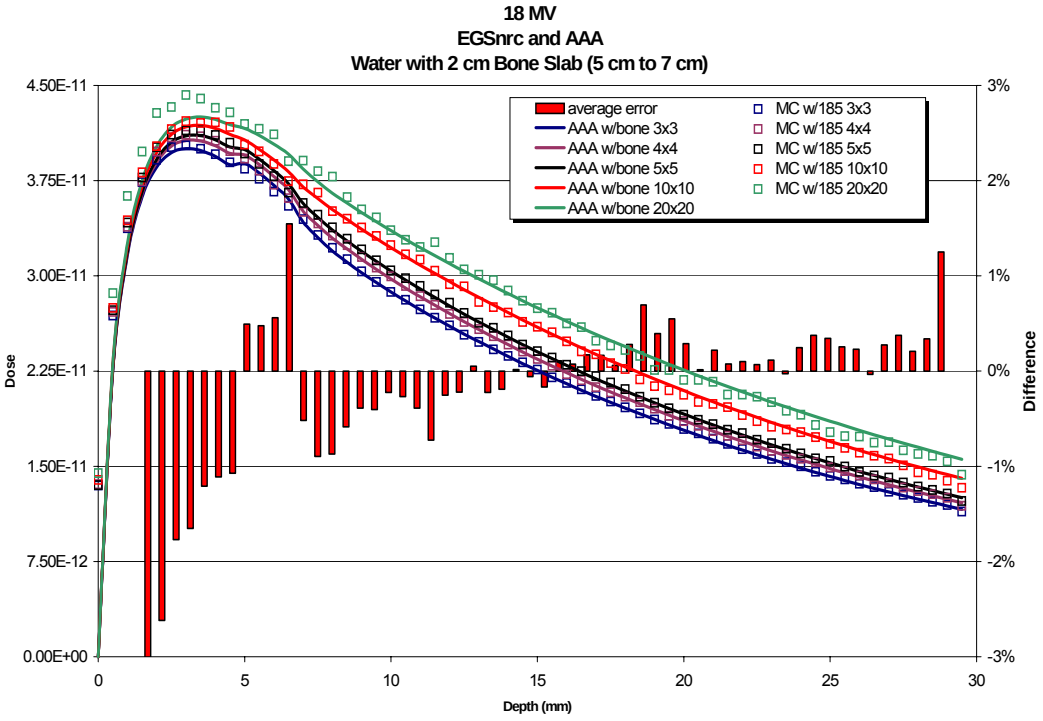

Figure 16: Percentage Depth Doses for 18 MV with Bone.

Conclusion

The AAA photon dose calculation model is a powerful tool for clinical radiotherapy planning that provides the increased speed and accuracy required for complicated modern treatment techniques. The AAA model also introduces a novel, physical method for modeling electron disequilibrium in tissue heterogeneities. The dose calculation accuracy has been tested by measurements in heterogeneous slab phantoms and the results demonstrate very good agreement with calculations. In general the agreement is better than ± 1.5 % of the dose maximum or ± 2 mm lateral shift of isodose lines in high dose gradient regions.

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