



Dose calculations

Collapsed Cone Convolution Superposition and Delta Pixel Beam

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Introduction

The Pinnacle³ 3D treatment planning system uses a Collapsed Cone Convolution Superposition (CCCS) computation to determine the dose distribution in patients from external photon beams. The CCCS dose model is a true three-dimensional dose computation that intrinsically handles the effects of patient heterogeneities on both primary and secondary scattered radiation. This computation method is inherently able to account for dose distributions in areas where the electronic equilibrium is perturbed, such as tissue-air interfaces and tissue-bone interfaces.

Intensity modulated radiation therapy (IMRT) requires both fast and accurate dose calculation. For this reason, a hybrid dose calculation of the CCCS and a finite pencil beam technique, Delta Pixel Beam, are used to maintain CCCS accuracy while providing speed for IMRT optimization.

The Convolution Superposition dose model

The Pinnacle³ CCCS dose algorithm is based on the work of Mackie, et al. Rather than correcting measured dose distributions, the CCCS algorithm computes dose distributions from first principles and, therefore, can account for the effects of beam modifiers, the external patient contour, and tissue heterogeneities on the dose distribution.

The CCCS dose model consists of four parts:

- Modeling the incident energy fluence as it exits the accelerator head
- Projection of this energy fluence through the density representation of a patient to compute a total energy released per unit mass (TERMA) volume
- A three-dimensional superposition of the TERMA with an energy deposition kernel using a ray-tracing technique to incorporate the effects of heterogeneities on lateral scatter
- Electron contamination modeled with an exponential falloff, added to the dose distribution after the photon dose is computed

The following sections describe each part of the model in more detail.

Modeling the incident energy fluence as it exits the accelerator

The incident energy fluence distribution is modeled as a two-dimensional array which describes the radiation exiting the head of the linear accelerator. The parameters defining this array are defined during physics data modeling.

The starting point for photon modeling is a uniform plane of energy fluence describing the intensity of the radiation exiting the accelerator head. The fluence model is then adjusted to account for the flattening filter, the accelerator head, and beam modifiers such as blocks, wedges, and compensators.

- The “horns” in the beam produced by the flattening filter are modeled by removing an inverted cone from the distribution.
- Off-focus scatter produced in the accelerator head is modeled by defining a 2D Gaussian function as a scatter source and adjusting the incident energy fluence based on the portion of the Gaussian distribution visible from each point in the incident energy fluence plane.
- The geometric penumbra is modeled by convolving the array with a focal spot blurring function.
- During planning, the shape of the field produced by blocks or multi-leaf collimators is cut out of the array, leaving behind the corresponding transmission through the shape-defining entity.
- Beam modifiers, such as wedges and compensators, are included in the array by attenuating the energy fluence by the corresponding thickness of the modifier. For static wedges and compensators, a radiological depth array is also stored, which allows for proper modeling of the beam hardening due to the presence of the beam modifiers during the projection of the incident fluence array.

Dynamic beam delivery with intensity modulation or dynamic wedges is easily handled using the incident energy fluence array. For these beams, the radiological depth array is not needed to account for beam hardening.

Projection of energy fluence through a CT patient representation

The incident energy fluence plane is projected through the CT patient representation and attenuated using mass attenuation coefficients. These coefficients are stored in a three-dimensional lookup table as a function of density, radiological depth, and off-axis angle. Patient heterogeneities are taken into account with the density dependence. Beam hardening through the patient is accounted for with the radiological depth dependence, and the off-axis softening of the energy spectrum is produced with the off-axis angle dependence. To account for the changes in the photon energy spectrum at different locations in the beam, the mass attenuation coefficient lookup table is produced using a weighted sum of several mono-energetic tables.

The TERMA volume is computed by projecting the incident energy fluence through the patient density volume using a ray-tracing technique. A given ray's direction is determined based on the position of the radiation source and the particular location in the incident fluence plane. At each voxel in the ray path, the TERMA is computed using the attenuated energy fluence along the ray and the mass attenuation coefficient at the particular density, radiological depth, and off-axis angle.

3D superposition of an energy deposition kernel

The three-dimensional dose distribution in the patient is computed by superposition of the TERMA volume with the energy deposition kernel. The kernel represents the spread of energy from the primary photon interaction site throughout the associated volume. Poly-energetic kernels are produced by combining a series of Monte Carlo-generated mono-energetic energy deposition kernels. The superposition is carried out using a ray-tracing technique similar to that used in the projection of the incident energy fluence. The kernel is inverted so that the dose can be computed in only a portion of the patient (TERMA) volume if desired. This allows for point dose computation and decreases computation time.

The rays from the dose deposition site are cast in three dimensions. At each voxel of the TERMA traversed along a ray, the contribution of dose to the dose deposition site is computed and accumulated using the TERMA and the kernel value at the current radiological distance. Using the radiological distance along the ray also allows the kernel to be scaled to account for the presence of heterogeneities with respect to scattered radiation in all directions.

The dose computation described above determines the dose from a single beam. Multiple beams are computed independently and the entire 3D dose distribution is created by adding the dose from each beam together according to the corresponding beam weight.

Adaptive Convolution Superposition

An Adaptive Convolution Superposition approach has also been implemented in Pinnacle³. This uses the calculation technique described above with some slight modifications. The speed of the computation is increased by adaptively varying the resolution of the dose computation grid depending on the curvature of the TERMA and dose distribution. First, the dose in a coarse 3D grid is computed and then the curvature in the TERMA distribution is assessed. In regions where the curvature is high, the dose is computed at intermediate points to provide higher resolution. The system adaptively increases the resolution in regions of high curvature until an acceptable resolution is used. In regions of low curvature, the dose is interpolated from the coarse dose grid. This technique decreases the computation time by a factor of 2-3 without compromising the accuracy of the CCCS calculation in the presence of heterogeneities.

The Delta Pixel Beam dose for optimization of IMRT

During the iterative optimization process to determine the appropriate intensity modulation for the treatment beams for IMRT, it is necessary to repeatedly compute dose. In order to have an efficient optimization, the dose computation must be fast. Most approaches to the high speed iterative dose computation sacrifice dose accuracy for speed. The Delta Pixel Beam method provides high speed dose computation without sacrificing the accuracy of the CCCS.

The Delta Pixel Beam dose computation is a hybrid between a finite-sized pencil beam algorithm and the CCCS method. The basic method is to first optimize the IMRT treatment to an intermediate solution using the pencil beam algorithm. Then the full CCCS dose is computed for each beam. For the remainder of the optimization, the dose is updated by first determining the change in each beam's intensity distribution. Then, the perturbation in the dose from the change in the intensity distribution is determined with the pencil beam algorithm. This dose perturbation is then used to update the base CCCS dose. This technique provides the fast dose computation needed for the optimization while also maintaining required accuracy in the presence of heterogeneities to insure that the optimal dose distribution can be realized in the patient.

The pencil beam algorithm is based on the model for the CCCS and requires no additional modeling beyond the CCCS modeling process in Pinnacle³.

This paper provides an overview of the Convolution Superposition dose computation method used in the Philips Pinnacle³ 3D treatment planning system. For further information on this method and other dose computation techniques, please refer to bibliography.

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