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Implementation of Monte Carlo Dose Calculation for CyberKnife treatment planning

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Abstract. Accurate dose calculation is essential to advanced stereotactic radiosurgery (SRS) and stereotactic radiotherapy (SRT) especially for treatment planning involving heterogeneous patient anatomy. This paper describes the implementation of a fast Monte Carlo dose calculation algorithm in SRS/SRT treatment planning for the CyberKnife® SRS/SRT system. A superposition Monte Carlo algorithm is developed for this application. Photon mean free paths and interaction types for different materials and energies as well as the tracks of secondary electrons are pre-simulated using the MCSIM system. Photon interaction forcing and splitting are applied to the source photons in the patient calculation and the pre-simulated electron tracks are repeated with proper corrections based on the tissue density and electron stopping powers. Electron energy is deposited along the tracks and accumulated in the simulation geometry. Scattered and bremsstrahlung photons are transported, after applying the Russian roulette technique, in the same way as the primary photons. Dose calculations are compared with full Monte Carlo simulations performed using EGS4/MCSIM and the CyberKnife treatment planning system (TPS) for lung, head & neck and liver treatments. Comparisons with full Monte Carlo simulations show excellent agreement (within 0.5%). More than 10% differences in the target dose are found between Monte Carlo simulations and the CyberKnife TPS for SRS/SRT lung treatment while negligible differences are shown in head and neck and liver for the cases investigated. The calculation time using our superposition Monte Carlo algorithm is reduced up to 62 times (46 times on average for 10 typical clinical cases) compared to full Monte Carlo simulations. SRS/SRT dose distributions calculated by simple dose algorithms may be significantly overestimated for small lung target volumes, which can be improved by accurate Monte Carlo dose calculations.

1. Introduction
The Monte Carlo method has been demonstrated to be the most accurate dose calculation method for radiation therapy treatment planning and dosimetry verification with significant differences shown between Monte Carlo calculated dose distributions and those calculated by conventional dose calculation algorithms [1-7]. The commissioning and clinical validation of commercial electron and photon algorithms have been performed by several investigators [8-10] and a number of IMRT (intensity-modulated radiotherapy) treatments planned using Monte Carlo dose calculation were reported [11]. Because Monte Carlo algorithms can accurately calculate dose distributions for complex beam delivery configurations and heterogeneous patient geometry, and because rapidly increasing computing power and better computational algorithms have reduced the dose calculation time to a clinically acceptable level, Monte Carlo is expected to become widely used for dose calculation in...
radiation therapy treatment planning. One special procedure of radiation therapy is stereotactic radiosurgery (SRS) and stereotactic radiotherapy (SRT). For extracranial SRS/SRT, especially for the treatment of small target volumes in lung the use of Monte Carlo dose calculation may significantly improve the dosimetry accuracy [12-13].

In this work, we have implemented Monte Carlo dose calculation for SRS/SRT treatment planning using the CyberKnife® system (Accuray, Inc., Sunnyvale, CA), which consists of a 6-MV X-band linear accelerator installed on a robotic arm. In order to implement Monte Carlo simulations for routine clinical dose calculation we developed accurate and efficient algorithms for tracking particle transport and scoring energy deposition in heterogeneous patient geometry. We also developed practical photon source models and beam commissioning procedures to facilitate the widespread application of Monte Carlo dose calculation for CyberKnife SRS/SRT treatment planning. This paper describes the source model, the beam commissioning procedure and the patient dose calculation algorithm. Phantom dose distributions and patient treatment plans are compared between the new Monte Carlo dose calculation algorithm, the CyberKnife treatment planning system (TPS) and measurements.

2. Materials and Methods

2.1. Source modelling and beam commissioning
A multiple source model has been developed for commonly used clinical linear accelerators [14-17]. Since the CyberKnife does not have a flattening filter, we have developed a single source model for the phase space presentation and reconstruction of the CyberKnife 6 MV photon beams for accurate and efficient phantom and patient dose calculations. The source model and its parameters were initially derived from the Monte Carlo simulated phase space data of an actual CyberKnife system that ensure accurate regeneration of dose distributions, which will match the measured dose distributions. The source model parameters can be modified automatically in the beam commissioning process (see section 2.1.4) to match the measured dose distribution for other CyberKnife systems.

2.1.1. The Monte Carlo simulation. The CyberKnife SRS/SRT system was first simulated using the NRCC BEAM Monte Carlo code system [18]. The phase space data of 6MV photon beams for all the 12 collimation cones were obtained based on the geometric and material specifications provided by the vendor. The incident electron energy was adjusted in the simulation to match the measured depth dose curves (PDD), which were also provided by the vendor. After many iterations of trial and error with the electron beam parameters, the final electron energy was determined to be 6.8 MeV. A Gaussian beam profile was used for the incident electron beam at the target surface and the Gaussian parameters were adjusted to match the measured beam profiles at different phantom depths (with $\sigma = 1.15$ mm). The phantom dose calculation was performed using the MCSIM Monte Carlo dose calculation system with a standard set of transport parameters and energy cutoffs [19-20].

2.1.2. Source model presentation. The source model used in this study to reconstruct phase space consists of a single photon source located at the target level of the Cyberknife treatment head with a spatial and energy distribution. The photon source is assumed to be cylindrically symmetrical with a planar fluence distribution as a function of the off-axis distance that can be determined based on the measured in-air cone output factors. The energy spectrum of the photon source is determined from the measured central axis PDD in water using a 60 mm cone defined at 80 cm SSD. The planar fluence distribution at the patient plane is also cylindrically symmetrical that is derived from the measured profile at $d_{max}$ without the cone collimator. The dose contribution of the contaminant electrons is found to be in the order of $10^3$ of the maximum dose based on Monte Carlo simulations and therefore is omitted in the phase space representation and reconstruction.
2.1.3. Phase-space reconstruction 

Based on the phase-space presentation for the CyberKnife source model, one can reconstruct the phase space of the CyberKnife 6-MV photon beam and use it as source input for forward Monte Carlo dose calculations. As shown in figure 1, the sampling plane is located at the middle of the cone collimator while the phase space reconstruction plane can be anywhere above the patient/phantom surface. The sampling plane is divided into two parts, inside the cone inner surface (the collimator opening) and from the cone inner surface to 5 cm in radius (the cone material). A cone transmission factor is used to adjust the weight of the particle that passes through the cone material. The starting point of the particle on the source plane is sampled using the source planar distribution. The direction of motion of the particle is determined by the line linking the starting point on the source plane to the point sampled at the collimator plane. The particle is weighted by the planar fluence distribution at 80 cm SSD to match the measured dose distribution. Finally, the energy of the particle is sampled from the source energy distribution. The spatial variation of the photon energy spectrum on the phase space reconstruction plane is negligible since there is no flattening filter in the Cyberknife unit.

Two important modifications are made for cone collimators of different radii. The photon energy spectrum derived from the 60 mm cone PDD is used for all the cone collimators except for the 5 mm cone. Annihilation photons with energies about 0.511 MeV have a significant contribution to the final dose for the 5 mm cone, reducing the average energy of the energy spectrum by about 4% (figure 2). Another special situation is the collimator shape. According to the design, the inner surfaces of the larger cones (> 7.5 mm diameter) are divergent, which focus on the target position while for the 5 mm and 7.5 mm collimators, the inner surfaces are parallel (straight tubes). The radius of the collimator opening at the sampling plane has to be reduced accordingly since the lower part of the collimator will block some of the particles (i.e., resulting in partial transmission).

2.1.4. Beam commissioning 

An automatic beam commissioning procedure has been investigated and software developed to derive source model parameters automatically based on measured beam data, which can be used for efficient and accurate Monte Carlo dose calculation for the CyberKnife system. Our commissioning procedure assumes a single photon source located at the target location of the CyberKnife treatment head. The photon source is assumed to be cylindrically symmetrical with a source distribution, which will be determined based on the measured in-air cone output factors. The energy spectrum of the photon source will be determined from the measured central axis PDD in water for a 60mm cone defined at 80 cm SSD. The fluence distribution of the photon source is obtained from the measured profile at the depth of maximum dose without any cones (secondary collimators). With the energy spectrum, the fluence and source distributions known, one can reconstruct the beam phase
space one particle at a time with a beam sampling routine, and use the reconstructed phase space as source input for forward Monte Carlo dose calculations in the patient geometry.

For the PDD data, the depths required are from 0 to 30 cm in water. For the in-air output factor data, all the 12 cones are measured at 80 cm source-to-detector distance (SDD) and normalized to the 60 mm cone reading. For the profile data, the cone is removed before taking the measurement. Assuming good symmetry for the cones used in the CyberKnife system, only profile along one major axis is needed. The scan range is from –8.0 cm to 8.0 cm. The depth at which the scan is performed is at the depth of the maximum dose to avoid the effect of electron contamination (though we found it <0.1%).

2.2. Dose calculation

2.2.1. The code

The code The MCRS code, which is the software for Monte Carlo dose calculation for the CyberKnife radiosurgery system, is designed for dose calculations in a 3D rectilinear voxel (volume elements) geometry that is used to model the patient heterogeneous anatomy. Every voxel can be assigned to a different material and its density can be variable though the density-effect corrections for the stopping powers of the material remain unchanged. The conversion of the CT numbers (or electron density values) to materials and mass densities has been described previously [7,20].

The beam information, such as the monitor units (MU), the cone size and the coordinates of the node and target, is taken from the Accuray CyberKnife treatment plan XML file. The node coordinates (x, y, z in mm) and target coordinates (x, y, z in mm) are defined in the patient (CT) coordinate system. MCRS will read the beam information and discard all the beams with a zero MU. The rotation matrix with reference to the primary direction (z direction) for the non-zero-MU beam is then calculated based on the node and target coordinates before starting the dose calculation.

MCRS can read particle phase space information directly from a phase space file generated by other Monte Carlo beam simulation code, such as BEAM [18] and MCBEAM [21]. It can also reconstruct the phase space parameters of all the particles in a beam at the sampling plane defined at 80cm SSD, perpendicular to the beam axis from a source model (see above) that can be commissioned based on measured beam data.

2.2.2. Variance reduction techniques

In order to make the simulation more efficient, MCRS utilizes pre-simulated interaction/track data. The mean free paths in 4 basic materials (air, tissue, bone and water), major interaction types, scatter photon information as well as the electron track information in water were recorded in a binary tree for mono-energetic photons from 25 keV to 7.7 MeV during Monte Carlo simulations using a benchmarked Monte Carlo code MCSIM [19,20]. The number of photons for each energy (bin) is between 1000 and 10000 based on the spectral distribution of the CyberKnife 6 MV photon beam. The hard disk storage space for the interaction data is about 50 Megabytes, which will be stored directly in the memory during Monte Carlo dose calculation.

In a photon beam simulation, many photons may penetrate through the patient without any contributions to the dose distribution. To increase the simulation efficiency, the incident photon is split into N photons and forced to interact along the ray line of the incident photon. The weights of the resultant particles are reduced N times first and then further reduced more according to the probability for the photon to interact in the geometry. The locations of the N photon interactions are randomly

![Dose Profile at Depth of Maximum Dose](image)
sampled based on the interaction probability distribution between the entry point and exit point, which is calculated based on the mean free path. N was set to 90 for this work. The interaction type and the secondary photon and electron parameters are taken from the recorded data.

As a result of the photon splitting and interaction forcing, many scattered photons with small weights are generated and they make little contributions to the target dose distribution. It is more efficient to perform Russian roulette to these scattered photons so that, on average, only one scattered photon will survive for every original incident photon. The scattered photon will be forced to interact and split again along its ray line in the same way as described above. This will continue until all the resultant particles are terminated when their energies are below a predetermined cutoff or when they move out of the simulation geometry defined by the patient external contours.

The resultant electrons from photon interactions are simulated by repeating the recorded electron tracks. The electron step length is stretched/shrunken proportionally based on the local density and modified based on the stopping power ratios and scattering power ratios of the local material with respect to energy deposition and electron multiple scattering [19,20]. This way, the electron tracks can be repeated in both soft tissue (with variable density) and other materials such as bone and air (with variable density).

Energy deposition in a voxel is proportional to the fraction of the equivalent step length in that voxel. The dose of every voxel is calculated by dividing the accumulated energy deposition by the mass of the voxel. History-based relative statistical uncertainty estimation was employed.
3. Results and Discussion

3.1. Phantom verification

Using Monte Carlo simulated phase space data and source models with parameters derived from measurement data extensive dose calculations in a water phantom were performed using MCRS and MCSIM. The Monte Carlo results were compared with measured data. Good (<1% of maximum dose) agreement was achieved in depth dose curves and beam profiles for all the cones between Monte Carlo simulations and measurements. Figures 4 and 5 show depth dose curves and lateral dose profiles in water for a 5 mm and a 12.5 mm collimator cone, respectively, calculated by Monte Carlo and measured using a diode detector (PTW 60008). Similar results were obtained for other collimator cones between phase space, source model and measurements [22,23]. The Monte Carlo calculated output factors in water were also compared with measured results using the diode detector for collimator cones of diameters from 5mm to 60mm. It can be seen in Table 1 that good agreement was achieved for cone sizes from 60mm down to 12.5mm (with 2% or less differences). The discrepancies between Monte Carlo and measurements were 3-4% for the three smallest cones, which were possibly caused by the effect of electron scattering at the metallic parts of the detector shielding [24].

Table 1: Output factors at an 1.5cm depth in water for different collimator cone sizes defined at 80 cm SSD calculated by MCRS and by measurements.

<table>
<thead>
<tr>
<th>D(mm)</th>
<th>Meas</th>
<th>MC</th>
<th>Diff (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>5</td>
<td>0.714</td>
<td>0.684</td>
<td>-4.2</td>
</tr>
<tr>
<td>7.5</td>
<td>0.876</td>
<td>0.838</td>
<td>-4.3</td>
</tr>
<tr>
<td>10</td>
<td>0.915</td>
<td>0.884</td>
<td>-3.4</td>
</tr>
<tr>
<td>12.5</td>
<td>0.945</td>
<td>0.926</td>
<td>-2.0</td>
</tr>
<tr>
<td>15</td>
<td>0.961</td>
<td>0.949</td>
<td>-1.2</td>
</tr>
<tr>
<td>20</td>
<td>0.976</td>
<td>0.970</td>
<td>-0.7</td>
</tr>
<tr>
<td>25</td>
<td>0.983</td>
<td>0.982</td>
<td>-0.1</td>
</tr>
<tr>
<td>30</td>
<td>0.987</td>
<td>0.986</td>
<td>-0.1</td>
</tr>
<tr>
<td>35</td>
<td>0.989</td>
<td>0.993</td>
<td>0.4</td>
</tr>
<tr>
<td>40</td>
<td>0.992</td>
<td>0.994</td>
<td>0.2</td>
</tr>
<tr>
<td>50</td>
<td>0.996</td>
<td>0.999</td>
<td>0.4</td>
</tr>
<tr>
<td>60</td>
<td>1.0</td>
<td>1.0</td>
<td>0</td>
</tr>
</tbody>
</table>

3.2. Plan comparison

We have compared dose distributions of realistic patient treatment plans calculated using our new Monte Carlo code MCRS using source models and MCSIM using the simulated phase space data with the results of the CyberKnife treatment planning system. Figure 6 shows the isodose distributions and dose-volume histograms (DVH) of a lung treatment plan. The target volume was 92 cc treated with 126 beams using a 20mm cone. MCRS and MCSIM gave consistent results (dashed lines and symbols in the DVH plot). Large differences were observed in the target dose (up to 15% for some lung patients [22,23]) between Monte Carlo and the CyberKnife planning system, which employed a TMR-based dose calculation algorithm. The differences emphasize the demand of more accurate dose calculation methods than simple correction-based methods for SRT/SRS planning.

Figure 6: Comparison of isodose distributions (left) and dose volume histograms (right) for a lung treatment plan calculated by Monte Carlo simulations and the CyberKnife planning system. Thin lines in the isodose plots and dashed lines in the DVH plot were calculated using MCRS. Open circles were from full Monte Carlo simulations using the MCSIM code.
Figure 7: Isodose distributions (left) and DVHs (right) for a head & neck treatment plan calculated by Monte Carlo simulations and the CyberKnife planning system. Thin lines in the isodose plots and dashed lines in the DVH plot were calculated using MCRS. Open circles were from full Monte Carlo simulations using the MCSIM code.

Figure 7 compares isodose distributions and DVHs for a head and neck treatment plan between Monte Carlo simulations and the CyberKnife treatment planning system. The target volume was very small (< 1cc), which was treated with 123 beams using a 5 mm collimator cone to achieve a desired dose distribution. The tissue-air and tissue-bone interfaces did not seem to affect the target dose distributions. The CyberKnife results agreed with both MCRS and MCSIM to within 1-2%. Figure 8 compares isodose distributions and DVH curves for a liver treatment, where the target was very large (>800 cc). This treatment plan was designed to test the ability of the CyberKnife system for planning large targets that require the placement of many small beams to cover the target volume. A 60mm collimator cone was used in this case with 232 beams to achieve a uniform and conformal dose distribution to the large target volume. Again, the CyberKnife system generated consistent results with Monte Carlo simulations using MCRS; the differences in the liver dose were within 3% between MCRS and the CyberKnife system.

It should be noted that the CyberKnife planning system only calculates the dose in a user-defined 3D volume containing the target while Monte Carlo simulations generate dose distributions for the entire patient geometry. Therefore, we observed some additional differences in the isodose distribution outside the CyberKnife scoring region and in the DVH curve for large anatomic structures such as lungs and the whole body in figures 6-8.

Figure 8: Comparison of isodose distributions (left) and DVHs (right) for a liver treatment plan calculated by Monte Carlo simulations and the CyberKnife planning system. Thin lines in the isodose plots and dashed lines in the DVH plot were calculated using MCRS.
3.3. Computation efficiency

We have investigated the computation efficiency of the new Monte Carlo code MCRS by comparisons with full Monte Carlo simulations using MCSIM with and without variance reduction techniques for 5 treatment plans with two different voxel sizes. In all the cases, MCRS showed better computation efficiency than full Monte Carlo simulations without variance reduction techniques and MCSIM with variance reduction techniques. As can be seen in Table 2, the mean speed up factor for MCRS is 45.7 versus full Monte Carlo simulations (MCSIM without variance reduction techniques). On average, MCRS is 3.2 times faster than MCSIM with variance reduction techniques for the 10 cases. We also investigated the number of particle histories required for MCRS to achieve a 2% statistical uncertainty with respect to the number of voxels in the target volume in order to provide guidelines for routine clinical applications. Figure 9 shows the results based on the 10 cases investigated, which exhibits a linear relationship on a log-log plot.

Table 2. The CPU time required by MCRS and MCSIM with variance reduction techniques and without variance reduction techniques (full MC) to achieve a 2% statistical uncertainty. The speed up factors were calculated as ratios of the CPU times for full MC to those for MCRS and MCSIM and are shown in brackets. The target size, voxel size, number of voxels, cone size and number of beams are also given for each treatment.

<table>
<thead>
<tr>
<th>Case #</th>
<th>Site</th>
<th>Voxel size (mm)</th>
<th>Target size (cc)</th>
<th># of voxels</th>
<th>Cone size (mm)</th>
<th># of beams</th>
<th>Full MC CPU T (min)</th>
<th>MCSIM CPU T (min)</th>
<th>MCRS CPU T (min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Lung</td>
<td>3.9 x 3.9 x 3</td>
<td>4.03</td>
<td>88</td>
<td>12.5</td>
<td>158</td>
<td>57.0</td>
<td>3.5 (16.3)</td>
<td>1.6 (35.6)</td>
</tr>
<tr>
<td>2</td>
<td>Lung</td>
<td>1.95 x 1.95 x 3</td>
<td>4.72</td>
<td>413</td>
<td>12.5</td>
<td>158</td>
<td>228.9</td>
<td>17.8 (12.9)</td>
<td>5.7 (40.2)</td>
</tr>
<tr>
<td>3</td>
<td>Lung</td>
<td>3.9 x 3.9 x 3</td>
<td>91.6</td>
<td>2002</td>
<td>20</td>
<td>126</td>
<td>324.4</td>
<td>19.9 (16.3)</td>
<td>5.9 (55.0)</td>
</tr>
<tr>
<td>4</td>
<td>Lung</td>
<td>1.95 x 1.95 x 3</td>
<td>96.5</td>
<td>8436</td>
<td>20</td>
<td>126</td>
<td>842.4</td>
<td>79.2 (10.6)</td>
<td>18.8 (44.8)</td>
</tr>
<tr>
<td>5</td>
<td>Lung</td>
<td>3.9 x 3.9 x 3</td>
<td>62.1</td>
<td>1356</td>
<td>25</td>
<td>134</td>
<td>231.5</td>
<td>15.7 (14.7)</td>
<td>4.5 (51.4)</td>
</tr>
<tr>
<td>6</td>
<td>Lung</td>
<td>1.95 x 1.95 x 3</td>
<td>68.4</td>
<td>5977</td>
<td>25</td>
<td>134</td>
<td>710.6</td>
<td>43.7 (16.3)</td>
<td>13.4 (53.0)</td>
</tr>
<tr>
<td>7</td>
<td>Head</td>
<td>2.2 x 2.2 x 2</td>
<td>0.73</td>
<td>74</td>
<td>5</td>
<td>123</td>
<td>35.2</td>
<td>2.5 (14.1)</td>
<td>0.84 (41.9)</td>
</tr>
<tr>
<td>8</td>
<td>Head</td>
<td>1.1 x 1.1 x 2</td>
<td>0.86</td>
<td>351</td>
<td>5</td>
<td>123</td>
<td>112.9</td>
<td>8.0 (14.1)</td>
<td>2.6 (43.4)</td>
</tr>
<tr>
<td>9</td>
<td>Liver</td>
<td>3.9 x 3.9 x 1.25</td>
<td>817.0</td>
<td>42860</td>
<td>60</td>
<td>232</td>
<td>2379</td>
<td>171.2 (13.9)</td>
<td>80.1 (29.7)</td>
</tr>
<tr>
<td>10</td>
<td>Liver</td>
<td>1.95 x 1.95 x 1.25</td>
<td>839.7</td>
<td>176324</td>
<td>60</td>
<td>232</td>
<td>12911</td>
<td>807.8 (16.0)</td>
<td>207.8 (62.1)</td>
</tr>
</tbody>
</table>

The relationship between the CPU time and the number of voxels in the target volume is depicted in Fig. 10, which can be fitted to a power function. The power function is expressed as

\[ t_{CPU} = \alpha \times n_{\text{voxel}}^\beta \]  

where \( t_{CPU} \) is the CPU time in minute, \( n_{\text{voxel}} \) is the voxel number in the target volume. Parameter \( \beta \) has a similar value for the three codes. It equals 0.6492 for full Monte Carlo simulation (MCSIM without variance reduction techniques), 0.6849 for MCSIM with variance reduction techniques and 0.6544 for MCRS. Parameter \( \alpha \) is dependent on the machine used for calculation and the statistic uncertainty achieved. It equals 2.7784 for full Monte Carlo, 0.1505 for MCSIM and 0.0597 for MCRS for a 2%
statistical uncertainty using a PC with 3.2GHz Intel duel-core CPU and 2GB memory. This equation can be used to estimate the CPU time as well as the total history numbers needed for the dose calculation when it is implemented for routine clinical applications.

4. Conclusions
We have developed source modeling, beam commissioning and dose calculation software for the clinical implementation of Monte Carlo dose calculation for CyberKnife SRS/SRT treatment planning. Good agreement (<1%) was achieved between Monte Carlo dose calculations and measurements in PDD and dose profiles. Small differences were found in the output factors for cone sizes smaller than 12.5mm diameter, which were mainly caused by the finite size of the detector and possible response variation. The treatment plan comparison results showed that there were no significant discrepancies between Monte Carlo and the Cyberknife dose calculation for the 5mm cone or 60mm cone in homogeneous patient geometry with target volumes between 1-800 cm³. Large differences (8-11%) were observed for small collimator cones between Monte Carlo and the Cyberknife planning system for lung cases where the equivalent path length correction algorithm overestimated the target dose because of the lack of photon scatter correction at low densities and the lack of electron transport in heterogeneous geometries. The Monte Carlo computation time is 5 min or less for a typical CyberKnife plan with a <100cm³ target volume (consisting of <1000 voxels) on a Pentium4 3.2GHz PC.

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