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Implementation of Monte Carlo Simulations for the Gamma Knife System

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Abstract. Currently the Gamma Knife system is accompanied with a treatment planning system, Leksell GammaPlan (LGP) which is a standard, computer-based treatment planning system for Gamma Knife radiosurgery. In LGP, the dose calculation algorithm does not consider the scatter dose contributions and the inhomogeneity effect due to the skull and air cavities. To improve the dose calculation accuracy, Monte Carlo simulations have been implemented for the Gamma Knife planning system. In this work, the 201 Cobalt-60 sources in the Gamma Knife unit are considered to have the same activity. Each Cobalt-60 source is contained in a cylindrical stainless steel capsule. The particle phase space information is stored in four beam data files, which are collected in the inner sides of the 4 treatment helmets, after the Cobalt beam passes through the stationary and helmet collimators. Patient geometries are rebuilt from patient CT data. Twenty two Patients are included in the Monte Carlo simulation for this study. The dose is calculated using Monte Carlo in both homogenous and inhomogeneous geometries with identical beam parameters. To investigate the attenuation effect of the skull bone the dose in a 16cm diameter spherical QA phantom is measured with and without a 1.5mm Lead-covering and also simulated using Monte Carlo. The dose ratios with and without the 1.5mm Lead-covering are 89.8\% based on measurements and 89.2\% according to Monte Carlo for a 18mm-collimator Helmet. For patient geometries, the Monte Carlo results show that although the relative isodose lines remain almost the same with and without inhomogeneity corrections, the difference in the absolute dose is clinically significant. The average inhomogeneity correction is (3.9\pm0.90)\% for the 22 patients investigated. These results suggest that the inhomogeneity effect should be considered in the dose calculation for Gamma Knife treatment planning.

1. Introduction
Gamma Knife was developed in 1968 by a neurosurgeon, Lars Laksell, and a physicist, Börje Larsson. The Gamma Knife unit is now produced by Elekta Instruments Inc., a Swedish company, based on the
inventions of Lars Leksell. Gamma Knife is widely used in stereotactic radiosurgery (SRS) now as an alternative tool to treat deep-seated intracranial tissues, benign or malignant tumors, which are inaccessible or unsuitable for conventional invasive surgery.

The Gamma Knife system delivers a single high dose of ionizing radiation from 201 Cobalt-60 sources [1, 2]. Each application of radiation, a “shot”, has an ellipsoidal-shaped dose distribution that varies with collimator sizes and iso-center location. Each individual Cobalt beam is mechanically focused through the collimating system onto the target. Each Leksell Gamma Knife includes a radiation unit, several helmets with different collimator sizes, a patient couch and recently an Automatic Positioning System (APS). The collimators are 4, 8, 14, and 18 mm in diameter, individually. Lesions of different sizes and shapes can be treated using different collimators. Individual beams can be blocked to avoid critical structures such as optical nerves. Because the radiation is highly concentrated on the vicinity of the iso-center, usually multiple shots are required to produce a conformal dose distribution to match the irregular tumor volume [3].

There are four generations of the Leksell Gamma Knife units, i.e., model-U, model-B, Model-C and the latest Model, Gamma Knife Model Perfexion. Mode Perfexion was introduced to radiosurgery in June, 2006, which has a new collimator system that allows for increased treatable target volume. The first Gamma Knife Perfexion was installed in France in Mid-July, 2006. Model U has similar source geometry and dimensions as models B and C but it has a larger latitude angles; thus some sources at the superior area of the patient are frequently plugged to protect the optical chiasm. Model B and Model C are widely used Gamma Knife units. They have the same geometries and radiation units except that Model C has a motorized APS. The APS avoids the manual positioning of the target and enables many isocenters in a reasonable treatment schedule resulting in shorter treatment time and more conformal treatment.

Currently the Gamma Knife is accompanied with a treatment planning system, Leksell GammaPlan (LGP), which is a standard, computer-based treatment planning system for radiosurgery using Gamma Knife. In LGP, the patient geometry is created as a 3-dimensional simulation of the skull inside a stereotactic frame based on the measurements of 24 pre-selected points on the surface of the patient skull. The dose calculation is based on a single beam calculation and the contributions from all 201 Cobalt source are summed up for each point of interest. For a single beam, the dose on the beam axis can be calculated using the inverse square law and linear attenuation exponential formula. At other points, the dose calculation also uses the off-axis ratio, which is measured in a single beam profile. The dose delivered by Gamma Knife is calculated in a $31 \times 31 \times 31$ matrix of points. The distance between these points is decided by the grid size.

The LGP dose calculation algorithm ignores the scatter photon contribution and assumes homogeneous patient geometry [1, 4]. Although brain tissues are relatively homogeneous, beams that pass through low density air cavities or high density skull bones are expected to be perturbed. Variations in attenuation and absence of electronic equilibrium adjacent to air-tissue inhomogeneity could cause errors in dose calculation [5, 6].

Monte Carlo simulations have been applied to radiotherapy treatment planning for some time and are capable of accurately predicting various dosimetric parameters as well as doses in regions where electronic equilibrium is lacking. The necessity of more accurate dose calculation using algorithms such as Monte Carlo is now recognized by the medical physics community [7]. Previous Monte Carlo simulations of the GammaPlan output show no differences in homogeneous phantoms [8, 9 and 10]. Recent studies show substantial differences between commonly used treatment planning algorithms and Monte Carlo results when heterogeneous patient anatomy is applied in SRS using megavoltage photon beams [5, 6]. Air-tissue inhomogeneity leads to rapid dose fall-off near the interface due to the lack of electron equilibrium. It is possible that the dose model utilized in the LGP will fail in certain situations. Verification of Gamma Plan is necessary with accurate evaluation techniques such as
Monte Carlo simulations in realistic patient geometries.

In this work, Monte Carlo has been implemented for Gamma Knife treatment planning dose calculation, including the generation of phase-space files for Cobalt source simulation, the conversion of patient geometry from CT data and the dose calculation in real patients as well as in QA phantoms. We also studied the inhomogeneity effect due to the high density skull and low density air cavity on Gamma Knife SRS using realistic patient geometries.

2. Materials and Methods
The Monte Carlo code implemented for this research is MCSIM (MCDOS) [17]. The implementation of Monte Carlo simulations for Gamma Knife took the following three steps. The first step is the Cobalt source simulation, which generates the phase-space files for the simulation. The next is to create QA phantoms and convert patient geometries from CT data. The third step is to put the Cobalt source data and patient geometries in the Monte Carlo program to perform phantom/patient dose calculation.

2.1 Gamma Knife source simulation
For Model B and Model C Gamma Knife units, there are 201 Cobalt-60 sources arranged on a hemispherical surface with a radius of about 400 mm. These are distributed along five radii separated by an angle of 7.5 degrees. Each source is composed of 20 Cobalt-60 pellets that are 1 mm in diameter and 1 mm in length. The final size of a single Cobalt-60 source is 1 mm in diameter and 20 mm in length. Figure 1 illustrates the geometry model of a Gamma Knife beam channel. Each Gamma beam from source is collimated by a stationary collimator and a final helmet collimator. The stationary collimator is assembled with the Gamma Knife unit while the helmet collimator has different sizes, being selected according to treatment needs. Each beam channel in the stationary collimator consists of a 67 mm long cylinder with a radius of 2 mm in a pre-collimator followed by a 92.5 mm cone in a tube-collimator. In the helmet, each beam channel ends in a 60 mm final collimator. When the helmet moves to a treatment position, the entire collimator system forms a cone shaped passage from the Cobalt source to the isocenter. Four different final collimator helmets create 4, 8, 14 and 18 mm nominal beam diameters at the isocenter, respectively. The inner and outer final helmet collimator

![Figure 1. The geometry of a single beam channel in the Gamma Knife unit](image-url)
diameters are 2.5 and 2 mm for the 4 mm helmet, 5.0 and 3.8 mm for the 8 mm helmet, 8.5 and 6.3 mm for the 14 mm helmet, and 10.6 and 8.3 mm for the 18 mm helmet, respectively.

In the Monte Carlo simulation, each Cobalt source is modeled by a cylinder 1 mm in diameter and 20 mm in length. The starting points of Cobalt gamma rays are evenly distributed in the cylinder. The initial angular distribution of the gamma rays is isotropic in each point of the source. To increase the efficiency of simulation, the initial theta of gamma ray is limited to 10 degrees toward the iso-center. This limitation has no effect on the final results [11]. The geometries of the stationary collimator and the helmet collimator are reconstructed in the Monte Carlo simulation exactly according to the Gamma Knife unit. The materials of the collimators are selected as close to the original material as possible. All particles including photons and electrons are traced through the stationary collimator and the helmet collimator until they pass the inner surface of the helmet or are absorbed.

The energy cutoffs in the Gamma Knife source simulation are 700 KeV for electron transport ($ECUT$ and $AE$) and 10 KeV for photon transport ($PCUT$ and $AP$). The electron transport step-length is confined such that the maximum fractional energy loss per electron step is 4% (i.e. $ESTEP=0.04$). The ICRU recommended compositions and stopping power values are used for the materials in the simulation. The phase-space data contains all the particles scored at a plane below the inner surface of the helmet collimator. The number of particles is about 50 million in each of the 4 phase-space files corresponding to each of the 4 helmet collimators of different sizes. In the Monte Carlo simulation, each source is considered to have equal activity although the source activity varies by up to 2% in reality. Each phase-space file represents the one of 201 Gamma Knife sources used for every shot in the treatment.

2.2 Monte Carlo dose calculation
The EGS4 (electron gamma shower version 4, [12]) user code, MCSIM [13], is used in this work for the dose calculation. MCSIM is designed for dose calculation in a 3D rectilinear voxel geometry. Voxel (volume element) sizes are completely variable in all three dimensions. Every voxel can be assigned to a different material. The cross section data for the materials used are available in a pre-processed EGS4 cross-section data file. The mass density of the material in a MCSIM calculation is varied based on the patient’s CT data although the density effect corrections for stopping power of the material remain unchanged. The voxel dimensions and materials are defined in a MCSIM input file together with the transport parameters such as $ECUT$, $PCUT$, $ESTEP$, and the parameters required by PRESTA [14]. Several variance reduction techniques have been implemented in the MCSIM code to improve the calculation efficiency. These include photon interaction forcing, particle splitting, Russian roulette, electron range rejection and region rejection, particle track displacement and rotation, and correlated sampling. Detailed description and application of these techniques has been given elsewhere [15, 16, 17, 18, 19 and 20].

![Figure 2. The collimator helmet viewed from behind and side.](image)
Figure 2 shows the corresponding collimator helmet as viewed from behind and side. In the Monte Carlo dose simulation, all the sources are put in the exact position as they are in the Gamma Knife unit. The exact geometry location can be found in [22]. The sources can be blocked by inserting Tungsten plugs into the helmet. However, in MCSIM, the sources locations are indexed with distances from the source plane to the isocenter and two relative angles of the sources and the phantom: theta and phi. It provides great flexibility to put source planes to the place where is convenient for Monte Carlo dose calculation, especially with the use of phase-space beam data. In this study, the source planes are just beneath the helmet, the distance from source planes to isocenter is 16.5 cm. The angles phi and theta can be derived from the locations provided in reference [22] with a formula:

\[
\tan(\phi) = \frac{y}{z} \\
\cos(\theta) = \frac{x}{\sqrt{x^2 + y^2 + z^2}}
\]

where \(x, y\) and \(z\) is the index for source locations in Gamma Knife unit and isocenter is at the origin as (0, 0, 0).

For patient dose calculation, the simulation phantom is built from the patient’s CT data with up to 516×516×516 voxels. The size of a voxel varied from 0.075 cm to 0.4 cm for this study. A separate inhouse program is developed to convert the patient’s CT data to required dimensions, material types and densities. The organ contours are also obtained for dose calculation and analysis. Cobalt-60 source phase-space from the beam simulation is used as a source input with 201 source positions and beam incident angles.

MCSIM produces dose data files that contained geometry specifications such as the number of voxels and their boundaries in all the three directions as well as the dose values and the associated statistic uncertainties in the individual voxels and organs (structures). The EGS4 transport parameters are \(ECUT=AE=700\text{keV}, PCUT=AP=10\text{keV}\) and \(ESTEPE=0.04\) for the patient dose calculation. The statistical uncertainty in the dose is generally 2% or smaller of the maximum dose value.

2.3 Monte Carlo geometry reconstruction

Two kinds of geometries are investigated in this study. One is a spherical QA phantom for the experimental measurement and the other is the patient geometry generated from CT data.

The QA phantom is provided with the Gamma Knife unit. It consists of tissue-equivalent plastic materials. It has a spherical shape and can be mounted on the Gamma knife unit for output measurement. The center slice is designed with a tunnel, which allows for the insertion of a pin-point chamber. The center of the chamber is located at the isocenter of 201 Cobalt beams with coordinates \((XA, YA, ZA=100, 100, 100)\) in LGP system. At such coordinates, it is possible to calculate the dose rate of the 201 isocentric Cobalt 60 sources according to the ICRU 24 report to verify the linearity of the electrometer response connected to the chamber.

The QA phantom in the Monte Carlo simulation is reconstructed as a 16 cm diameter sphere with a unity density with air surrounding it. The center of the phantom is located at the isocenter of the Gamma Knife unit. Due to limitations of the computer memory and speed, the voxel dimension is set to 0.75 mm in each direction. The QA phantom is simulated with a 220×220×220 matrix.

The patient geometry in the Monte Carlo simulation is reconstructed from patient CT data. The patient selected for this study was treated at our institution. The CT data were taken from a GE Lightspeed CT scanner with a 70 cm bore. An inhouse program has been developed to read DICOM format CT data and generate EGS4 format geometry files. The CT number has been converted to tissue density using a bilinear formula [21]. The CT resolution has been kept unchanged in \(x, y\) and \(z\) directions. This study is focused on the treatment target in the head, and the major media in the head are soft tissues and skull bones. The skull is assigned as bone material based on density and location. The average bone density of the skull is around 1.5 g/cc although it varies between patients. To reduce
the memory requirement and to save CPU time, the patient geometry is simulated using a 256×256 matrix in the x-y plane with the original CT resolution by removing those voxels that contain air surrounding the patient.

3. Results
Monte Carlo simulations have been performed on the sphere phantom as well as the patient geometry. The comparison of the beam output between measurement and Monte Carlo simulations for the QA phantom provides an excellent index to verify the implementation of Monte Carlo for the Gamma Knife. Monte Carlo also has been applied on real patients to investigate the inhomogeneity effect from high density skull bones and low density air cavities.

3.1 Comparison of Measurement and Monte Carlo results for the QA phantom
To verify the implementation of Monte Carlo simulations for the Gamma Knife, the QA phantom is investigated by comparisons of Monte Carlo results with measurements with and without a lead cover.

![Figure 3. The QA phantom and lead-covered QA phantom](image)

Figure 3 shows the QA phantom with and without a lead covering. The lead-cover is home-made by putting a 1.5 mm lead sheet around the sphere QA phantom. Lead is chosen due to its softness, malleability and ductility. Several rectangular lead sheets are cut into small pieces and pressed to cover the round surface of the QA phantom tightly. Those small lead pieces are taped together around the phantom carefully that the overlaps or gaps between those pieces are minimized. The average overlap or gap is less than 1mm. Eventually the QA phantom is completely surrounded by the lead covering except the hole for the ion chamber, which is not in the entry paths of any cobalt beams.

The photon mass attenuation of lead and compact bone is $6.21 \times 10^{-3} \text{ m}^2/\text{kg}$ and $6.13 \times 10^{-3} \text{ m}^2/\text{kg}$ for Cobalt-60 gamma rays, respectively. The lead sheet of 1.5mm thickness provides similar attenuation as that of compact bone (skull) of 10mm thickness at a density of 1.8 g/cc. Several measurements were performed in the QA phantom with and without the lead cover.

The output reading was first taken in the lead-covered QA phantom. The average output reading is 0.9474 nC in 1 minute. After the lead was removed, the average reading is 1.055 nC in 1 minute. Since the half life for Cobalt decay is more than 5 years and the time interval between 2 measurements is within 10 minutes, the output difference caused by different measurement time can be ignored. The output difference in these measurements comes from the extra 1.5 mm lead attenuation. The ratio of the measurement is $(0.898\pm0.007)\%$.

The error of the reading came from the imperfect connection of lead pieces. The overlap or gap between two neighboring lead pieces will cause more attenuation or less attenuation for the beams, which go through it than the 1.5 mm lead. The corresponding result is that the reading of the output
would be lower or higher than the real output reading. For estimation of error from this imperfection of connection, the overlap or gap area has to be measured. There are total 28 pieces of 1.5 mm lead sheet, which are used to cover the QA phantom. The total perimeter of all these lead sheets is measured as 587.5 cm. For conservative estimation, the width of the connection is set to be 2 mm. The total connection area is the product of the half perimeter and the width, which is $587.5 \times 2 = 1175$ mm$^2$. At the same time, the total area of the QA phantom surface is 80425 mm$^2$ and the ratio of the connection area and the total surface is 7.3%. There are 201 Cobalt-60 sources and the connection area distributed randomly over the phantom surface, it is expected that the probability for beams to fall into the connection area is proportional to the area ratio. Thus the error of output caused by the beam goes through the connection area is the product of the ratio output difference which is about 0.73%.

Figure 4 illustrates the relative isodose lines in the QA phantom from Monte Carlo simulations. The material of the QA phantom in the Monte Carlo simulation is water with unity density. The cover in the figure is made of lead with density of 11.350 g/cc. The statistic error at the isocenter is about 1%. There are two sets of relative isodose lines in the figure, which are the isodose lines in the QA phantom with and without the lead-cover for the exact setup and beam arrangement. Even there is dose difference between two doses in figure 4, the relative isodose lines are very similar. They could not be distinguished from each other if they were normalized individually. The lead-cover has little effects on the isodose shape (relative doses).

The absolute dose output has been reduced as expected. A small volume (0.15 cc) which is about the same with pin-point chamber has been created around the isocenter. The volume is about the size of the pin-point chamber used in the measurement. The output is set to the average dose in the volume. The setups of Monte Carlo simulations are identical for the QA phantom with and without the lead cover. The ratio of the simulated outputs with and without the lead cover is 0.893 ± 0.010. The ratio matches the measurement result very well. It shows that the implementation of Monte Carlo simulations for the Gamma Knife is accurate.
3.2 Preliminary results of inhomogeneity effects on patient dose calculation
Monte Carlo simulations have been performed on patient geometries converted from CT data. Twenty-two patients have been selected for this study. The patient CT data was transferred from the treatment planning system to a Linux machine. An inhouse conversion program converts the CT data into EGS4 simulation geometry files. The isocenter of each simulation is chosen roughly at the geometry center. Isocenters close to the high density skull and low density air cavities are also created for relative isodose and output comparison. The 18mm helmet is chosen for this study since it is most frequently used in output check of Gamma Knife unit. The statistical error of Monte Carlo results is 1% at the isocenter for each simulation. For every patient geometry, simulations are performed in both heterogeneous geometry and homogeneous geometry (assuming unity density).

![Dose comparison in CT Geometry](image)

(a) ![Inhomogeneity effect vs. skull bone density](image)

(b) Figure 5. Inhomogeneity effect vs. skull bone density

Figure 5a shows the doses in homogeneous and heterogeneous geometries. The relative isodose lines are almost the same for both geometries. For the simulations in which the isocenters are close to the skull or an air cavity, the isodose lines are also similar for homogeneous and heterogeneous geometries. This means that the inhomogeneity effect on the isodose distribution is very limited. The DVH for a small volume located at the isocenter shows the absolute dose difference. Figure 5b shows the mean dose difference between homogeneous phantom and heterogeneous geometries for 22 patients is 3.9%. The dose difference between two phantoms comes from D10 in DVH comparison. The absolute dose difference results from the attenuation of the high-density skull and increases with the skull density.

4. Conclusion
Monte Carlo simulations have been applied to dose comparison in CT-based geometries with and without inhomogeneity corrections. Monte Carlo simulations agreed very well with measurements for the spherical QA phantom, demonstrating the accuracy of the Monte Carlo implementation for the Gamma Knife system. The results show that the inhomogeneity effect on relative isodose distributions is very limited with a single shot. However, inhomogeneity has a significant effect on the absolute dose. For the patients investigated, the average dose received by the patient is 3.9% less than the planned dose. The inhomogeneity correction should be considered in gamma knife planning.

The inhomogeneity correction can be applied with a global correction factor for a patient based on the current Gamma Knife planning system. In the future, more accurate planning systems with
inhomogeneity corrections should be applied to Gamma Knife treatment planning dose calculation for individual patients.

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