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MONTE CARLO SIMULATION FOR LEKSELL GAMMA KNIFE® RADIOSURGERY PLAN VERIFICATION

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ABSTRACT

This paper reviews the current status of the Indiana University Department of Radiation Oncology project on Monte Carlo based treatment planning verification for the Leksell Gamma Knife®.

Key Words: Leksell Gamma Knife, radiosurgery, Monte Carlo simulation, treatment plan verification system

1. INTRODUCTION

The Gamma Knife® (see Figure 1) was developed in 1968 by Professor Lars Leksell of the Karolinska Institute in Stockholm, Sweden and Professor Borge Larsson of the Gustaf Werner Institute at the University of Uppsala, Sweden. Gamma Knife radiosurgery spares healthy areas of the brain from high-dose exposure to radiation, and eliminates many of the risks inherent in traditional invasive surgery.

The Gamma Knife® focuses photon radiation from 201 60Co sources distributed on the surface of the partial sphere on the target area. The beams exit the primary and final collimator system, creating beam diameters of approximately 4, 8, 14 and 18 mm at the isocenter of the beams (focus). Individual source fluences can be blocked to create optimal dose distributions. Converging photon beams are directed toward the lesion with accuracy better than 0.3 mm [1]-[4]. The Gamma Knife® unit is accompanied by a treatment planning system, GammaPlan®, a computer based dose planning system specifically designed for simulation and planning of stereotactic Leksell Gamma Knife® radiosurgery based on tomographic and projection images [5].

The patient's head is placed inside the helmet (partial sphere) and held at four points to the skull. Each of the "drain holes" is an aperture that creates a beam of varying diameters. When the individual beams converge, that area receives the full treatment dose of gamma radiation.

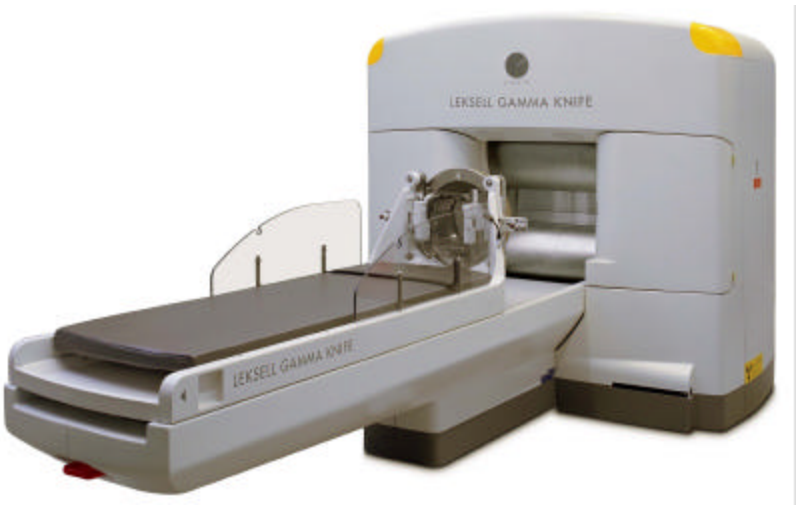

Figure 1. Gamma Knife unit installed in IU Department of Radiation Oncology

Stereotactic dose calculation algorithms, implemented in GammaPlan® are semi-empirical and are based on a homogeneous media approach. Algorithms utilize an isocentric technique with a known beam profile and a constant linear attenuation through tissue. Fields are generally considered to be small enough that a "primary only" formalism can be used. Scatter is included only in the measured off-axis ratios. The GammaPlan® system also makes the assumption that the target material is composed entirely of unit density material ([1], [2] and [5]). While brain tissue is relatively homogeneous, beams that pass through the low density air cavities of sinuses or high density of bone are susceptible to potential perturbation. Due to the loss of side scatter equilibrium in small photon beams, small low density tissue heterogeneities can cause marked changes in the central axis dose [6]. These calculations show substantial differences between commonly used treatment planning algorithms and Monte Carlo results depending on whether or not homogeneity is assumed in stereotactic radiosurgery. Tissue heterogeneities produce a lateral broadening of the beam, resulting in a smaller volume contained within the higher isodose levels 80-90%) with a corresponding increase in volume treated at the lower isodose levels (<50%). This suggests that the GammaPlan® will be inaccurate in certain situations. Verification of dose calculations by GammaPlan® is necessary with more accurate evaluation techniques such as Monte Carlo simulation.

Quality assurance programs require that the calculation output of computerized planning systems be verified by measurements [7]. Small photon beams present numerous difficulties in physical dose measurement. As an alternative to benchmarking, investigators have shown that the Monte Carlo method is capable of accurately predicting standard dosimetric parameters as well as dose in regions where electronic equilibrium is lacking. Small beams are perturbed by tissue heterogeneities in a manner that conventional methods are unable to predict. Utility of a Monte Carlo approach to radiotherapy treatment planning is now clearly recognized [8].

The purpose of this paper is to give a review of the current status of Monte Carlo based treatment planning verification system for Leksell Gamma Knife®. PENELOPE general-purpose code [9] was selected as the core engine for this system.

2. SOURCE MODELING AND HOMOGENEOUS PHANTOM DOSIMETRY

Monte Carlo data used to verifying Gamma Plan® output showed no differences with data obtained using a homogeneous phantom. Published data (see review in [10]) show differences up to 25% between Leksell Gamma Plan® dose calculations and Monte Carlo results in predicted doses near tissue interfaces and in the area of dose edges. These differences were not addressed in the cited manuscripts.

In the first stage of this project [10], radiation from a single 60Co assembly (See Figure 2.a) traversing the collimator system was simulated, and phase space distributions at the output surface of the helmet for photons and electrons were calculated. Figure 2.b illustrates comparison of the calculation results with film measurements on the output surface of the helmet. No significant discrepancies were observed by comparing Monte Carlo data and measurements.

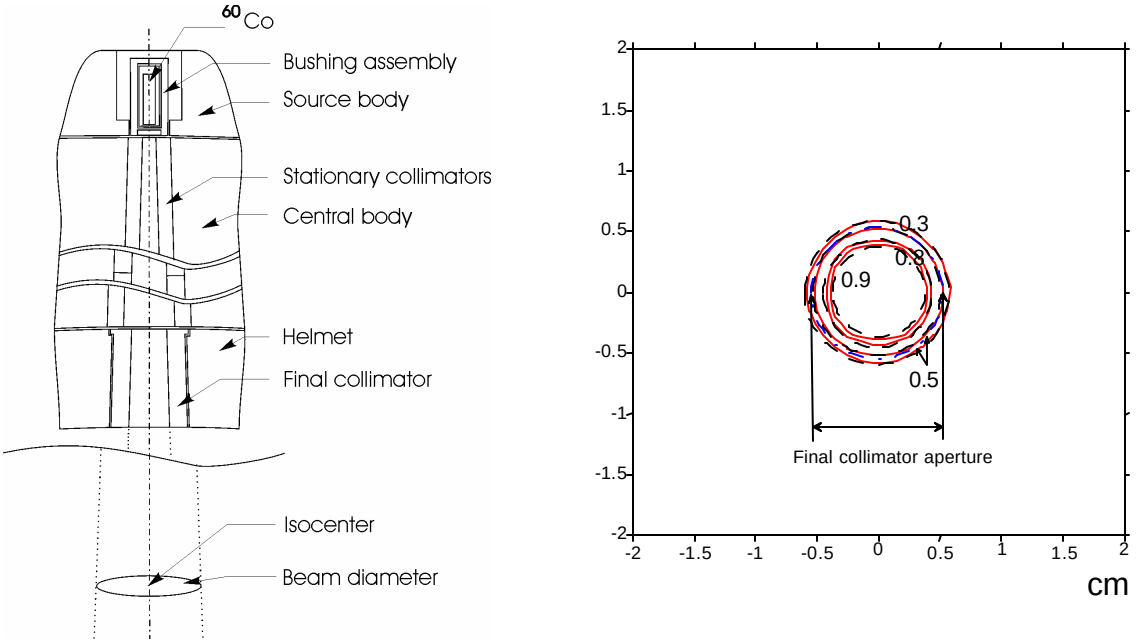

Figure 2: a) Draft of the geometry used for simulation of the radiation from single source, b) Dose on the output surface of the helmet: solid lines are Monte Carlo simulation; dashed ones are the results of the measurements.

The characteristics of the final emitted beam were used to build a two-stage Monte Carlo simulation of irradiation of a target. Monte Carlo model has the following features:

- PENELOPE-based Monte Carlo engine;
- Semi-empirical description for photon angular distribution based on the results of a single source simulation;
- Secondary electrons from the helmet are not simulated in most situations;
- Photon leakage from plugged sources and through the final collimator is taken into account in the simulation.

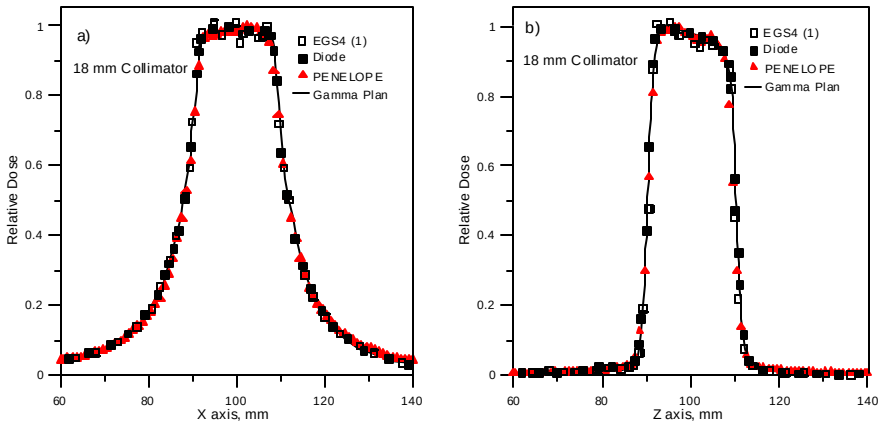

Figure 3 (a-b): Comparison of relative dose in 160 mm spherical polystyrene phantom from 201 collimated sources calculated with Monte Carlo method and measured. Figure presents dose data for 18 mm final beam diameter versus x-axis (a) and z-axis (b). Filled triangles are the results of present Monte Carlo study with the PENELOPE code; open squares and crosses are the EGS4 data (labeled as EGS4 (1) taken from [11], and EGS4 (2) taken from [12], respectively. Solid lines represent the dose profiles from Gamma Plan treatment planning system. Filled quadrics and open circles represent the diode measurements [13] and film measurements [12], respectively.

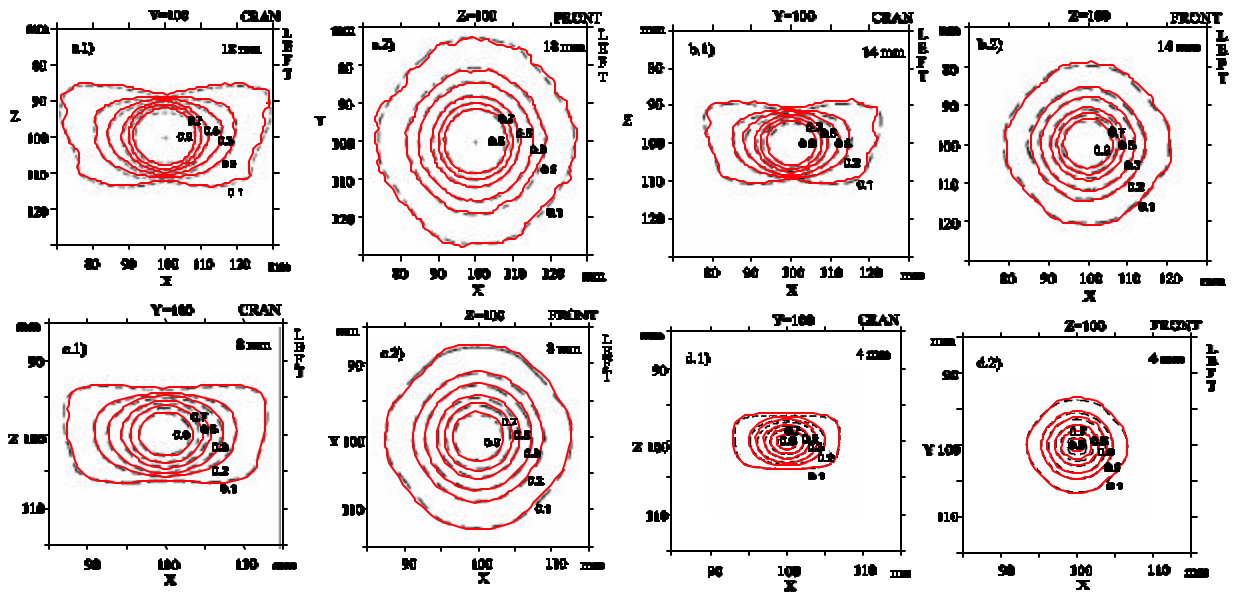

Figure 4: Comparison of Monte Carlo simulation (solid lines) with Gamma Plan® output (dashed lines) for spherical homogeneous phantom for different final beam apertures.

Following the standard Leksell Gamma Knife® dosimetry protocol, a dose field inside a standard spherical polystyrene phantom has been computed and compared with experimental results, with calculations performed by other authors with the EGS4 Monte Carlo code, and data provided by the treatment planning system Gamma Plan®. Good agreement was found between these data and results of simulation in homogeneous media (see Figure 3 a,b). Due to this established accuracy, it was concluded that PENELOPE is suitable for simulating problems relevant to

stereotactic radiosurgery. The Monte Carlo code PENELOPE was used to simulate photon flux from the Leksell Gamma Knife®. Throughout comparison was made with the Gamma Plan® output. There were no differences observed between results of Monte Carlo simulation for homogeneous spherical calibration phantom LUCY and Gamma Plan® dose distribution calculations (See Figure 4). It was concluded that both system yield identical results for homogeneous media.

3. INHOMOGENEOUS PHANTOM DESIGN AND SIMULATION

Benchmarking of the system requires a special protocol for inhomogeneous dosimetry for Leksell Gamma Knife®. The inhomogeneous phantom CYCLOPS was designed on the basis of LUCY (Sandstrom Trade & Technology) and included air cavities to simulate sinuses and an aluminum slab to simulate bone. Figure 5-a presents a sketch of the phantom design. The phantom itself is shown in Figure 5-b.

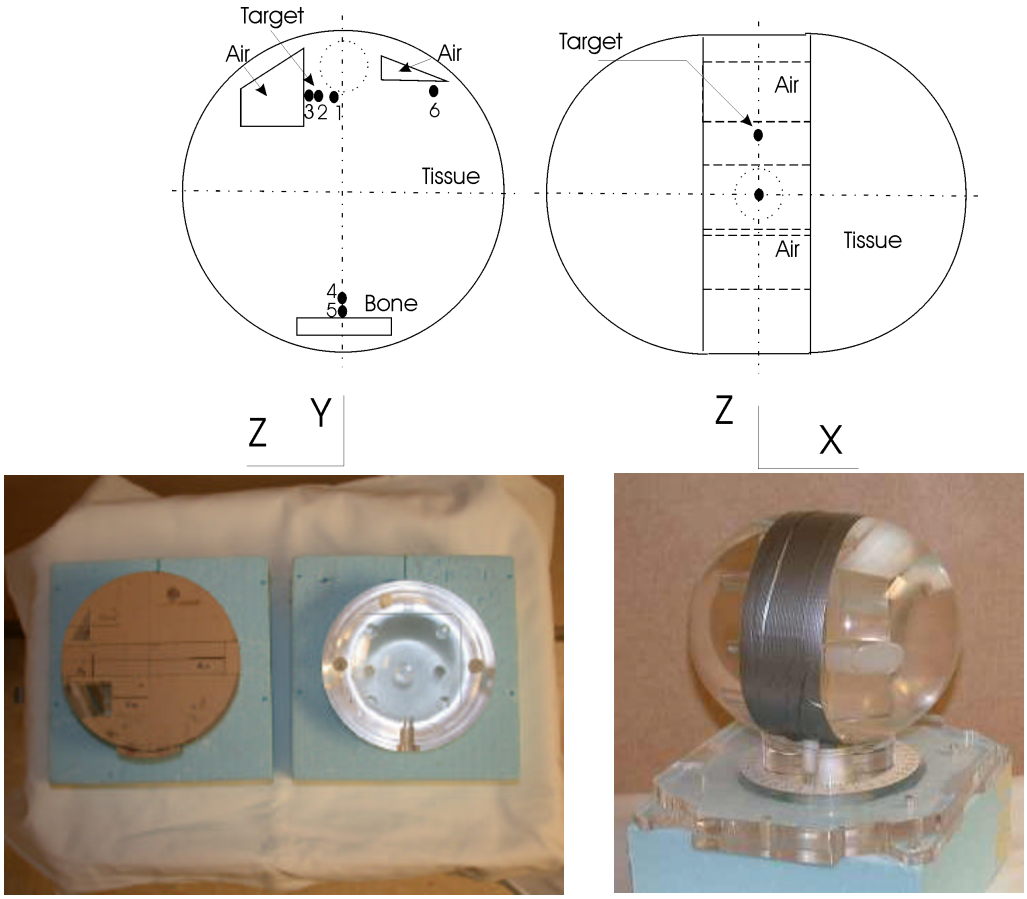

Figure 5: Inhomogeneous phantom “CYCLOPS”.

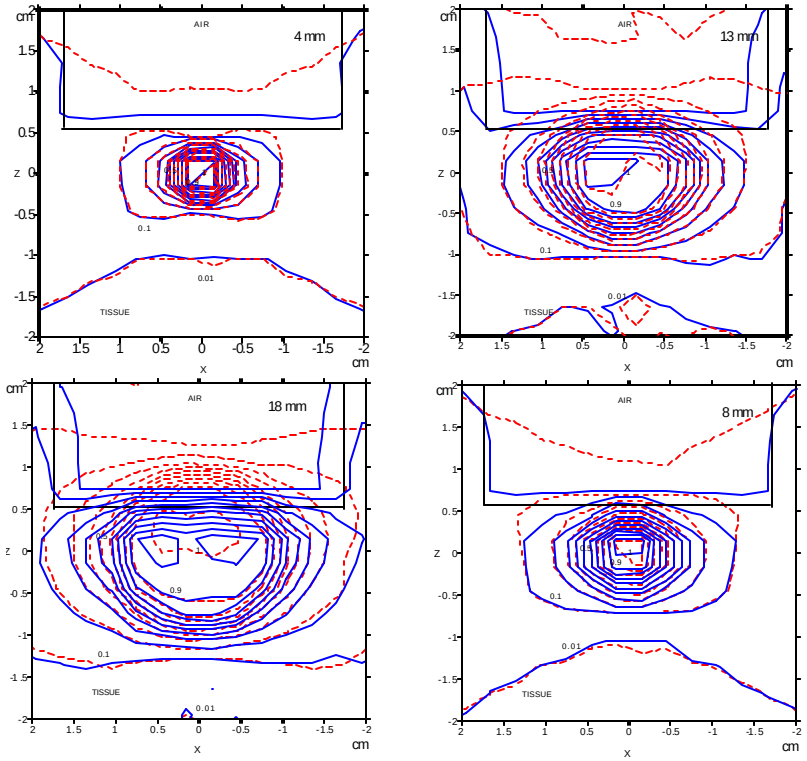

Figure 6: Dose from 201 sources for different final beam sizes close to air-tissue inhomogeneity. Comparison of calculations for inhomogeneous phantom (solid lines) and homogeneous media (dashed lines)

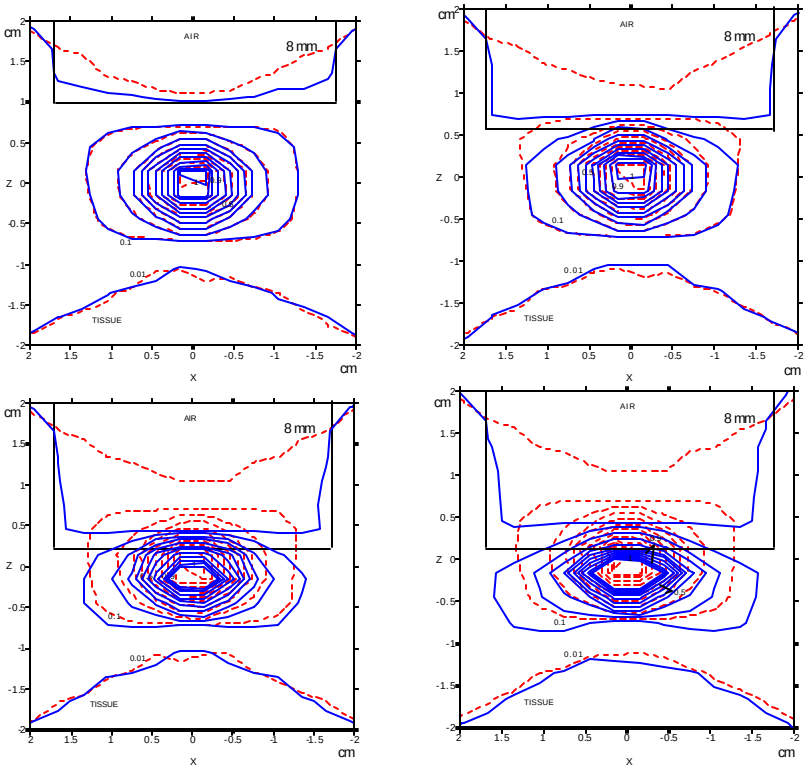

Figure 7: Dose in homogeneous (dashed lines) and inhomogeneous (solid lines) media for 8 mm final beam size for positions of the isocenters at different distances from the air-tissue interface. The distances to the air cavities are 10 mm (a), 5 mm (b), 2 mm (c) and 1 mm (d).

Results of previous work [10] on Gamma Knife® source modeling were used to setup initial angular and intensity distributions of photon beams. The cutoff energy of secondary electron transport and the threshold of secondary electron generation were selected in such way that residual electron range will be considerably less than the characteristic size of the spatial scoring bin (voxel). A voxel of 0.3 cm was selected for preliminary calculations. All 201 sources were assumed to be unplugged in simulation. Single shots were considered. Targeting points were placed at different distances from the interfaces between plastic (tissue) and air, plastic (tissue) – bone (aluminum) (see Figure 5-a). Results of simulation are presented in Figures 6-7. Calculations show that the dose perturbation effect at the air-tissue interface can have a significant impact on the dose distribution.

Air-tissue interface: Figure 6 shows perturbations in the dose field for different final beam sizes when the targeting point is placed 5 mm from the air-tissue interface. Solid lines represent isodose curves in inhomogeneous media and dashed lines show results of simulation in a homogeneous phantom. The dose filed is normalized to the value of the dose at isocenter.
It is seen from the figure that dose perturbation may be significant when the interface crosses the area covered by beam. Dose perturbation increases with proximity to the interface, according to results presented at the Figure 7. The distance at which perturbation is observed depends on the isodose level considered and the distance between the targeting point and the interface. The characteristic size varies from 2.5 mm to 10 mm. Figure 7 shows that dose distribution both of 90% and 50% isodoses in homogeneous media are perturbed by inhomogeneity effect.

Bone-tissue interface: Bone interface does not show significant dose perturbation (see Figure 8).

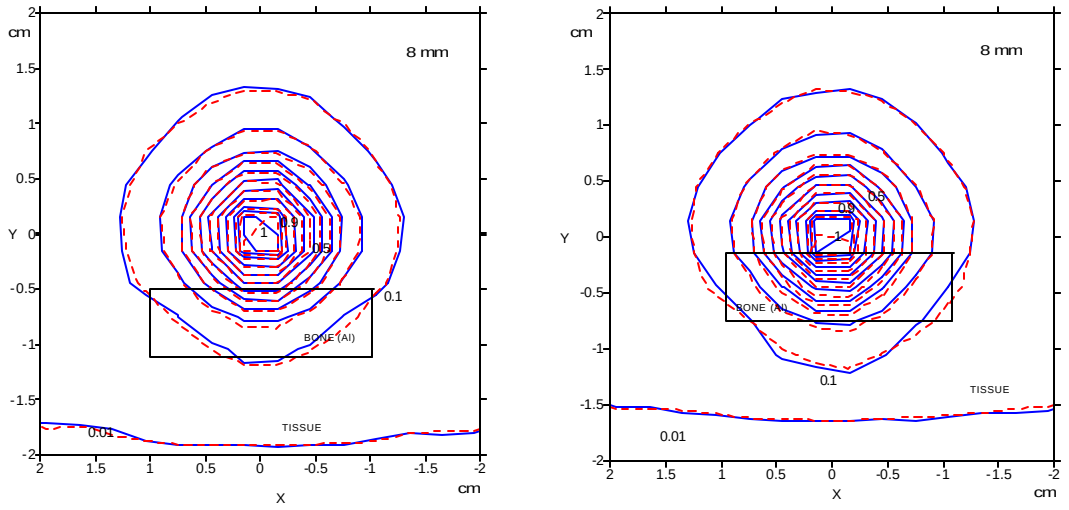

Figure 8: Dose in homogeneous (dashed lines) and inhomogeneous (solid lines) media for bone-tissue interface.

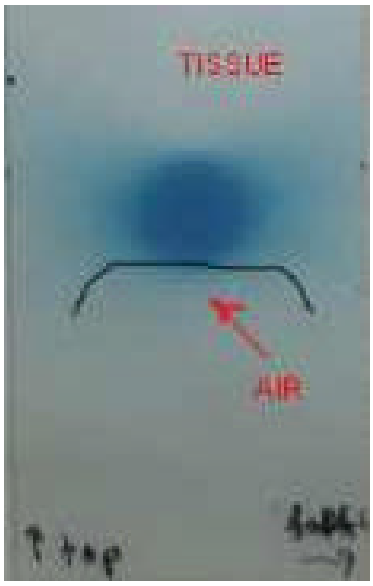

Figure 9: One of the preliminary measurements of dose field in the presence of air –tissue inhomogeneity.

Measurements to verify the simulation results were done with the dosimery film Gafchromic®. Preliminary results show deformation of the dose field in the presence of air-tissue inhomogeneities (see Figure 9 in comparison with Figure 7). Results of this investigation and proposed corrections based on inhomogeneity will be published separately.

Results of the simulation for the inhomogeneous phantom show that significant dose changes may be observed near inhomogeneities compared with homogeneous media. Used for treatment planning for the case of Leksell Gamma Knife®, Gamma Plan® does not yield correct results in the presence of inhomogeneities due to limitations of its physics model. This justifies our efforts to development a Monte Carlo based treatment planning verification system.

4. DESIGN OF TREATMENT PLAN VERIFICATION SYSTEM

The main steps in implementing a Monte Carlo based treatment planning verification system are

- Computer Tomography (CT) based geometry and detailed description of the irradiated target,
- Voxelization of PENELOPE to build a primary voxel-based Monte Carlo tool and algorithm modification to reduce run-time,
- PC based prototype of the treatment verification system and PENELOPE-based client server application adjusted for parallel computing,
- Connection with GammaPlan treatment planning system input and output, and
- Design alternative to PENELOPE code specialized for radiosurgery with cobalt unit.

The flow chart diagram of the system under development is presented in Figure 10.

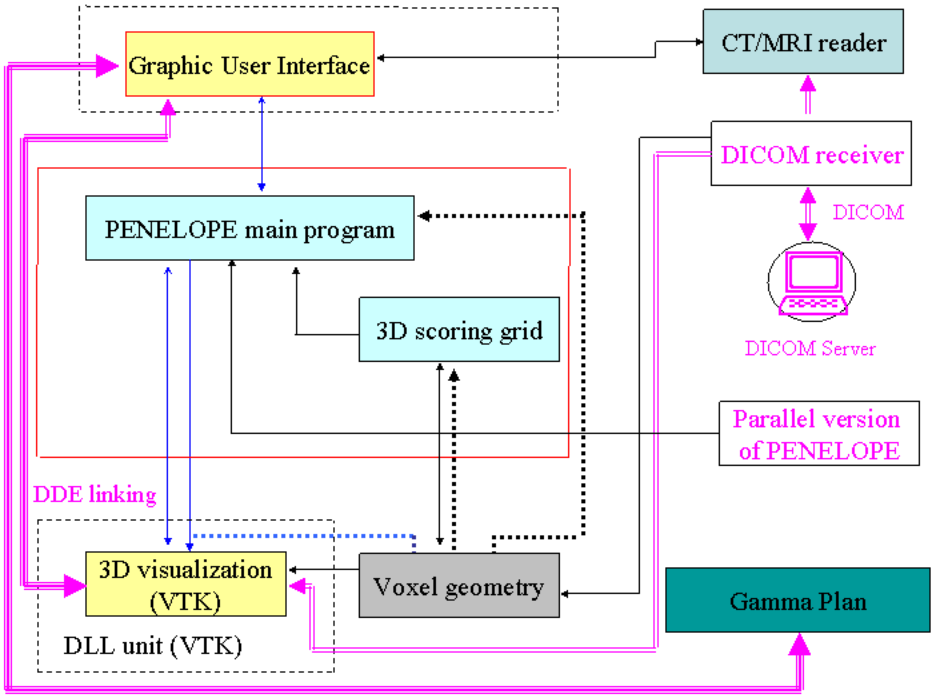

Figure 10: Flow chart diagram of the treatment plan verification system

Coding of the unit MCscan has been completed for transferring images from the CT scanner into for Monte Carlo simulation. Current efforts are ongoing to evaluating uncertainty in dose calculations caused by noise introduced by the CT scanner in the image and influence of magnification of the original image to reduce the number of voxels in the simulation. Image changes due to magnification are illustrated in Figure 11. It is shown that resolution of geometry representation decreases with magnification.

Though PENELOPE code is adequate for simulating dose deposition in homogeneous media and simple geometries, radiotherapy applications involve numerous variations of material and density over small distances. Patient geometry is usually simulated by a map of densities over a large number of relatively small (from 0.1 to 1 mm) parallelepipeds (voxels) (512x512x200 (typical amount of slices)=52,428,800 voxels). Currently, Monte Carlo simulation of absorbed dose for a such large scale problem is feasible only when employing computer resources on a scale not commonly available in medical centers. Therefore two different implementation of the treatment plan verification system are in a progress. One is IBM PC-based, and, the second is parallel computing-based. Parallelization of PENELOPE code will be presented in a separate paper [14]

At the present time work is focused on development of two versions of the Monte Carlo code, working with geometry acquired from CT images. The PC version of basic voxel–based algorithm was already implemented and under testing. The basic voxelization is an algorithm where each voxel contains different material. This algorithm has low efficiency and will be used for benchmarking of the modified scheme and for the parallel computer version of the treatment planning verification system. Modification of the algorithm to trace particles in combined voxels with similar radiation physical properties is under consideration. Usage of this approach may

significantly reduce run-time. Next, parallel computing will be implemented. Results of simulation with the voxelized geometry versus PENGEOM2 implementation of quadric geometry are promising. Details of the implementation of the Monte Carlo-based treatment verification system will be presented later pending benchmarking the current versions.

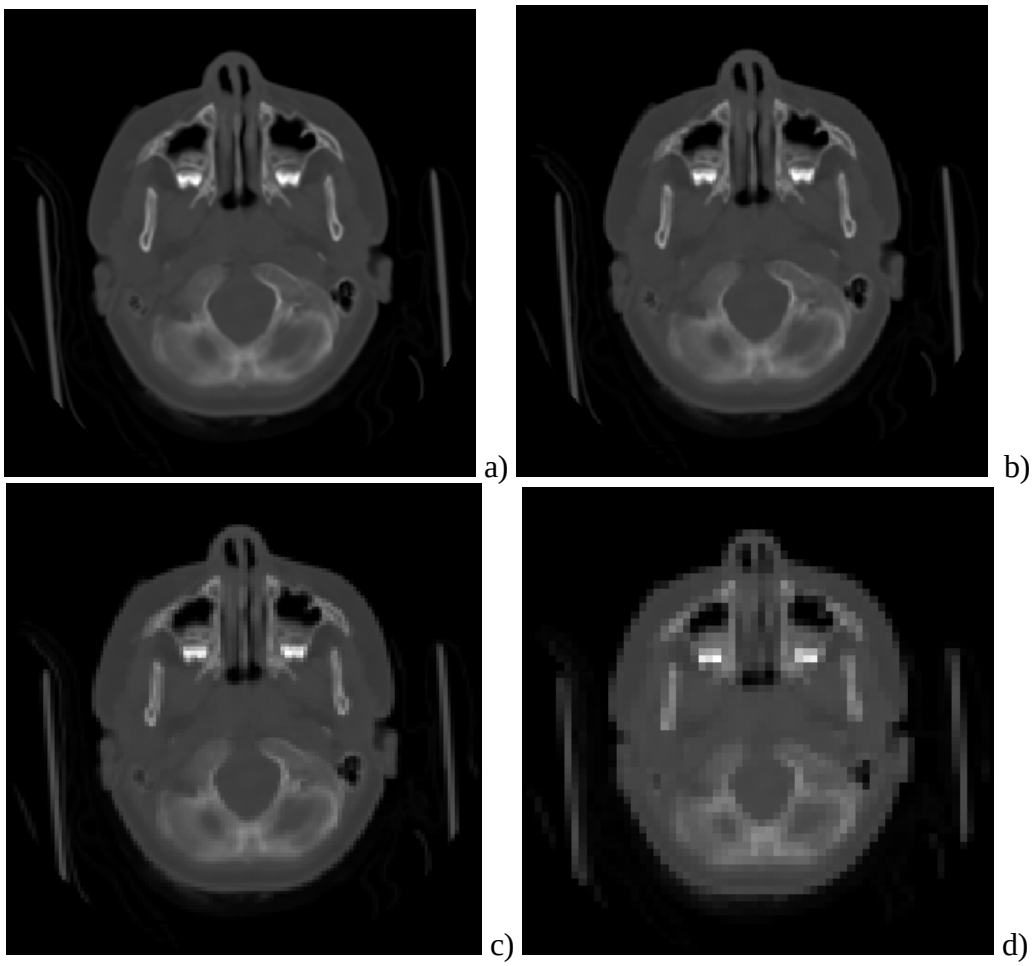

Figure 10: Magnification of CT image with MCscan; a) original image (512x512 voxels of 0.418 mm size); b) magnification x4 (256x256 voxels of 0.836 mm size); c) magnification x6 (128x128 voxels of 1.612 mm size) and b) magnification x8 (64x64 voxels of 3.344 mm size).

5. CONCLUSIONS

When treating neurologic disorders of the brain with ionizing radiation, it is prudent to accurately characterize the radiation delivery and its interaction with normal tissue. While the mechanical design of Gamma Knife® provides high accuracy its computer planning software relies on older technology and algorithms. These algorithms have been shown to have shortcomings in accurately simulating the treatment, especially when treating near the surface, the skull or in the region of the eyes. Monte Carlo techniques have been long recognized as most rigorous and accurate for a wide variety of geometries. Historically these techniques have not been widely implemented due to limitations of computer capability and speed. With modern computer hardware and faster algorithms, these techniques are now feasible. The results

presented in this paper will help to identify and correct the shortcomings of current treatment planning, eventually leading to improve quality of treatment.

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