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Clinical Implementation and Commissioning of Monte Carlo Dose Calculation Algorithm for CyberKnife Treatment Planning

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Abstract: ***The purpose of this study is to commission and evaluate the Monte Carlo (MC) dose calculation algorithm for the CyberKnife S7 Robot system (Accuray, Inc.), which utilizes a 6 - MV X - band linear accelerator on a robotic arm with a Multi Leaf Collimator (MLC). We implemented a MC dose calculation algorithm for SRS/SRT treatment planning with the CyberKnife system. This algorithm simulates energy transport and dose deposition in tissue using three independent probability distributions for initial photon characteristics, based on dosimetric data from the Accuray Precision® system. The model was optimized by adjusting maximum energy and source full width at half maximum (FWHM) to minimize discrepancies with measured tissue phantom ratios. MLC plans were calculated and irradiated on various phantoms. Dose and profile measurements were taken with a PTW farmer - type ionization chamber and PTW microdiamond chamber and EBT3 Gafchromic films, followed by comparisons to TPS doses to determine dose differences. Patient - specific quality assurance (QA) was conducted using global gamma index criteria of 2%/2 mm. The optimal parameters for maximum energy and source FWHM were found to be 6.4 MeV and 1.8 mm, respectively. Results of calculations made using these parameters agreed approximately with those actually measured to within ±1% only. Monte Carlo simulation model of MLC covering of CyberKnife is clinically acceptable but tends to underestimate the delivered dose by an average of −1.3%. After 11 cm depths the dosage difference added on further. It is pragmatically advised to rely on the Monte Carlo model with the MLC primarily in heterogeneous regions [(7)] such as lungs***

Keywords: Commissioning, CyberKnife, Monte Carlo algorithm (MC), EBT3 Gafchromic films, Multi Leaf Collimator (MLC)

1. Introduction

The CyberKnife (CK) system is a sophisticated radiation therapy system designed to deliver stereotactic radiosurgery (SRS) and stereotactic body radiation therapy (SBRT) to tumors using a robotic arm and real - time image guidance. It offers three types of collimators, including fixed collimators with diameters ranging from 5.0 to 60 mm, Iris collimators with variable circular apertures from 7.5 to 60 mm, and the multileaf collimator (InCise™ Multileaf Collimator, 2.5 mm Leaf width), which accommodates larger, irregularly shaped lesions with a maximum treatment area of 115.0 × 100.1 mm². [(1, 5)]

The CK's Precision treatment planning system (TPS) offers three dose calculation algorithms: RayTracing, Finite Size Pencil Beam (FSPB), and Monte Carlo (MC). While the RayTracing algorithms are useful for certain treatment scenarios, they can be less accurate in heterogeneous tissues, such as the lungs. [(4)] This is due to their reliance on effective path length calculations, which may not adequately account for the complex interactions of radiation with varying tissue densities.

The Finite Size Pencil Beam (FSPB) algorithm is exclusive to the MLC. It divides the beam into small rectangular pencil beams, calculating the dose distribution through the convolution of energy fluence and dose deposition kernels. Similar to RayTracing, the FSPB algorithm is also inadequate for heterogeneous media due to its reliance on effective path length for dose calculations.

In contrast, the Monte Carlo (MC) algorithm, initially available for fixed and Iris collimators, has now been adapted for the MLC. This algorithm effectively accounts for lateral electronic scatter and electronic disequilibrium, enhancing dose calculation accuracy, particularly in heterogeneous environments.

The Monte Carlo method is recognized as the most precise dose calculation technique for radiation therapy treatment planning and dosimetry verification, demonstrating significant differences compared to conventional algorithms. Its ability to accurately model complex beam delivery configurations and heterogeneous patient geometries, combined with advancements in computing power and algorithms, has made MC calculations feasible within clinically acceptable timeframes.

This study aims to commission and validate the MC calculation algorithm for the MLC, with a particular focus on lung treatments. To facilitate routine clinical dose calculations using Monte Carlo simulations, we developed efficient algorithms for tracking particle transport and scoring energy deposition in heterogeneous geometries. Additionally, we established practical photon source models and beam commissioning procedures to support the widespread application of Monte Carlo dose calculations in CyberKnife SRS/SRT treatment planning. This paper details the source model, beam commissioning procedures, and patient dose calculation algorithms, comparing phantom dose distributions and patient treatment plans between the new Monte Carlo algorithm, the CyberKnife TPS, and experimental measurements.

2. Materials and Methods

In this study, we used the Precision Treatment Planning System (TPS) version 2.0.1.1 to develop beam models for various algorithms, including RayTracing (with both fixed

and Iris collimators), Finite Size Pencil Beam (FSPB) (MLC), and Monte Carlo (MC) (MLC).

For MC model creation, we measured tissue phantom ratios (TPRs), dose profiles, and output factors across 11 square field sizes for the MLC, ranging from 7.6 × 7.7 mm² to 115.0 × 100.1 mm². [3, 1] Data acquisition was conducted using a PTW MicroDiamond chamber placed in a PTW MP3 water tank. Additionally, commissioning measurements, including TPRs, dose profiles in both the X and Y directions, and output factors, were collected to implement the FSPB algorithm with the MLC. Input data for these models were consistent with those acquired during the CyberKnife system commissioning process. [3, 5]

During initial measurements, Accuray recommended using either the stereotactic MicroDiamond or a diode detector, despite limitations in accurately measuring dose in the tails of profiles due to their non - water equivalence. The Monte Carlo (MC) model was iteratively refined by minimizing the differences between measured and MC - calculated tissue phantom ratios (TPRs) and dose profiles. Two primary parameters were optimized: the maximum energy (Emax), representing the peak value of the energy spectrum, and the source size (S), defined by its full width at half maximum (FWHM).

TPRs were used to optimize energy Spectrum Emax, while profiles were used to adjust source size (S). Although S has a minor effect on TPRs, Emax significantly influences both profiles and TPRs. As a result, TPRs for all field sizes and depths must be recalculated with the selected Emax to derive accurate profiles.

Energy spectra, which characterize the distribution of initial photon energies and are defined by Emax, are pre - calculated in the TPS and serve as inputs for TPR calculations. Accuray recommended initiating the optimization by calculating TPRs for small (15.4 × 15.4 mm²) and large (84.6 × 84.7 mm²) field sizes at five depths (10, 15, 100, 200, and 300 mm). The statistical uncertainty for MC calculations was set to 0.5%, with S set to 1.8 mm, as per Accuray's guidelines. TPRs were calculated using Emax values ranging from 6.6 to 6.8 MeV in 0.1 MeV increments for the specified field sizes and depths. [5]

The open field fluence distribution generated from open beam profile maps for the Finite Size Pencil Beam (FSPB) algorithm was reviewed, an appropriate beam hardening option was chosen, and an initial energy spectrum of 6.6 MeV was used. Calculated TPR values were compared to measured values, with acceptable differences observed (less than 1%)

For optimizing the source FWHM, the source distribution, which represents the photon emission direction from the target, was modelled as a Gaussian function. The photon source FWHM (S) was initially set to 1.8 mm. Based on whether the LINAC exhibited beam hardening, we selected an energy spectrum, starting with an input of 6.6 MeV for profile calculations. Following Accuray's recommendations, profiles were calculated for two field sizes (15.4 × 15.4 mm² and 84.6 × 84.7 mm²) at a depth of 100 mm. Profiles were computed for S values of 1.4, 1.6, and 1.8 mm and compared to the measured profiles, with the small field size profile comparison being particularly influential in determining the final value of S. [5]

3. Results

Model Validation

Output Factors: Once the maximum energy (Emax) and source size (S) were optimized, output factors were calculated for all field sizes without further adjustments. These calculated values were compared to measurements taken at a source - axis distance (SAD) of 800 mm. In accordance with AAPM guidelines for medical linear accelerator quality assurance, output factors were considered acceptable if the differences between calculated and measured values were less than 1%.

Additionally, output correction ratios (OCRs) were calculated at a depth of 100 mm for field sizes of 15.4 × 15.4 mm² and 84.6 × 84.7 mm², with a statistical uncertainty of 0.5%. The calculated OCRs in the penumbra region (80% - 20%) were compared with measured values and found to be within acceptable limits.

To ensure accurate dose calculations for all field sizes and depths, TPRs and OCRs were calculated for all configurations, closely aligning with measured data. Output factors were calculated to a 0.2% uncertainty. A reference depth of 1.5 cm was used for all collimators and field sizes. All measurements were performed using a PTW MicroDiamond chamber in a PTW MP3 water tank. [5, 8]

Determination of Optimized Source FWHM: To initiate source modelling, we set the photon source FWHM to 1.8 mm. We reviewed the open field fluence distribution created from open beam profile maps obtained during FSPB measurements. Based on this analysis and Accuray's recommendations, we selected the appropriate beam hardening option and set the initial energy spectrum to 6.6 MeV. [8] See Figure 1:

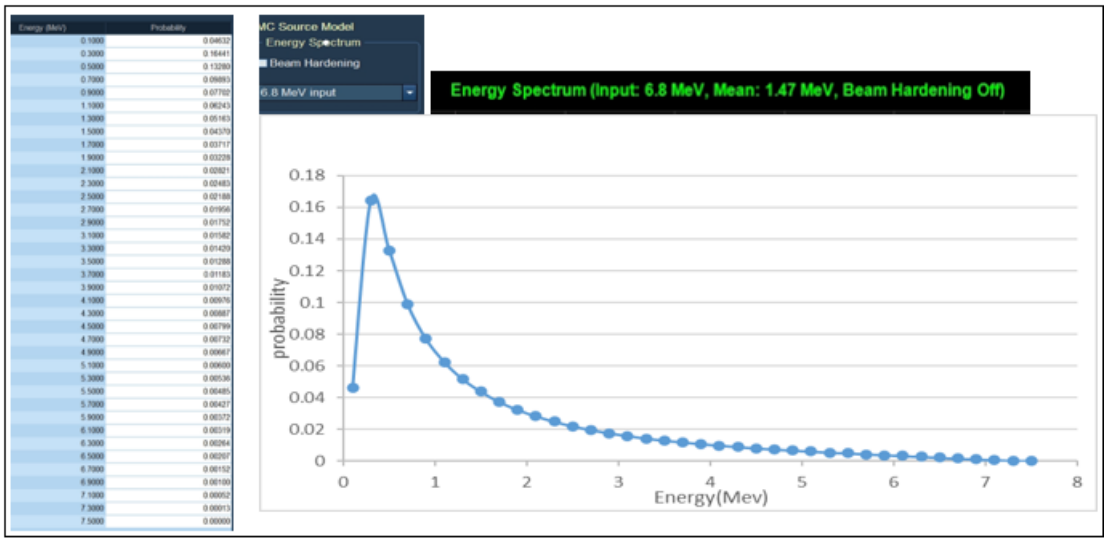
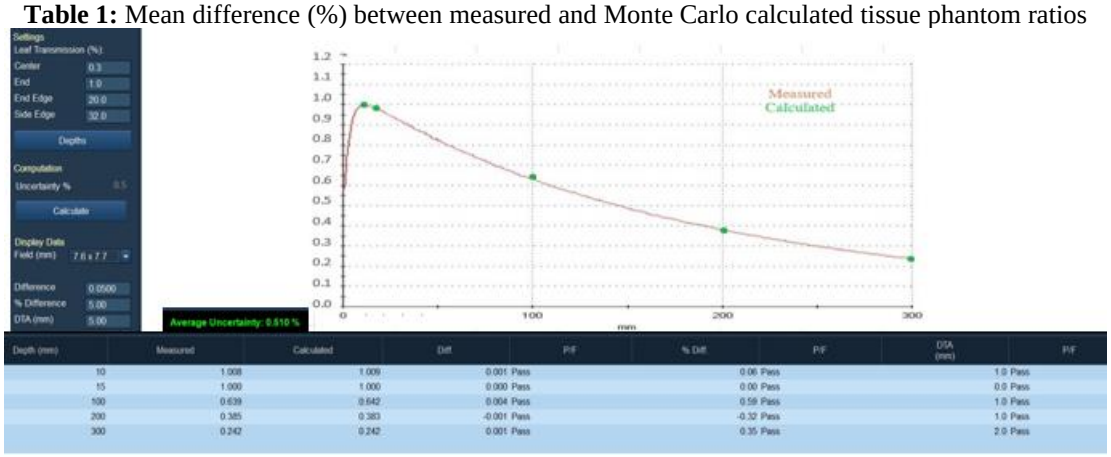


Figure 1: Determination of Optimized Source FWHM

MLC MC TPR Step

We calculated the tissue phantom ratios (TPRs) and verified them for consistency with measured data, maintaining a statistical uncertainty of 0.5%. This process included field sizes of 15.4 × 15.4 mm² and 84.6 × 84.7 mm² at depths of 10, 15, 100, 200, and 300 mm. To enhance alignment between calculated and measured TPRs, we adjusted the energy spectrum selection as needed, recalculating and comparing the TPRs after each modification. Once the calculated TPRs matched the measured data to the desired accuracy, we moved to the next step in the validation process. ([5, 8]) See TABLE 1.

Table 1: Mean difference (%) between measured and Monte Carlo calculated tissue phantom ratios



Depth (mm)	Measured	Calculated	Diff	P/F	% Diff	P/F	DTA (mm)	P/F
10	1.008	1.009	0.001	Pass	0.06	Pass	1.0	Pass
15	1.000	1.000	0.000	Pass	0.00	Pass	0.0	Pass
100	0.639	0.642	0.004	Pass	0.59	Pass	1.0	Pass
200	0.385	0.383	-0.001	Pass	-0.32	Pass	1.0	Pass
300	0.242	0.242	0.001	Pass	0.35	Pass	2.0	Pass

MLC Monte Carlo OCR Step

We calculated the output correction ratios (OCRs) and compared them to measured data, maintaining a statistical uncertainty of 0.5%. This evaluation was performed at a depth of 15 mm for a field size of 7.6 × 7.7 mm². To further refine the model and improve the agreement between calculated and measured OCRs, we adjusted the source size FWHM. After each adjustment, OCRs were recalculated and compared to the measured data. Once a satisfactory match was achieved, we proceeded to calculate TPRs and OCRs for all field sizes and depths. ([5, 8]) See Figure 2.

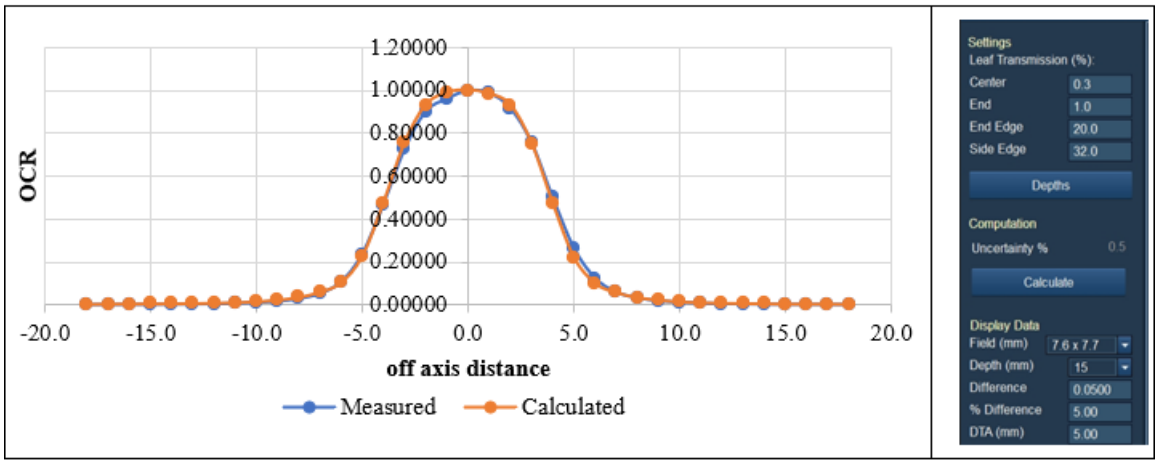


Figure 2: MLC Monte Carlo OCR, measured and calculated Values.

MLC Monte Carlo OF Step

After confirming the accuracy of TPRs and OCRs by comparing them to measured data, we calculated the output factors (OFs) with a statistical uncertainty of 0.2%. These calculated OFs were then compared to the measured data. Upon successful validation, the results were approved for phantom deliveries.[5, 8] See Figure 3

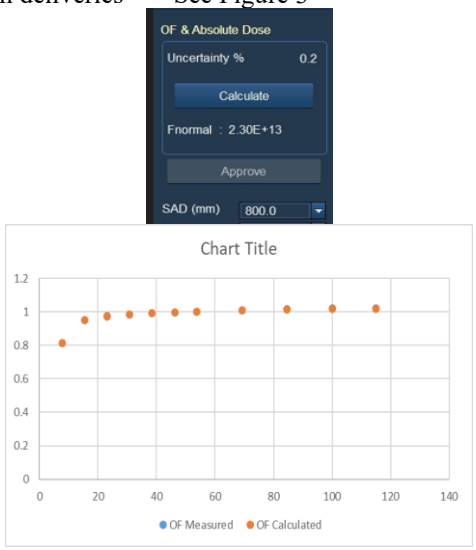


Figure 3: MLC Monte Carlo OF. Measured and Calculated Values.

MLC Monte Carlo Final Approval: Final approval for patient plans is granted after algorithm verification measurements are completed. Leaf transmission settings are maintained at their default values. Monte Carlo utilizes a CT number to mass density table to generate patient model. Accuray Precision System supports both conventional 12 - bit CT scans (4096 discrete CT values) and 16 - bit CT scans (65536 discrete CT values) with Max HU Number allowed in table is 31743. Value of density curve at HU of 31743 is used for all HU values higher than 31743. The mass density curve should have a density of 0.001 g/cm3 for - 1000 HU.

Verification:

Phantom Measurements:

We validated the Monte Carlo (MC) model by comparing TPS - calculated doses to those measured in various phantoms. Measurements were performed using the Standard Imaging Stereotactic Dose Verification Phantom recommended by Accuray (see Figure 5a &5b). For the phantoms shown in Figure 5b, ionization chamber measurements were taken with a PTW Farmer ionization chamber, calibrated for the MLC. Additionally, EBT3 Gafchromic films (Ashland Inc., Wayne, NJ, USA) were used with the phantom depicted in Figure 5b. Each dose measurement was adjusted for daily output variations of the CyberKnife.

The model validation process comprised five steps. In steps 1 through 4, we assessed the dose differences (ΔD) between MC model calculations and measurements across various configurations to evaluate model accuracy. A maximum ΔD of ±2% was set as the threshold for clinical acceptability, while a ΔD within ±1% was considered highly accurate. Step 5 involved calculating actual patient treatment plans. Details of each step are provided in the following sections.

Step 1 - Single Beam in Homogeneous Phantom (Ionization Chamber Measurements):

The Single Beam QA path was set up with a single node positioned 800 mm from the alignment center along the Z - axis. This setup allowed us to assess calculation accuracy across a range of beam and phantom geometries, covering percentage depth doses (PDDs), field widths, buildup regions, heterogeneity corrections, oblique incidences, and irregular surfaces.

The aim of this step was to compare Monte Carlo (MC) - calculated doses with measured doses in a homogeneous phantom (Figure 4), focusing on the center of the beam at a depth of 5 cm. Ten treatment plans with different equivalent square field sizes, ranging from 20.0 to 55.0 mm, were created in the TPS using a single beam incident perpendicular to the phantom surface. These plans were then exported to the CyberKnife system and delivered to the phantom. Dose measurements were taken with a PTW Farmer - type ionization chamber and compared to the TPS - calculated dose to determine the dose difference (ΔD). A ΔD close to 0 indicated greater accuracy of the MC algorithm.

Additionally, to assess whether the MC algorithm would be suitable for all clinical applications, including homogeneous regions where the FSPB algorithm could also be used, we calculated ΔD for five of the ten plans using the FSPB

algorithm and compared these dose differences with those from the MC algorithm.

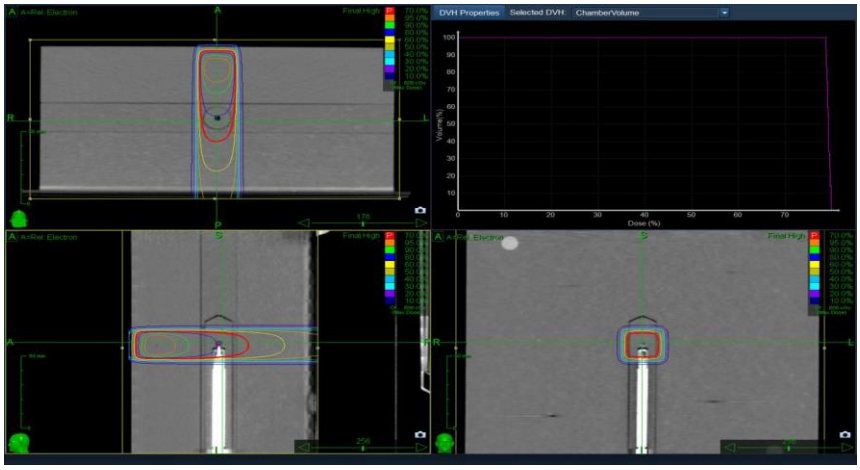


Figure 4: Single Beam in Homogeneous Phantom

Step 2 - Single Beam in Homogeneous Phantom (Film Measurements): To assess the accuracy of the MC model in high - dose gradient regions, we used EBT3Gafchromic films ([6]). These films were calibrated on an Elekta Infinity linear accelerator (Elekta AB, Stockholm, Sweden) with a PTW Farmer ionization chamber (PTW, Germany) at a 6 MV beam energy. The energy independence of Gafchromic films enabled their effective use with the CyberKnife beam. The uncertainty for film dosimetry was estimated to be ±2%. Five plans were designed to generate fields of various dimensions, and dose profiles were measured with EBT3 Gafchromic films in a homogeneous phantom (Figure 5b) at a depth of 5.1 cm. To compare the measured and MC - calculated doses, we performed a gamma index (GI) analysis on each film using a local criterion of 2% of the maximum dose difference (DD), a 2 mm maximum distance - to - agreement (DTA), and a dose threshold of 10% of the maximum dose.

Step 3 - Single Beam in Heterogeneous Phantom: This step followed the same methodology as Step 1, but calculations and measurements were performed using a heterogeneous phantom with a lung insert (Figure 5a). Ten plans with various equivalent square field sizes, ranging from 20.4 to 45.2 mm, were created using a single beam. Dose measurements were taken at the beam center at a depth of 10 cm using a PTW Farmer - type (0.6 cc) ionization chamber.

Step 4 - Multiple Beams in Homogeneous and Heterogeneous Phantoms: In this step, we calculated four plans in the homogeneous phantom (Figure 5b) using 8, 12, 14, and 26 beams at various entry angles. Dose measurements were taken at the beam center at a depth of 5 cm in the homogeneous phantom. The same approach was applied to a heterogeneous phantom (figure 5a), where 6, 10, 14, and 20 beams were used with different entry angles. In the heterogeneous phantom, dose measurements were taken at the beam center at a depth of 10 cm.

Figure 5a

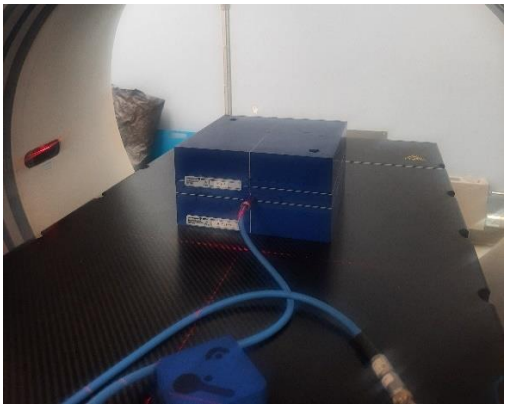


Figure 5b

Figure 5a and 5b: Standard Imaging Stereotactic dose verification Phantom

Step 5 - Patient - Specific QA with Monte Carlo and SRS MapCheck: To ensure that the dose delivered to the patient closely matched the planned dose, we used the Sun Nuclear SRS MapCheck device, calibrated with known beam configurations. A StereoPHAN phantom was employed to simulate patient anatomy, with the SRS MapCheck positioned to record the delivered dose. Independent Monte Carlo calculations were conducted to predict the dose distribution, which was then compared with the measurements obtained from the SRS MapCheck.

Gamma Passing Rates: In this study, we analyzed 20 patient plans and found that using a 2%/2 mm criterion for gamma analysis resulted in passing rates exceeding 95%. This high agreement between planned and delivered doses indicates that the MC calculations effectively identified potential issues in treatment plans before delivery. (2)

Clinical Implementation: Incorporating Monte Carlo calculations into the QA workflow improves both the efficiency and accuracy of the verification process, promoting enhanced patient safety and treatment outcomes.

Figure 6: Patient Specific QA with SRS MapCheck and StereoPHAN Phantom.

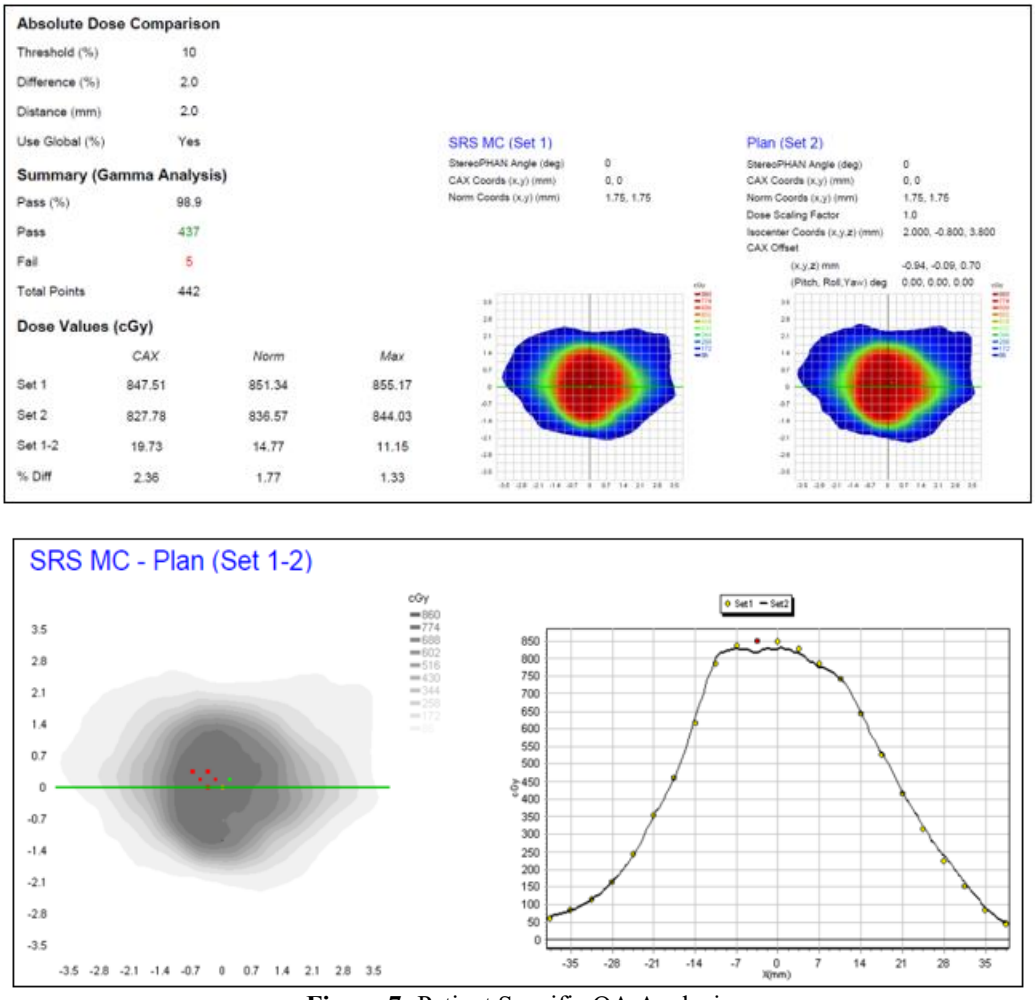


Figure 7: Patient Specific QA Analysis

4. Conclusion

We successfully developed a beam model for the Monte Carlo (MC) algorithm integrated with the multileaf collimator (MLC) within the Precision CyberKnife Treatment Planning System (TPS). Our findings highlighted notable differences between the MC calculations and the measured tissue phantom ratios (TPRs) and dose profiles. However, the dose differences observed across various configurations remained within acceptable limits, confirming the clinical adequacy of the MC algorithm despite some accuracy limitations.

All patient - specific quality assurance (QA) evaluations met the established criteria for clinical acceptance of treatment plans. Consequently, we conclude that while the MC model paired with the MLC provides significant advantages, its application should be approached with caution due to its inherent accuracy constraints.

Additionally, the integration of Monte Carlo algorithms with the SRS MapCheck device establishes a robust framework for patient - specific QA, thereby enhancing the reliability of radiation therapy treatments. By ensuring that the delivered dose closely matches the planned dose, these methodologies contribute to improved patient safety and treatment effectiveness.

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