Validation of a commercial Monte Carlo algorithm for stereotactic radiosurgery and stereotactic body radiation therapy

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Abstract

This thesis aimed to validate the Monte Carlo (MC) algorithm in BrainLab’s iPlan treatment planning system, used in conjunction with stereotactic radiosurgery (SRS) mode of the Varian Novalis TX linear accelerator for clinical use. Specifically, the iPlan algorithm was “benchmarked” by comparing results obtained with a BEAMnrc model developed for the Novalis TX’s SRS mode. The BEAMnrc model was obtained by modifying an existing model for a Varian linac to include the different SRS flattening filter and the high definition 120 leaf multi-leaf collimator (HD120 MLC) of the Novalis TX. Characterization of the source model used a recently published procedure to fit beam energy, source size and angular spread, and an existing BEAMnrc Monte Carlo component module (DYNVMLC) was reprogrammed to model the HD120 MLC of the Novalis TX linac. For the latter, the interleaf air gap and leaf density were adjusted such that simulations matched interleaf leakage profiles measured with film. Validation of the iPlan MC algorithm was accomplished through comparisons between both MC codes and film measurements for MLC defined fields, depth dose curves of square fields incident on heterogeneous slab phantoms, and more clinically realistic plans incident on a Lucy® stereotactic QA phantom and a Rando® head phantom. The source characterization procedure and the modeling of the HD120 MLC were successful, with subsequent simulations performing well compared to measurements of output factors, profiles in water and dose planes of MLC defined fields. Some discrepancies were observed between either MC code and film measurements, but calculations with iPlan MC and EGSnrc MC codes agreed well with each other in all cases. These results suggest that the iPlan Monte Carlo dose calculation algorithm is capable of accurately predicting radiation dose for complex fields in heterogeneous media.
Résumé

Ce mémoire visait à valider l’algorithme Monte Carlo (MC) dans le cadre du système de planification de traitement iPlan de BrainLab où il est conjointement utilisé avec le mode radiochirurgical stéréostatique (SRS) de l’accélérateur linéaire Novalis TX de Varian. Plus particulièrement, l’algorithme iplan a été validé en comparant les résultats obtenus avec un modèle BEAMnrc du mode SRS de Novalis TX. Le modèle BEAMnrc a été créé en modifiant un modèle existant d’accélérateur Varian afin d’y inclure le filtre compensateur SRS et le collimateur multilames de haute définition de Novalis TX (HD120MLC). La caractérisation de la source a utilisé une procédure récente pour ajuster l’énergie, la taille et l’ouverture angulaire de la source. Par ailleurs, un module multilames de BEAMnrc existant (DYNVMLC) a été reprogrammé pour simuler le collimateur multilames de haute définition (HD120MLC). Pour ce dernier, l’écart entre les lames et la densité des lames ont été ajustés de sorte que les simulations correspondaient aux profils de fuites interlames mesurées par films. La validation de l’algorithme iPlan a été réalisée par comparaisons entre les deux codes MC et des mesures de films pour des champs définis par le collimateur multilames, pour des courbes de la dose en profondeur de champs carrés administrés sur des fantômes hétérogènes et de plans plus réalistes du point de vue clinique administrés sur un fantôme stéréostatique Lucy® et un fantôme de tête Rando®. La procédure de caractérisation de la source et la modélisation du collimateur multilames (HD120MLC) ont été réussies ainsi que les simulations ultérieures correspondaient bien aux mesures des facteurs d’ouverture, des profiles dans l’eau et des distributions de dose des champs définis par le collimateur multilames. Des différences ont été observées entre les codes MC et les mesures de films, mais les calculs avec les codes IPlan MC et EGSnc correspondaient bien dans tous les cas. Ces résultats suggèrent que l’algorithme MC de Iplan peut prédire précisément les doses de rayonnement pour des champs complexes dans des médias hétérogènes.
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Chapter 1

INTRODUCTION

1.1 Stereotactic techniques

Stereotactic techniques, such as stereotactic radiosurgery, or SRS, and stereotactic body radiotherapy, or SBRT, are designed to deliver large doses of radiation to a highly localized area in one or few fractions. The etymology of the word "stereotactic" combines "stereo" which means three-dimensional in Greek, and the Latin word for touch: "tactus." By its definition, stereotactic treatments differ from conventional radiotherapy primarily in the degree of positional accuracy required to perform the treatment.

Figure 1.1: An example of a stereotactic head frame.
The origin of stereotactic techniques was in the late 1940s, developed by Lars Leksell, a neurosurgeon, to destroy dysfunctional loci in the brain. [1, 2] While the first 3D treatment of a brain lesion with x-rays was performed in 1948 [3], Leksell performed his first radiosurgery, as described, in 1950. [2, 4] Today, a rigid metal frame is fixed to the patient's skull, so that a coordinate system can be imposed on CT or MRI images. An example of a stereotactic frame can be seen in Figure 1.1. Fiducial markers on the frame can be located on the CT images, placing each CT slice in an exact location within the frame. An intracranial target's location, therefore, can be defined precisely in reference to the frame. Following localization, a large number of non-overlapping small fields are used to deliver a large radiation dose to the target with a steep fall-off to spare healthy brain tissue. The treatment is delivered in a small number of fractions. This technique is formally known as stereotactic radiosurgery and it can be accomplished using a Gamma Knife irradiator™, a linear accelerator or with protons.[1] Stereotactic radiosurgery is defined by the accurate location and definition of the target in reference to a stereotactic frame, and the large concentrated dose with rapid dose fall-off delivered in few fractions. Radiosurgery usually implies intracranial treatment, as opposed to SBRT, which is extracranial.

SBRT, or stereotactic body radiotherapy, is an extension of SRS to extracranial lesions. SBRT is used to treat areas in abdominopelvic and thoracic cavities, as well as at spinal or paraspinal sites.[5] In the same manner as SRS, SBRT delivers high localized doses using a large number of small fields in few fractions. SBRT also places demands on spatial accuracy and image-guided target localization has allowed the extension of stereotactic techniques to extracranial sites without the use of a rigid frame. The technique is further complicated by the motion present at extracranial sites, which introduces problems for imaging and treatment. Multiple imaging modalities, such as 4D-CT, MRI and positron emission tomography, can be used to define the location of the target. Several techniques exist to manage moving targets, such as breath-hold techniques and respiratory gating. The reader can refer to the report of the AAPM task group 76 for a more complete understanding on how respiratory motion is managed in radiation oncology.[6] Although motion management factors in greatly into the imaging and treatment for SBRT, it is not focused on in this work. In summary, when compared to conventional radiotherapy, SBRT differs by having a high dose per fraction, a small number of fractions, a large number of beams used for treatment, tighter target margins, stricter geometric verification and respiratory motion management.[5]

1.1.1 Dosimetric challenges

As mentioned above, stereotactic techniques employ a large number of small fields to achieve a high localized dose with a steep dose falloff. Small fields pose a serious dosimetric
challenge, both for measurement and dose calculation. A “small” field is a relative concept, but commonly, can be considered as such if it has one dimension smaller than 3 cm. Specifically, a field is small when it presents three issues: the collimating devices occlude the source, the field size is comparable to the lateral range of electrons resulting in a loss of lateral charged particle equilibrium, and the field size is comparable to the size of the devices used to measure it. [7] The occlusion of the source and the lack of lateral charged particle equilibrium results in a lower output, and an underestimation of field size, respectively, as shown in Figure 1.2. The presence of the detector can significantly perturb the charged particle fluence of small fields. Furthermore, given the Gaussian shape of small fields, the sensitive volume of a measuring device may not be uniformly irradiated (if the detector is large compared to the field), resulting in an erroneously low dose measurement. Finally, small fields are subject to significant perturbations in low density medium such as lung tissue. [7] As will be discussed in section 1.2.1, many dose calculation algorithms can fail to accurately predict these effects.

In SBRT, the problems of small field dosimetry are amplified, as a large number of fields are incident on the heterogeneous geometry of the abdominopelvic and thoracic areas. As will be discussed in the following section, the accuracy of dose calculation algorithms depends firstly on the input data entered in by a physicist. Inaccuracies in the measurement used in the input data for small fields will be reflected in the calculations. And these errors will, in turn, be amplified by the number of fields used and the complex geometry in SBRT cases. As will be discussed in section 1.5, the purpose of this work will be to evaluate a tool designed to address the dosimetric challenges associated with SBRT, specifically the iPlan Monte Carlo dose calculation algorithm located within the iPlan treatment planning system.
Figure 1.2: A diagram illustrating some problems associated with small fields. Given that field sizes are defined by their full-width at half-maximum, or FWHM, (indicated by the black horizontal arrows), as field size decreases, this definition breaks down as the measured FWHM would be larger than the actual field size setting. Furthermore, the source is obscured by the collimating devices in smaller fields, resulting in a lower output. [7]

1.2 Treatment planning

The modern radiotherapy process consists of several steps: 1) acquisition of CT or MRI images, 2) image registration, 3) delineation of structures, such as the tumour and organs at risk, 4) selection of beam arrangements and weights, 5) dose calculations to predict what dose would be delivered by the planned beam arrangement, 6) review by physicists and physicians, 7) exportation of beam setup to the treatment machine, and 8) the treatment itself. Steps 1-6 could be said to be part of the “treatment planning” process, where dosimetrists, physicists and physicians collaborate to craft an appropriate treatment that achieves the prescription dose to the target while minimizing doses to organs at risk. The CT or MRI images are used for planning purposes to visualize the patient geometry. On these images, structures are contoured, which essentially means that they are outlined, so that any dose given to the contoured structures can be kept track of. The macroscopic tumour structure visible on the image is labelled as the gross tumour volume, or GTV. Surrounding the GTV, a clinical target volume, or CTV, is drawn to encompass microscopic disease not visible on the image. In SRS or SBRT, it is common to assume that the GTV and the CTV are the same. And finally a planning target volume, or PTV, is drawn around the CTV to account for errors in setup or delivery. For stereotactic techniques, PTV

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margins are small to non-existent due to improved target localization. [8] A physician then prescribes a dose to the PTV and sets limits on doses that can be delivered to organs at risk. Given restrictions imposed by physicians and physicists, steps 4-6 would be repeated until an acceptable plan is reached. The treatment planning system that is the focus of this work is the iPlan treatment planning system and it is described in section 2.1.2. There are many modalities within the umbrella term of radiotherapy that can be used to treat lesions, depending on its location and nature. In this work, the SBRT treatment with linear accelerators is the focus. Therefore, the gantry angle, couch angle, collimator angle, jaw position and MLC shape can be adjusted for each field to achieve an optimal treatment plan. The specifics of the Novalis Tx radiosurgery unit (Varian Medical systems, Palo Alto, CA) will be expounded in section 1.4, as the Novalis is the machine used for SBRT treatments at the Montreal General Hospital.

1.2.1 Dose calculation

Within each treatment planning system, there are dose calculation algorithms that predict the dose to be delivered to the patient. In this work, the sole focus will be photon dose calculations. The ability to accurately deliver a prescription dose to a target within a patient is dependent first and foremost on calibration measurements made under reference conditions. Additionally, before dose can be predicted in more complex patient geometry, the radiation beam must be thoroughly understood in homogeneous water. The input of every treatment planning system is different but it typically consists of percent depth dose curves, cross beam profiles, and relative output factors in water. It is assumed, in this work, that the reader has an understanding of what this data set would look like. Dose calculation algorithms then extrapolate from this measured data to calculate dose to points of interest within a patient. However, the patient, in reality, is not composed of pure water, but is composed of variety of tissues and cavities with radiological properties different from water, such as lung, bone, and air cavities. Therefore, an algorithm’s ability to accurately predict radiation dose is contingent on the accuracy of the input data, the extrapolation of this input data to patient cases, and the ability to account for tissue inhomogeneity. [9] The photon dose calculation problem can be summarized in Figure 1.3. Other sources of inaccuracy in the radiotherapy process include limitations in diagnostic imaging and target localization, and errors in set-up and treatment of the patient.
The specific aspect of tissue inhomogeneity correction within the process of dose calculation is of interest to this work. There are a variety of techniques to correct for inhomogeneity employed by different treatment planning systems that vary in calculation speed and accuracy. Until the late 1970s, most dose calculations were done assuming patients were composed entirely of water, as there was a lack of patient-specific anatomical information. With the advent of the CT scanner, it became possible to begin accounting for heterogeneities within a patient as the Hounsfield numbers of the CT image could be converted into electron densities. [10] With patient specific information, calculations could take one of two forms: one where correction factors are applied to adjust a distribution calculated in water, or one where dose is calculated a priori in heterogeneous media. A simple example of the first form of calculation is to correct the dose on a point by point basis by calculating an effective attenuation coefficient based on an effective radiological depth. The inhomogeneity correction factor (ICF) can be described in the following equation:

\[
ICF = e^{\mu'(d-d')}
\]  

(1.1)

Where \( \mu' \) is the effective attenuation coefficient of water for a particular beam quality, \( d \) is the physical depth from the surface to the point of calculation, and \( d' \) is the “radiological depth,”
“equivalent depth,” or “equivalent path length” and is calculated by summing the depth to the point weighted by mass density. [10] It must be noted that this method, and methods similar to it, only can perform 1D corrections, precluding any corrections for lateral scattering. More complex techniques can account for 3D density information, such as the equivalent tissue-air ratio, or ETAR, method. [11] However, such methods still make certain assumptions such as charged particle equilibrium. More complex methods, such as superposition-convolution, that calculate dose a priori in heterogeneous media, do not make as many assumptions. In general, superposition-convolution techniques involve convolving an energy fluence distribution with a scatter spread kernel. [12] Based on the complexity of the algorithm, the fluence and scatter kernels can be considered in one or several dimensions. In general, the more complex the technique, the slower the calculation, but greater calculation accuracy is achieved. Depending on the site of a lesion and the surrounding heterogeneities, certain algorithms may be sufficient. The geometry of the brain, for example, is less complex than that of the lung.

The accuracy of dose calculation plays a significant role in the overall accuracy of the radiotherapy process. The sources of uncertainty of the whole process as reviewed by a large number of authors are summarized in table 1 in AAPM’s report number 85. [10] Considering the estimations of present uncertainties, the uncertainty of dose calculation algorithms must be under 3% to achieve the overall recommendation of 5% uncertainty in delivered dose. [10] These estimations are contextualized by how uncertainty plays into radiobiological models, how dose uncertainty affects the sample size required for clinical trials, and what dose difference can be detected clinically. Current radiobiological models have established that normal tissue complication probability (NTCP) and tumour control probability (TCP) have sigmoidal dependence with dose. [13] Given the steepness of the sigmoidal curves, a 5% difference is delivered dose could result in as much as a 10-20% difference in TCP, and a 20-30% difference in NTCP. [14] In clinical trials, a sample size must be sufficiently large to demonstrate statistically significant findings. One study by Orton et al. showed that increasing dose calculation accuracy (by applying heterogeneity corrections) allowed for a smaller sample size. [15] And finally, a 7% difference in dose has been shown to be clinically detectable.[16]

In general, the AAPM task group on inhomogeneity corrections recommends applying inhomogeneity corrections to increase dosimetric accuracy, although each site poses unique challenges. Specifically, the sites at the head and neck and near the thoracic cavity require more complex algorithms to account for heterogeneities and loss of electronic equilibrium in low density areas such as lung or air cavities. [10] It is, however, the responsibility of each individual institution to ensure that the results obtained from a dose calculation algorithm are indeed valid.
Every dose calculation algorithm must, therefore, be validated thoroughly, as will be explained in section 1.3.

1.2.2 The Monte Carlo Method

The Monte Carlo method, presented explicitly in section 2.1, is considered now as the gold standard for dose calculation accuracy. The Monte Carlo method does not make any assumptions concerning energy fluence or scatter, as done by other inhomogeneity corrections, but rather simulates explicitly the interactions of photons with matter. From a theoretical standpoint, Monte Carlo dose calculation algorithms have the potential to solve the dosimetric challenges presented by SBRT, as described in section 1.1.1. They have the ability to circumvent the problems associated with small fields, as well as being able to fully account for inhomogeneity in the patient. However, it must be stated here that Monte Carlo techniques are still only models of radiation transport, the assumptions they make are simply smaller than other algorithms. In this work, the Monte Carlo dose calculation algorithm within the iPlan treatment planning system will be investigated, as described in section 2.1.2.

1.3 Commissioning of a dose calculation algorithm

By definition, commissioning of the dose calculation algorithm is the process in which a dose calculation is readied for clinical use, with knowledge of where the algorithm is appropriate and where it is not. Commissioning begins with performing the necessary input measurements for the algorithm, and it ends with a clinical physicist allowing the algorithm to be used clinically. Once the algorithm is functioning, the purpose of the algorithm, and most importantly, the judgement of the clinical physicist in charge of the commissioning determine the extent that the algorithm is tested to ensure that it is fit for use. A certain planning system, for example, may only be used for photon treatments, so its commissioning will not include tests concerning electron treatments. A newly acquired algorithm may already be well understood so the testing might include a literature review and relatively few measurements. In general, however, it is recommended that a detailed testing of the algorithm be performed to minimize any errors in the treatment planning process.

At the end of the commissioning process, the overall testing of an algorithm includes the uncertainty associated with all the steps taken by a treatment planning system to predict dose. The overall testing process bifurcates into “validation” and “verification.” Validation is the process of checking the validity of physics approximations made by the algorithm. As all dose calculations algorithms must make assumptions to make the calculation feasible, and Monte Carlo simulations are no exception. Only correcting for scatter in one dimension, for example,
may not be valid in lung tissues. Verification, in contrast, is the process of checking that the
algorithm is indeed doing what it is supposed to do. Errors in programming, for example, could
be responsible for improper extrapolation of the algorithm’s input data. [18] The end to end
testing simultaneously combines validation and verification, checking the functioning of a system
as a whole by comparing to measured data. The disadvantage of end to end testing is that
discrepancies between calculations and measured data may be a result of invalid assumptions
made by the algorithm, or they may simply be the result of a programming error.

The reader is asked to refer to Figure 1.4 to contextualize the above paragraph.
Verification is depicted in the figure in between steps, implying that there can be errors in
translation. There might be errors in the interpretation of the CT image by the treatment
planning system, for example, without anything being wrong with the CT images. Another
perspective is to consider the interface between the planning system and the treatment
machine. One needs to verify that the patient specific beam arrangements are properly
communicated to the treatment machine: that a specific planned gantry angle actually translates
to that angle at the machine. However, the details surrounding the quality assurance of a linear
accelerator is outside the scope of this work. It is presented here to highlight the uncertainties of
the commissioning process. Validation is shown between the dose distribution and the dose
calculation to imply that the algorithm’s assumptions must be tested in different geometries to
determine where the algorithm is appropriate and where it is not. And finally, the overall
validation, or testing, is shown between dose distributions and measurements, incorporating all
aspects of the process.
Therefore, a thorough evaluation of a dose calculation algorithm requires verification, validation and a broader overall testing. This work, however, focuses solely on the overall testing of the iPlan Monte Carlo dose calculation algorithm, so specific validation and verification tests will not be discussed here. For information of specific validation of dose calculation algorithms in heterogeneous media, please refer to section V of the AAPM report no. 85 on tissue inhomogeneity corrections. [10] For verification of many aspects of a treatment planning system, the reader can refer to Jacky and White, 1990. [18]

Overall validation tests for photon beams, as compiled by Van Dyk et al. [9], are arranged into three categories: point dose calculations and factors, dose distributions in homogeneous media and dose distributions in inhomogeneous media. The point dose calculations and factors include tissue-air-ratios (TAR), tissue-phantom-ratios (TPR) and percent depth dose (PDD) data for square, rectangular and irregular fields. This category also includes
inverse square law corrections, as well as factors to compensate for attenuators in the path of the beam (wedges, compensators, etc.). The dose distributions in homogenous media include profiles at multiple depths for square and rectangular fields, as well as combinations of multiple beams. Oblique incidence, off-axis calculations and collimator rotation are also tested. Finally, in the category of dose distributions in inhomogeneous media, the accurate transfer of CT images and their conversion to electron density are tested, as well as a single square field incident on a number of different phantoms containing heterogeneous media such as air or bone. [9] The criteria for acceptability for such tests depend on the nature of each test, and also the region where the agreement to measurements is being examined. More importance is placed on agreement within a treatment field, as opposed to outside the field, for instance. Many authors have compiled criteria for acceptability, but for the reader’s convenience, table 4-4 of AAPM’s Radiation Therapy Committee Task Group #53 report [19] provides a concise overview of the criteria.

While the final part of the commissioning process always relies on measurements similar to those described in the paragraph above, another way to test an algorithm is to compare it to another trusted algorithm. Typically, one commissions an algorithm by comparing it to another more accurate algorithm.[20-23] Monte Carlo simulations are often used as the benchmark other dose calculation algorithms, as Monte Carlo simulations are considered as the goal standard for dose calculation. Given that EGSnrc is a widely available and trusted Monte Carlo code, it will be used to benchmark the iPlan Monte Carlo dose calculation algorithm in this work.[24] Both the EGSnrc and iPlan Monte Carlo algorithms are described in section 2.1. The comparison to EGSnrc complements the validation measurements to aid in commissioning the iPlan code, as the EGSnrc Monte Carlo code can provide a standard where measurements are difficult or impossible.

1.4 The Novalis Tx radiosurgery unit

The treatment machine focused on in this work is Varian’s Novalis TX radiosurgery unit (Varian Medical Systems, Palo Alto, CA) shown in Figure 1.5. The Novalis TX is a versatile linear accelerator which can perform radiosurgical techniques as well as conventional radiotherapy. It is capable of treating patients with a variety of electron and photon energies. Of special interest to this work is the stereotactic, or SRS, mode (Energy ≈ 6 MeV) of the Novalis TX, which is capable of delivering 1000 MU/minute, as opposed to the 600 MU/minute in normal mode. The Novalis TX accomplishes the high dose rate by utilizing a significantly smaller flattening filter when SRS mode is selected. Given that SRS mode is typically used to treat smaller lesions, the small size of the flattening filter is made possible by restricting the field size available in SRS mode. The
maximum field size in SRS mode is 15 x 15 cm$^2$, as opposed to 40 x 40 cm$^2$ in normal mode. The Novalis TX is equipped with the HD120 MLC, as can be seen in Figure 1.6. Multileaf collimators are described in a general manner in section 2.3. The HD120 MLC has three different types of leaves: quarter leaves, which project 0.25 cm at isocenter; half leaves, which project 0.5 cm at isocenter and outboard leaves, which are located at the outside edges of each leaf bank. All types of leaves have both isocenter and target versions, depending on whether the thicker part of the leaf is closer to the isocenter or target, respectively. The small size of the quarter leaves allows for the delineation of conformal fields for small lesions, in comparison to the larger leaves of a previous Varian MLC: the Millennium MLC. The Novalis TX contains on-board imaging devices that acquire kV planar images and software to allow for cone beam CT reconstruction. The Novalis TX is unique in that it has a stereoscopic imaging system that is used in conjunction with a camera system and couch robotics for stereotactic localization. Furthermore, an ExacTrac system is employed for target localization and motion management. The acquisition of images during treatment and the use of the ExacTrac system allows for reduced target margins and frameless stereotactic treatments.
1.5 Purpose of this work

The purpose of this work is to validate the iPlan Monte Carlo dose calculation algorithm by benchmarking it with a BEAMnrc model of the Novalis TX and comparing both Monte Carlo codes to measurements. This work is part of the commission process for the iPlan Monte Carlo dose calculation algorithm. In order to model the Novalis TX, an accelerator model in BEAMnrc is modified to include the SRS flattening filter and the HD120 MLC. BEAMnrc is used to model accelerators and is part of the EGSnrc Monte Carlo code system: please refer to section 2.1.1 for more information. To model the HD120 MLC, a module used to model the Millennium MLC is reprogrammed to account for the different leaves present in the HD120 MLC, due to the similarity of the leaf shapes of the two MLCs. Several free parameters within the BEAMnrc model of the Novalis TX are adjusted to match measured data. In a similar way to commissioning the dose calculation algorithm, the Novalis TX model is commissioned by comparing to measurements. Following the creation of the model, three way comparisons between film or ion chamber measurements, the BEAMnrc model and the iPlan Monte Carlo code are made in homogeneous water, heterogeneous slab phantoms and in a RANDO® phantom.[25] With these three way comparisons, the performance of the iPlan Monte Carlo algorithm will be assessed.
This work serves to complement additional commissioning work done to incorporate the iPlan Monte Carlo dose calculation algorithm into clinical use here at the Montreal General Hospital.

1.6 References


Chapter 2
THEORY

2.1 Monte Carlo dose calculation

The Monte Carlo method is a stochastic technique for solving complex problems. Originally developed in the late 1940s to study nuclear processes, the basic idea behind the technique is that individual particles are transported throughout geometry where random number generators are used to determine the interactions of that particle. [1-4] In Medical Physics and in this work, Monte Carlo techniques are employed for accurate patient dose calculation. Particles are transported through simulated linear accelerator head geometry and then into a phantom or patient geometry. Photon interactions and their outcomes follow experimentally determined probability distributions. With each step the particle takes, these distributions are sampled by random number generators in the Monte Carlo code, determining whether an interaction will take place, and the type of interaction. Other particles, electrons for instance, can be created in this process, and are also transported in a similar way. An example of this process can be seen in Figure 2.1. The entire series of events associated with a single particle and its progeny is called a history. Histories end when all the particles involved either
exit the geometry of interest or fall below a cut off energy. Along the way, stochastic quantities such as dose can be scored in regions of interest. By simulating large amounts of histories, the Monte Carlo technique will converge towards a solution, such a dose distribution, for example.

Figure 2.1: An example of Monte Carlo method within a patient geometry. Photons (yellow) enter the patient, and have interactions within the patient, producing secondary particles (electrons in light blue, positrons in red), depositing dose. This image was based on an image presented in BrainLAB’s white paper concerning the iPlan Monte Carlo code. [5]

In general, there are two categories of Monte Carlo simulations: analog and condensed history simulations. In analog simulations, all particle interactions are modeled explicitly. In condensed history simulations, given that the majority of electron interactions lead to very small changes in energy and direction, large numbers of these interactions are grouped into “steps.” With each step, the particle loses energy and can change direction, but these changes are not modeled explicitly. In radiation transport calculations, analog simulations are not practical considering the large number of interactions that must be simulated. The Monte Carlo codes in this work are of the condensed history variety, where interactions are grouped into “hard” and “soft” collisions, where only the “hard” collisions are simulated explicitly. Hard collisions result in large changes of energy and direction, as opposed to soft collisions where these changes are minimal. [6]

The uncertainty associated with stochastic quantities such as dose depends on the number of histories simulated. For a number of histories, \( N \), the uncertainty associated with a stochastic quantity is proportional to \( \frac{1}{\sqrt{N}} \). [7] The uncertainty of a simulation is also affected by the simulation geometry. In the case of simulating leakage through MLC leaves, for example, large numbers of histories have to be run simply to get a reasonable amount of particles that have made it through the MLC to score dose in a phantom. Such a simulation might require running upwards of hundreds of millions of histories.
The calculation of such enormous number of histories can require a lot of computing power. In general, the computation time is related to the uncertainty in the same way as histories, assuming the calculation time for each history is comparable. Given limited computed power, one simply cannot run infinite number of histories to achieve reasonable uncertainty. Introducing approximations or modifications to the Monte Carlo code can make calculations more efficient. These techniques are called “variance reduction techniques,” and are frequently employed in Monte Carlo methods to achieve low uncertainties within a reasonable time frame. An example would be bremsstrahlung splitting, where multiple photons, instead of just one, can be created from a bremsstrahlung event. Variance reduction techniques, however, must be used with care, as the approximations they introduce may not always be valid for a given simulation.

In general, calculating dose with Monte Carlo methods is useful because it can calculate dose in complex geometry where analytical methods may be difficult or impossible. Many traditional analytical dose calculation algorithms may predict dose accurately in water, but can perform poorly when heterogeneities are introduced. Considering that Monte Carlo methods are based on first principles, they can also help predict phenomena where measurements cannot be made. The introduction of Monte Carlo techniques into the treatment planning enables more accurate dose prediction in heterogeneous regions such as in the case of lung stereotactic body radiotherapy.

2.1.1  EGSnrc, BEAMnrc and DOSxyznrc

Developed by the National Research Council of Canada, EGSnrc is a Monte Carlo code developed model particle transport through materials. [4] BEAMnrc and DOSxyznrc are codes built on the EGSnrc simulation code, are used to model linear accelerator geometry and calculate dose in patient or phantom geometries, respectively. Simply put, EGSnrc contains the “physics,” in that it takes care of all the particle transport and interactions; BEAMnrc models a realistic beam, and this beam is then incident on either a phantom or a patient, where DOSxyznrc calculates the dose distribution. BEAMnrc and DOSxyznrc are run using input files (EGSINP), which contain all necessary information for the simulation. In these codes, all material information is stored in a file called a PEGS file. The PEGS file will contain all the radiological properties necessary for particle transport, such as interaction cross-sections, stopping power ratios, etc. All these codes can employ variance reductions techniques, as described in the previous section.

Linear accelerators are modeled in BEAMnrc with “component modules,” or CMs, which are independent geometric building blocks. For example, the jaws used to collimate the beam
can be modeled by the CM: JAWS, which consists of four rectangular blocks with angled sides. There is a whole selection of pre-coded CMs within BEAMnrc that can be used to model different portions of the accelerator. However, as the reader will see in the section 3.4, users can also code their own CMs to model any geometry.

Within the code for each CM, there are several macros that define the geometry. Specifically, WHEREAMI, HOWFAR and HOWNEAR are some of the most important. WHEREAMI determines which region within the CM the particle is in. HOWFAR determines the distance to the nearest boundary along the particle’s trajectory. And HOWNEAR determines the minimum distance in any direction to the nearest boundary. These macros are used to determine the step size of the particles as they approach boundaries, and therefore, define the geometry of the component being modeled. To model an entire accelerator, several CMs are placed in succession to model all the individual parts. Figure 2.3 depicts exactly this, pointing to all the different components all modeled by different CMs.

Figure 2.2: A linear accelerator head, as simulated by BEAMnrc. This is not the model used to simulate the Novalis Tx radiosurgery machine. The vertical axis here is the X axis, and the horizontal axis is the Z axis, with the dimensions being in centimeters. This image was produced with BEAMnrc, by previewing the accelerator.

Another important aspect of BEAMnrc is the source model, i.e. the source of the electrons or photons that will eventually exit the accelerator. There is a wide variety of source
models that can be used. The one used mainly in this work is source 19, an elliptical source with a Gaussian distribution. This source is used to simulate the electron beam hitting the target, which will produce photons via bremsstrahlung. Just as CMs must be tailored to model the components of the accelerator, the source model must also be adjusted to properly model the beam produced by an accelerator.

The output of BEAMnrc can be a phase-space file, which is a file that stores all the particles at a plane with their directions, energies, charge, etc. Phase-space files are typically used as sources for DOSxyznrc, but they can also be used as sources in BEAMnrc. For example, given that simulations often differ only below the jaw level in BEAMnrc, a phase-space can be scored above the jaws, and then used as a source incident on the jaws, so that the simulation above the jaws only has to be done once. BEAMnrc can also be linked to DOSxyznrc, forming a “shared library” simulation, where no phase-space file is produced and dose is scored directly in DOSxyznrc.

Anything that can be described by Cartesian geometry can be modeled in DOSxyznrc. Other codes can be used to calculate dose in other types of geometries, such as in DOSrznc which calculates dose in cylindrical geometries. However, in this work, virtually all dose calculations are done with DOSxyznrc. A source is needed to deposit these doses and sources models can be employed in the same way as in BEAMnrc described above. The geometries in DOSxyznrc are divided into “voxels” which are essentially the three-dimensional version of pixels, so whenever scoring dose is mentioned, it is always scored in a voxel. DOSxyznrc allows the user to “voxelize,” or determine the size and placement of the voxels, however they wish. In certain areas, for example, one might need more spatial resolution, so voxel sizes could be made smaller in that area. In general, calculation efficiency decreases with decreasing voxel size, given that a particle will have to cross over more boundaries, increasing calculation time. Therefore, in simulated areas where the dose need not be examined, large voxels can be used to speed up particle transport. The output of a DOSxyznrc simulation is a 3DDOSE file, which is a file that contains all the position and size of the voxels, the dose scored in each voxel, and the statistical uncertainty associated with it.

2.1.2 iPlan treatment planning system

The iPlan treatment planning system is comprised of several programs designed to interpret CT images of patients, delineate structures, design treatments, and predict radiation doses. The program RT DOSE imports CT data sets, and converts them to electron densities using a Hounsfield unit to electron density conversion curve. Designed for radio-surgical precision, RT IMAGE can impose a coordinate system on the CT image with the help of fiducial markers.
Structures, such as organs at risk or tumor volumes, can then be contoured to guide the planning process. Following this, a treatment can be designed, using RT DOSE, by selecting a target, dose tolerances to organs at risk, and the prescription dose. Beam arrangements can be created and collimating devices can be adjusted accordingly. With the plan in place, the dose can be calculated with the either the pencil beam or Monte Carlo dose calculation algorithms. Usually, the Monte Carlo algorithm is used for plan validation after initial calculation with the pencil beam algorithm.

The interest of this work is iPlan’s Monte Carlo calculations of dose, as it is done differently than in the EGSnrc code system. Instead of modeling the entire gantry head of the linear accelerator, as is done in BEAMnrc, iPlan’s Monte Carlo algorithm utilizes a virtual energy fluence model, or VEFM. [5] Given that all the linear accelerator components above the jaws do not change in a given mode, it is assumed that the phase space of particles at that level is independent of field configuration. Therefore, the VEFM model consists of several photon sources that match the primary and scattered radiation that a full modeling would produce. These virtual photon sources are incident on a model of the collimating system, consisting of jaws, MLCs or stereotactic cones. Within these collimating devices, particle transport is simulated as described in section 2.1. With the VEFM, the iPlan Monte Carlo algorithm performs significantly faster any EGSnrc calculation, given that it does not have to repeat the detailed particle transport upstream from the jaws.

IPlan’s dose calculation engine is based on XVMC, originally developed by Kawrakow and Fippel. [8] Instead of using material information contained in PEGS files, as done in EGSnrc, all the required material information such as cross-sections and stopping power ratios, are calculated from electron densities, determined by the CT number to electron density conversion curve. Electron densities \(n_e\) can be converted to mass densities \(\rho\), or vice versa, using the following formula: [5]

\[
n_e = n_e^W \frac{\rho}{\rho^W} f_c(\rho) \tag{2.1}
\]

where \(n_e^W\) and \(\rho^W\) are the electron and mass densities of water, respectively. The form function, \(f_c(\rho)\) is defined as: [5]

\[
f_c(\rho) \approx \begin{cases} 
0.99 + 0.01 \frac{\rho}{\rho^W}, & \text{for } \rho \leq \rho^W \\
0.85 + 0.15 \frac{\rho^W}{\rho}, & \text{for } \rho \geq \rho^W
\end{cases} \tag{2.2}
\]

Cross-sectional information is then calculated in the following manner, using Compton interactions as an example:
\[ \mu_c = \frac{\rho}{\rho_w} f_c(\rho) \mu^w_c \]  

(2.3)

Referring to Kawrakow’s work concerning VMC, all quantities necessary to simulate interactions are calculated in a similar manner, but the form function \( f(\rho) \) may be different depending on what is being calculated.[8]

There are four options available for Monte Carlo dose calculations within iPlan: spatial resolution, desired mean variance, dose result type, and MLC modeling. The spatial resolution simply defines the resolution of the Monte Carlo calculation, much like how DOSxyznr divides a phantom into voxels. The voxels sizes are approximate, as they are formed by combining the CT voxels in all three spatial dimensions. As mentioned in section 2.1, the number of histories determines the statistical uncertainty of a given Monte Carlo simulation. Simply put, iPlan’s Monte Carlo algorithm will run sufficient histories in order to achieve the desired mean variance. The dose result type can be set to either “dose-to-medium” or “dose-to-water.” In both cases, the Monte Carlo calculation is essentially the same, but with “dose-to-water,” there is one important difference: Bragg-Gray cavity theory is assumed, applying stopping power ratios to calculate what the dose would be to a small mass of water in medium. Bragg-Gray cavity theory states that the dose to a small mass of water in medium can be calculated using the following formula:

\[ D_W = D_{med} \langle \frac{S}{\rho} \rangle^W_{med} \]  

(2.4)

where \( D_W \) is the dose-to-water, \( D_{med} \) is the dose-to-medium, and \( \langle \frac{S}{\rho} \rangle^W_{med} \) is the unrestricted stopping power ratio of water to medium. In the case of iPlan, the stopping power ratios are collisional stopping power ratios.[5] The discussion of the difference between dose-to-water and dose-to-medium will continue in the section 2.1.5. Finally, the MLC modeling option for Monte Carlo can either be “accuracy optimized” or “speed optimized”. If the accuracy optimized option is used, then the full tongue-and-groove design of the MLC’s leaves is modeled. If the speed optimized option is used, a simpler model of the MLC is used to improve the speed of the calculation.

2.1.3 MMCTP

The McGill Monte Carlo treatment planning system, or MMCTP, [9] is designed as a port between clinical treatment planning systems and the EGSnrc Monte Carlo code systems. It functions by importing CT images and treatment plans from other systems, interpreting them in order to modify a template input file for BEAMnrc and DOSxyznr , which perform the Monte Carlo calculations. One can modify most of the simulation parameters within MMCTP as one...
would do with the BEAMnrc and DOSxyznrc GUIs. It also has the ability to interpret contours and structure sets, allowing for calculation of dose-volume histograms for targets and organs at risk. MMCTP has also been coded to submit calculations on a remote cluster. To perform calculations in DOSxyznrc, MMCTP creates an EGSPHANT file from CT image data. When the simulations are finished, MMCTP can interpret the 3DDOSE files produced by DOSxyznrc and superimpose the doses unto CT images, just like other treatment planning systems.

In this work, MMCTP is used to perform dose calculation comparisons with the iPlan’s Monte Carlo code and the EGSnrc Monte Carlo code systems.

2.1.4 Dose-to-water vs. dose-to-medium

As mentioned earlier, the iPlan Monte Carlo algorithm can either calculate dose-to-medium or dose-to-water. Historically, most patient dose calculations assumed that the entire patient was composed of water. Much initial research was based on this assumption, forming the body of knowledge that physicians use to treat patients. For example, a clinical trial could conclude that a particular dose was most effective for local tumour control. Regardless of whether or not this was the “true” dose, physicians could rely on the results it produced. With the advent of Monte Carlo dose calculation, doses to specific media could be calculated explicitly, throwing into confusion the meaning of dose. Although a dose-to-medium calculation is sound, much clinical knowledge has been based on dose-to-water. The AAPM task group report number 105 recognizes the problem, stating arguments on both sides. [6] In brief, the arguments in favor of using dose-to-water are that clinical experience is based on calculations in water, all measurement protocols are based in water, and cells embedded in media are more like water than the media. Arguments in favor of using dose-to-medium are that doses to media are “true” doses computed by Monte Carlo algorithms, much clinical practice will not change as the difference will be small for tissue equivalent materials, and there are uncertainties involved in converting dose-to-medium to dose-to-water.

Several methods were subsequently developed to convert dose-to-medium to dose-to-water. The methods of converting dose-to-medium to dose-to-water mentioned here are all based on the assumptions of Bragg-Gray cavity theory. In truth, the conditions of Bragg-Gray cavity theory may not apply in all patient geometries, making the uncertainty of these conversion methods unknown. Bragg-Gray cavity theory is only valid in regions of transient or full charged particle equilibrium. Bone presents the greatest problem. For example, the application of stopping power ratios would only be valid if the bone was large enough so that transient charged particle could be re-established.
Siebers et al. [10] suggests a method to convert dose-to-medium to dose-to-water by binning patient data into air, lung, tissue, soft bone or cortical bone. These materials have well studied stopping power ratios with respect to water that are relatively constant with depth, energy and at interfaces (barring a few exceptions). Figure 4 in the Siebers paper illustrates the stopping power ratios for all the materials mentioned as a function of density. [10] Under the assumption of Bragg-Gray cavity theory, to calculate dose to water within a material, the stopping power ratios specific to that material can simply be applied. However, as Kawrakow and Fippel pointed out in their comment of the Siebers paper [11], the discontinuities associated with the Siebers method can introduce very large uncertainties. For example, at a density of 1.4 g/cm³, there would be a 6% difference in the calculation of dose-to-water depending if the material was soft or cortical bone. From CT data, only mass or electron densities can be determined, not material information, making the Siebers method ambiguous at certain densities. This explains the use of the form function to calculate stopping power ratios, as described in section 2.1.2, as it does not suffer these discontinuities. The reader is referred to Kawrakow’s VMC paper to determine how stopping power ratios are calculated in this case. [8] The iPlan Monte Carlo dose calculation algorithm employs Kawrakaw’s method.

2.2 Source Characterization

As mentioned in section 2.1.1 for BEAMnrc, the parameters associated with source models need to be adjusted to accurately simulate a beam. The source used in this work is source 19, in BEAMnrc, which has three main parameters: beam energy, beam size, and angular spread. Given that these parameters define the source, they affect every dose distribution. However, certain tests are more sensitive to one parameter than the others. Therefore, those particular tests are used to determine that single parameter. Procedures involving multiple tests that aid the determination the source parameters can be called source characterization procedures, where source characterization means the adjustment of free parameters so as to ultimately match profiles and percent depth dose curves in water. Two source characterization procedures are used in this work: one from the work of Sheik-Bagheri and Rogers [12], and a newer procedure from Almberg et al. [13]

2.2.1 The Sheik-Bagheri and Rogers Method

The landmark paper about the sensitivity of Monte Carlo simulations to source parameters by Sheik-Bagheri and Rogers was the logical place to start source characterization. [12] Aside from obtaining accurate geometric information of each element of the treatment head, they suggest a simple procedure for finding the source’s energy and size: using a 10 cm x
10 cm$^2$ percent depth dose curves measured in water and large field in-air off-axis ratios. The percent depth dose curve is used to find the source’s energy, as that is what the test is primarily sensitive to. Having found the energy of the beam, one can use in-air off-axis ratios to determine the source size. The in-air off-axis ratios are sensitive to both source size and energy, so it is suggested to modify the energy if a sufficient fit cannot be found by modifying the source size. Sheik-Bagheri and Rogers recommend rechecking the percent depth dose curve if the energy had to be changed to ensure the match is satisfactory. The authors ignore the angular spread of the electron beam since they claim that “credible divergences up to 0.5 degrees show no effect.”[12]

2.2.2 Almberg’s method

The method for source characterization presented by Almberg et al. in their recent paper, proposes three separate tests for adjusting the three free parameters of source 19 in BEAMnrc. A 5 x 5 cm$^2$ percent depth dose curve is used to determine the beam energy. A linear function is fitted to the local dose difference with measurements as a function of depth. The beam energy corresponding to a slope closest to zero of the fitted linear function is the energy to be used. An EBT-2 film measurement of a 5 x 5 cm$^2$ penumbra is then used to adjust the source size, described in Almberg’s paper by sigma, σ. Since the penumbra in measured in both the in and cross-plane directions, both dimensions of the source can be adjusted separately. In this work, the full-width at half-max, or FWHM, is used, where $FWHM = 2\sqrt{2\ln2}\sigma \approx 2.35\sigma$. Finally, the angular spread of the beam is adjusted to match large field profiles at depths of 1.5 and 10 cm. The advantage of Almberg’s method was that there was no changing of previously determined parameters, angular spread is introduced as an additional parameter and each dose distribution the procedure utilized was easy to measure.[13]

2.3 Multi-leaf collimators

A multi-leaf collimator, or MLC, is a beam collimating device located close to the jaws, which helps to further shape the beam into more conformal shapes, eliminating the need for custom shaped lead blocks. An MLC is composed of two banks of many interlocking leaves made of tungsten alloys that can be moved independently from one another by a driving screw embedded within the leaf. MLC beam shaping can be static or dynamic, allowing for intensity modulated treatments, or IMRT treatments (intensity modulated radiotherapy).

The first patient treated with a multi leaf collimator dated back to 1959. [14] Various commercial vendors have since designed different versions of MLCs: having different placement within the gantry head, different carriage design and different leaf design. Each vendor places their MLC in a different position relative to the jaws. As for carriage and leaf design, the Siemens
MLC®, for instance, replaces the lower jaw, and is mounted on an arching trajectory; allowing for the flat leaf ends of the Siemens MLC to always be parallel to the beam. [15] Other MLCs travel linearly, and therefore, employ rounded leaf ends. The leaf designs are different for each MLC, as companies solve the problem of leakage through the leaves in different ways. The following figure demonstrates different interlocking leaf designs.

![Interlocking leaf designs](image)

Figure 2.3: Interlocking leaf designs, from left to right: Elekta, Siemens and Varian.

The designs shown, however, are not the only ones used today. The HD120 MLC investigated in this work, for example, employs a more complicated interlocking design. MLCs can also contain multiple types of leaves within a single leaf bank, depending how the leaves interlock and the resolution of the fields the MLC is designed to shape.

2.3.1 The DYNVMLC model

There are many component modules available for the simulation of MLCs, within the BEAMnrc system, but given the similarity of the design of the leaves of the HD120 MLC with those of Varian’s Millennium MLC, the DYNVMLC component module was modified to model the HD120 MLC. The DYNVMLC component module was originally designed to model the Varian Millennium MLC.[16] As Figure 2.6 demonstrates, the DYNVMLC component module can model 3 types of leaves: full, target and isocenter leaves, the focused sides of the leaves, and also rounded or focused leaf ends. Complex and dynamic fields can also be modeled with this component module. The description of how the DYNVMLC component module was modified to model the HD120 MLC can be seen in section 3.4.
For many of the validation tests, the MLC shape is planned within a treatment planning system, which defines the MLC leaf positions they project at isocenter. In contrast, the DYNVMLC component module’s input, and consequently, the HDMLC component module’s input, as well, is such that the leaf positions are defined at the level of the MLC leaves. Given the curved nature of the MLC leaf ends, a simple scaling ratio cannot be applied to convert MLC positions at the MLC level to their projected field at isocenter. The following formula is used, therefore, to calculate the positions of the simulated leaves, $X_c$, knowing what the desired field is at isocenter:

$$X_c = \frac{X}{SAD} (SCD + R \sin \theta) - R (1 - \cos \theta)$$

where $SAD$ is the distance from the target to where the field is defined (100 cm), $SCD$ is the source to collimator distance (51.01 cm), $X$ is the field size at the $SAD$ distance, $R$ is the radius of the leaf end and $\theta$ is the angle the desired field size makes with the central axis. So whenever the MLC defines the field, the planned projections at isocenter are converted to leaf positions at the level of the MLC for simulation.
2.4 MATLAB®

MATLAB®, a programming environment produced by Mathworks [19], is used in this work to perform the majority of data analysis and figure generation. MATLAB® is an extremely versatile software tool that can perform a wide variety of functions such as algorithm development, data analysis, visualization, and numerical computation. In this work, many scripts have been written in MATLAB® for film analysis, extracting dose planes from 3DDOSE files produced by DOSXYZnrc, and quantifying agreement between simulations and measurements. All graphs shown in this work were produced by MATLAB® with the exception of the ray trace plots shown in section 4.3.1.

2.5 Experimental analysis metrics

The quantities used in the analysis of the results are local dose difference, root-mean-square-difference, or RMSD, and the gamma criterion. Local dose difference and RMSD are used to quantify differences between simulations and measurements. Local dose difference is defined in the following equation:

$$D_{if} = \frac{(MC - \text{Measurement})}{\text{Measurement}}$$

(2.6)
where $MC$ is the Monte Carlo simulated percent depth dose value. Local dose difference is used to determine trends in differences between simulations and measurements. The root mean square difference calculation is as follows:

$$RMSD = \sqrt{\frac{\sum(MC_i - M_i)^2}{n}}$$

(2.7)

where $MC$ is the Monte Carlo data, $M$ is the measurement data, and $n$ is the total number of data points. The RMSD is used as an objective function to aid in determining free parameters of a simulation, as multiple simulations are compared to a single measurement.

When comparing simulations to film, there are two sources of differences between distributions: dosimetric and positioning differences. For example, comparing corresponding points of both distributions, the difference between them may be real, or it may be a result of misalignment. Distance-to-agreement and dose-difference, therefore, are two quantities that are used to compare 2D distributions. Distance-to-agreement is defined as the nearest distance from a point on the measured dose distribution to the point of the same amount of dose on the simulated dose distribution. Dose-difference is the simple point-by-point percent dose difference and it assumes that the simulated and measured dose distributions are in perfect alignment. [20]

In Gamma analysis, distance-to-agreement and dose-difference are combined into a single variable:

$$\Gamma = \sqrt{\left(\frac{DTA}{C_{DTA}}\right)^2 + \left(\frac{DD}{C_{DD}}\right)^2}$$

(2.8)

where $DTA$ and $C_{DTA}$ are the distance-to-agreement and distance-to-agreement criterion, respectively, and $DD$ and $C_{DD}$ are the dose-difference and dose-difference criterion, respectively. Should the $\Gamma$ value for a point be less than 1, then that point is said to have passed the gamma analysis. Even if one parameter is close to failing its criterion, a point may still pass if the other parameter is small enough relative to its criterion. [20]

2.6 References


Chapter 3

MATERIALS AND METHODS

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3.1 Measuring devices

3.1.1 EBT-2 film

For the measurement of planar 2D dose distributions and for point dose measurements in slab phantoms, EBT-2 GAFCHROMIC™ (International Specialty Products, Wayne, NJ) film was used. EBT-2 film is specifically designed for radiation dosimetry: a self-developing film that darkens with radiation dose. It is a relatively cheap and versatile dose measurement tool: formulated to measure doses up to 800 cGy and to be energy independent above 50 keV into the MeV range. Specifically, the film itself consists of many layers: a polyester overlaminate, an adhesive layer, a topcoat, an active layer, and a polyester substrate. In particular, the active layer reacts to form a blue polymer when exposed to radiation, which appears green due to the yellow marker dye of the film, as seen in Figure 2.1. The film’s structure can be seen in Figure 3.2. Following irradiation, the films are scanned on an Epson Expression 1680 flatbed document scanner in RGB mode with the commercial software Film QA.[1] Only the red channel image is used, as suggested by Devic et al, 2005, [2] and optical densities are converted to doses using a calibration curve.
3.1.1.1 Orientation and Noise

The orientation of EBT-2 film as it is scanned on a flat-bed document scanner can affect the dose extracted from the film measurement. Considering the older version of the film, EBT, optical density can vary as much as 15% when rotated 360 degrees on the scanner. [2] There is a notch on one corner of each EBT-2 film, which aids the user to keep track of the orientation. Great care was employed to ensure that the calibration and measurement films were all scanned in the same orientation.

Film measurements can be noisy. This is partially due to scanner noise and partially due to minor imperfections in the film; both are inevitable sources of noise that must be dealt with. To remove scanner noise, one can simply apply an adaptive Wiener filter to the unconverted film image. [2] Wiener filtering is designed to remove noise from a signal (by estimating the local noise power spectrum for each pixel) while preserving the desired variations in the films optical density. The MATLAB command is \texttt{wiener2} (a built-in function), where the syntax is as follows: \texttt{filtered image = wiener2 (unfiltered image, (region))}, where “region” is the number of pixels one wants to use to estimate the local noise power spectrum. For calibration films, 7 x 7 pixel regions are used, so the syntax would be \texttt{image2 = wiener2 (image1, (7 7))}. For measurements, a smaller region of interest was chosen to ensure that systematic variations in the optical density are not
affected: a 3 x 3 pixel region was used. It must be repeated that the wiener filter is only applied to the raw images, NOT after this images are converted to dose.

As for dealing with film noise, there is a different procedure for calibration and measurement films. For calibration films, film noise can be removed by using multiple regions of interest to determine pixel values. With the help of MATLAB scripts, one can obtain the mean and standard deviation of the pixel values within 10 pixels in the x and y direction. It is recommended to do this procedure 5 times to obtain reasonable uncertainty.[2] Since the user selects a different point every time, different regions of interest are chosen and averaged, so random variations in the film will be averaged out as a consequence. For profiles extracted from measurement films, film noise is removed by averaging a number of pixels perpendicular to the direction of the profile. Taking a profile that is one pixel thick will be extremely noisy. By averaging over a few pixels, not only is film noise reduced, but scanner noise as well.

3.1.1.2 Calibration Curve

The calibration curve was generated by first cutting and labeling one film sheet into 8 rectangular pieces. The rectangular nature of the pieces and the dose label ensures that proper orientation is maintained in all senses. The labeling also gives the user a clear place to handle the film, as optical density will not be measured near the label. Before exposing the films, they were all scanned together, as if they were to reconstruct the original large film piece. The orientation of the unexposed calibration films on the scanner was noted, so that the exposed calibration and measurement films could be scanned in the same orientation. All scanned films were saved as 48 bit TIFF files. These films were then irradiated beyond depth of maximum dose in solid water: at 5 cm depth, SAD setup with 10 cm of solid water for backscatter. Each calibration film was irradiated to a different dose, in the range of 0 – 400 cGy. Tabulated output factors and TMR tables were used to calculate the doses delivered to the film plane. The MUs were simply adjusted to deliver the appropriate dose, accounting for the machine’s output. Following irradiation, twenty four hours passed before the films were scanned.

Having obtained the two TIFF files, the unexposed and exposed calibration films, a calibration curve is generated following the procedure set out by Devic et al. 2005, with the help of a few MATLAB scripts and excel. Specifically, as described earlier, a 7 x 7 adaptive wiener filter was applied to both TIFF files and five regions of interest were chosen on each film, both unexposed and exposed, to obtain mean pixel values and standard deviations. After converting the pixel values to optical density according to the Devic’s method[2] and inputting the known dose values, a function of following form was fitted to the data:
\[ D_{fit} = b \times \text{netOD} + c \times \text{netOD}^n \]  

(3.1)

Where \( D_{fit} \) is the delivered dose and \( \text{netOD} \) is the net optical density of the film. This function can now be used to convert the optical density changes into dose values. According to the uncertainty analysis of Devic [2], the uncertainty on the EBT-2 film measurements in this work is under 2\% for doses above 50 cGy. Below 50 cGy, the uncertainty rises sharply, and after 400 cGy, the calibration curve cannot be trusted as no calibration films were irradiated above that dose.

Another calibration curve was generated using the Film QA software in a similar manner. The pixel values were extracted from the irradiated films by creating regions of interest in a similar fashion as described above. Only one region of interest was chosen for each film, however, and the TIFF images were not filtered as they were using MATLAB. The dose vs. pixel value data was fitted to a six degree polynomial instead of equation 2.2.

### 3.1.1.3 Film processing

Films were analyzed using two methods: using the commercial software Film QA, and using customized MATLAB scripts. Film QA is a very versatile piece of software that allows the user to make comparisons of films with ease, allowing on-the-fly adjustments of image registration and normalization. The software allows for the easy extraction of profiles, as well as gamma analysis. However, it must be noted that Film QA seems to have limitations regarding its spatial resolution. As the reader will see in the results section, the demands on spatial resolution are high when measuring interleaf leakage, especially in the case of the quarter leaf bank. As stated earlier, all the inner workings of Film QA are unknown, but it appears that Film QA has a coarser sampling than a customized MATLAB method as can be seen in Figure 2.3. The MATLAB method used to analyze the film first applies an adaptive wiener filter as described in section 3.1.1.1, then makes use the improfile_integrated function[4]. This function allows the user to make a profile of a customizable pixel thickness between any two points, integrating in the direction perpendicular to the profile. Essentially, the MATLAB script deals with noise by wiener filtering and averaging over pixels perpendicular to the profile direction. In contrast, Film QA may deal with noise by reducing its sampling frequency, but this is not known for sure. The bottom line is that the MATLAB method was used when high spatial resolution was required, and Film QA was used in every other case.
Figure 3.3: Comparison of film processing techniques for an EBT-2 film measurement of transmission through a closed MLC. Transmission values are calculated by dividing the film dose by the dose that would be delivered using an open field. Note the difference in sampling frequency between the two techniques.

Another issue that is addressed in film analysis is the alignment of the film. While the Monte Carlo dose distribution is always centered at zero, the film measurements need to be shifted to the appropriate location. Film QA, with its easy to use registration tools, allows the user to center their film, at least to the precision of the room lasers given their marks on the film. One can translate and rotate the film manually until the laser markings match the superimposed axes. In Film QA, the profiles can be visualized on the fly, allowing for an estimate of positional uncertainty, as the film can be moved further after registration to obtain the best agreement. But while Film QA has some significant advantages, the MATLAB method is used for certain measurements because of the preservation of spatial resolution as mentioned above. This means that the problem of film alignment still has to be dealt with. The HDMLC can be imperfectly centered, in a given installation, resulting in misaligned MLC-defined profiles. Due to the difficulties of aligning the film, the film cannot detect misalignments below a millimeter or so. So for the purposes of this work, it is assumed that a film’s profile is produced from a perfectly centered HDMLC. A simple MATLAB script was written, therefore, to shift a film’s profile until it best fits the Monte Carlo profile. The script accomplishes this by interpolating the
Monte Carlo data to the positions of the film points, and then calculating the difference between
the film data and the interpolated Monte Carlo data. This difference is then squared and
summed across the entire distribution. This process is repeated as the profile is shifted,
recording the sum of the square differences each time. Then the script simply reports the shift
that corresponds to the minimum sum of the differences. This shift, therefore, corresponds to
the best alignment of the film.

3.1.2 Ion chambers

To measure profiles and output factors, both IC-10 (CNMC, Nashville, USA) and
microLion (PTW –Freiburg, Freiburg, Germany) ionization chambers were utilized. Ion chambers
used to measure dose typically consist of a cavity that lies between two electrodes. A potential
difference is setup between the two electrodes so that charged particles created by ionizations
within the cavity can be collected by a collecting electrode due to the electric field in the cavity.
The signal produced by the collecting electrode is measured with an electrometer, which a
device used to measure small currents. Every chamber has a host of correction factors th
include effective point of measurements shifts, temperature and pressure corrections, ion
recombination corrections, and polarity corrections, to name a few. However, barring a few
exceptions, ion chambers can be considering a simple measurement tool as they respond linearly
with dose, as opposed to film, where a calibration curve is necessary. In general, most
measurements done in this work are relative measurements, so signals can be normalized to
each other as if they were dose, assuming the same correction factors apply in the cases being
considered. For a complete understanding of absolute dosimetry, the reader is encouraged to
read the AAPM’s TG-51 protocol. [5]

The IC-10 chamber is an air-filled cylindrical ion chamber with a collecting volume of
0.13 cm³. It was used to measure percent depth dose curves, dose profiles and the Novalis Tx’s
output. Following the TG-51 protocol, the position of the chamber was adjusted to account for
the effective point of measurement shift of 0.6 \( r_{cav} \) where \( r_{cav} \) is the inner radius of the air cavity
of the chamber; 6 mm for the IC-10 chamber. Likewise, the percent depth dose curves were also
shifted upstream by 0.6 \( r_{cav} \). With the exception of measuring output factors, all other
measurements with the IC-10 chamber were relative measurements, so no further corrections
were applied. Before making film measurements, the IC-10 chamber is used in conjunction with
a solid water phantom to ensure that the machine’s output has not deviated significantly. The
TG-51 protocol was followed, placing the chamber so that its effective point of measurement is
at a depth of 10 cm. Using a 10 x 10 cm² jaw defined field, 500 MU was delivered to calculate the
output in cGy/100 MU. The chamber’s average reading per 100 MU: $\frac{R}{100 \text{ MU}}$ is then corrected for temperature and pressure using the following equation:

$$M_{T,P} = \frac{R}{100 \text{ MU}} \times \left( \frac{760}{P} \times \frac{273 + T}{295} \right)$$  \hspace{1cm} (3.2)

Where $M_{T,P}$ is the corrected reading, $P$ is the pressure in mm Hg, and $T$ is the temperature in degree Celsius. Next, the following equation calculates the output in cGy/100 MU at depth of dose maximum (1.5 cm):

$$\frac{D_{\text{water}}}{100 \text{ MU}} = \frac{100 \times M_{T,P} \times (N_{D,W})_{SW}}{PDD_{10 \text{ cm}}$$  \hspace{1cm} (3.3)

Where $(N_{D,W})_{SW}$ is the absorbed dose to water conversion factor and $PDD_{10 \text{ cm}}$ is the percent depth dose value at 10 cm. Both quantities were determined previously by clinical physicists here at the Montreal General Hospital. The $(N_{D,W})_{SW}$ factor contains the ion recombination correction factor: $P_{ion}$, as defined in the TG-51 protocol. [5] Please refer to section 2.6.1, for the details on the IC-10’s use in measuring output factors for different field sizes.

The microLion chamber is a miniature parallel plate ionization chamber with a collecting volume of 0.0017 cm$^3$ filled with isooctane, instead of air. It was used exclusively for the measurement of output factors as a function of field size, due to its small collecting volume. Since the work focuses on the SRS mode of the Novalis Tx, small fields (below 4 x 4 cm$^2$) are of particular interest. A known problem for small fields is that many detectors have dimensions comparable to the field size, so the chamber may only be partially irradiated, resulting in an incorrect measurement. [6] Therefore, the microLion chamber, with its small collective volume, was the ideal choice to measure the output of small fields. Whenever the microLion chamber is used, the integrated electronics of the Wellhöfer water tank (CNMC, Nashville, USA) are only used for positioning, and not for measurement. To avoid any interference from the positioning electronics of the water tank, once the chamber has been positioned, it is disconnected from the Wellhöfer electrometer and connected to a Keithly 6517A electrometer with a polarizing voltage of 800V.

Since the microLion’s cavity is filled with fluid instead of air, typical ion recombination correction factor do not apply. A procedure developed to correct for ion recombination as a function of the microLion’s signal was used. [7] First, the collection efficiency $F(Q_M)$ is calculated with the following formula:
\[ F(Q_M) = 1 - a_{\text{xu}} Q_M = \frac{Q_M}{Q_{\text{sat}}} \]  

(3.4)

where \( a_{\text{xu}} \) is an experimentally determined constant which depends on the measurement conditions, \( 1.34 \times 10^{-3} \), in this case, \( Q_M \) is the microLion chamber signal per 100 MU, and \( Q_{\text{sat}} \) would be the ideal signal with no loss of signal due to ion recombination. Therefore, the corrected signal can be calculated using:

\[ M_{\text{microLion}}^c = \frac{M_{\text{microLion}}}{f(Q_m)} \]

(3.5)

So whenever a measurement was made with the microLion chamber, first the signal was scaled according to the number of MUs delivered to find the signal per 100 MU, then the collection efficiency was calculated, and finally, the total signal was corrected using the collection efficiency.

### 3.2 Monte Carlo settings

The BEAMnrc and DOSxyznrc Monte Carlo user codes are used for the linear accelerator head modeling and dose calculation, respectively.[8, 9] These user codes are based on the EGSnrc code system for modeling electron and photon transport.[10] For an introduction to Monte Carlo simulations, please refer back to section 2.1. While some Monte Carlo parameters were modified for particular experiments, many parameters are common to all simulations done in the work. The phantom dimensions and voxel sizes, for example, will vary depending on the simulation. This section serves to expound the Monte Carlo parameters that are constant for all simulations. Unless otherwise specified, all Monte Carlo dose simulations were run using a shared library, which means that the BEAMnrc treatment head simulation is coupled directly to the phantom dose simulation of DOSxyznrc.

In BEAMnrc, the EGSnrc parameters are as follows:

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>max step size: ( 1 \times 10^{10} )</td>
<td>max fractional energy loss: 0.25</td>
</tr>
<tr>
<td>Xlmax=0.5 boundary crossing algorithm: EXACT</td>
<td></td>
</tr>
<tr>
<td>no skin depth for BCA</td>
<td>electron-step algorithm: PRESTA-II</td>
</tr>
<tr>
<td>spin effect: off</td>
<td>electron impact ionization: off</td>
</tr>
<tr>
<td>brems angular sampling: KM</td>
<td>brems cross sections: BH</td>
</tr>
<tr>
<td>bound Compton scattering: off</td>
<td>pair angular sampling: KM</td>
</tr>
<tr>
<td>pair cross sections:</td>
<td>photoelectron angular</td>
</tr>
<tr>
<td>Rayleigh scatter: off</td>
<td>atomic relaxations: on</td>
</tr>
<tr>
<td></td>
<td></td>
</tr>
</tbody>
</table>
BH | sampling: off
---|---
photon cross-sections: | no photon cross-sections output
| Table 1: EGSnrc parameters in BEAMnrc

The BEAM parameters used are as follows:

<table>
<thead>
<tr>
<th>directional bremsstrahlung splitting: varying splitting field radius</th>
<th>SSD 100 cm</th>
<th>brem splitting number 1000</th>
<th>Z of rejection plane: 85</th>
</tr>
</thead>
<tbody>
<tr>
<td>e-/e+ splitting in the flattening filter</td>
<td>e-/e+ splitting plane no. 12 (12.837 cm)</td>
<td>Z of Russian Roulette 12.67 cm</td>
<td>do not redistribute split e-/e+</td>
</tr>
<tr>
<td>augmented range rejection</td>
<td>brems cross-section enhancement: off</td>
<td>no photon splitting at CMs</td>
<td>incident particle: electrons</td>
</tr>
<tr>
<td>incident particle: electrons</td>
<td>global ECUT: 0.7 MeV</td>
<td>global PCUT: 0.01 MeV</td>
<td>electron range rejection: on with ECUTRR</td>
</tr>
<tr>
<td>ESAVE_GLOBAL: 2 MeV</td>
<td>photon forcing: off</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

| Table 2: BEAMnrc parameters

For the DOSxyznrc simulations, depending on the simulation, ECUT was either 0.521 MeV for simulations with small voxels, and 0.7 MeV for simulations with larger voxels, PCUT: 0.01 MeV, incident particles: all, source type: 9 (BEAM treatment head simulation), HOWFARLESS: on in the case of homogeneous water phantoms and off in all other cases, range rejection on with ESAVE: 2 MeV, and photon splitting number: 200. All EGSnrc parameters in the DOSxyznrc simulations are identical to those in the BEAMnrc simulations.

In general, all simulations were run with sufficient histories to achieve statistical uncertainty of under 0.5%. All simulations were performed on a BLADE cluster.

### 3.3 Source Characterization

Having modified a model of a Varian Clinac to simulate the Novalis Tx treatment head (Varian Medical Systems, Palo Alto, CA), the source parameters of source 19 (an elliptical beam with a Gaussian distribution in X and Y) had to be re-characterized due to the introduction of the
SRS flattening filter. For a description of source characterization please refer to section 2.2. A new flattening filter necessitates source characterization because the filter and the source parameters both affect the shape of the beam. Changing one, therefore, necessitates adjusting the other. The SRS flattening filter can be seen by removing some of the shielding surrounding the treatment head as shown in Figure 3.4.

Figure 3.4: Gantry head shielding must be removed (left) to expose the SRS flattening filter (right).

Having obtained the dimensions of the flattening filter from Varian in a non-disclosure agreement, the SRS flattening filter was incorporated into the BEAMnrc model.

3.3.1 The Sheik-Bagheri and Rogers Method

The procedure established by Sheik-Bagheri and Rogers [11] was to find the source's energy and size using a 10 cm x 10 cm$^2$ percent depth dose curves measured in water and large field in-air off-axis ratios, respectively. Please refer to section 3.3.2.1 for the description of how the percent depth doses were measured and simulated. Finding the beam energy using percent depth dose data is common to both source characterization procedures in this work, so it is only described once. Having found the energy of the beam, one can use in-air off-axis ratios to determine the source size.

Since in-air off-axis ratios had already been measured by a clinical physicist here at the Montreal General Hospital with the IC-10 ionization chamber a brass build-up cap at SSD’s of 80, 100 and 120 cm, for the BrainLAB Monte Carlo code, an attempt was made to follow the procedure set out by Sheik-Bagheri and Rogers using this data. The authors use a 40 x 40 cm$^2$ field size, but since this work involves the SRS mode of the Novalis TX, 15 x 15 cm$^2$ is the max field size available. Instead of simulating the details of the IC-10 chamber with the brass build-up
cap, collisional kerma was calculated from a phase-space file scored in air at 100 cm from the source to approximate the in-air measurements. The authors suggest the following calculation which can be carried out by the phase-space analysis tool “beamdp:"

\[ K_{\text{air}}(x) = \frac{1}{A} \sum_i w_i E_i \left( \frac{\mu_{\text{en}}(E_i)}{\rho} \right)_{\text{air}} \]  

Where \( A \) is total area of each bin, \( w_i \) is the statistical weight of each particle \( i \), \( E_i \) is the energy, \( \theta_i \) is the angle the particle’s direction makes with the z-axis, and \( \frac{\mu_{\text{en}}(E_i)}{\rho} \) is the mass energy absorption coefficient in air. This calculation is echoed in the work of Tonkopi et al.[12], and well as in the work of Belec.[13] The results of this characterization procedure will be detailed fully in the results section 4.1.1. However, for completeness it must be mentioned here that this procedure was not successful in producing acceptable agreement for profiles and percent depth dose curves in water. Lack of confidence in the measurements and the inability to match simulations to the measurements forced the abandonment of Sheik-Bagheri and Rogers’ procedure.

### 3.3.2 Almberg’s Method

A new procedure for source characterization from a Norwegian group was attempted given its inclusion of a third source parameter: angular spread, and that it did not employ in-air measurements. This method began in a similar manner to the one proposed by Sheik-Bagheri and Rogers, adjusting the beam energy to fit percent depth dose measurements of a 5 x 5 cm² jaw defined field. Next the source size was adjusted to agree with an EBT-2 film measurement of the penumbra of a 5 x 5 cm² jaw defined field at a depth 10 cm in solid water. Having determined the energy and source size, the angular spread was modified to fit large field profiles in water at depths of 1.5 cm and 10 cm.[14] The following sections describe in detail each step of the characterization process.

The advantage of Almberg’s method was that there was no changing of previously determined parameters, and each dose distribution it utilized was easy to measure.

#### 3.3.2.1 Beam Energy

The percent depth dose curves of three jaw defined fields in an SSD 100cm setup: 5 x 5 cm², 10 x 10 cm², and 15 x 15 cm², were measured with an IC-10 ionization chamber in the Wellhofer water tank. The measurements were smoothed and shifted by 0.6 \( r_{\text{cav}} \) upstream. No further corrections were made. The simulations of these fields were made in phantoms defined
according to the measurement geometry with the dimensions of the phantom matching the dimensions of the water tank. The phantom was divided into unequal voxel sizes to improve calculation efficiency. The X and Y dimensions of the voxels were 1 cm at the central axis, bounded by single voxels occupying the rest of the geometry. The Z dimension of the voxels was 0.2 cm, making for 1 x 1 x 0.2 cm³ voxels down the central axis. For these initial simulations, a source size of 0.1 FWHM in both the X and Y directions was used, with an angular spread of zero degrees. Several simulations were run, modifying the source’s energy from 6 to above 7 MeV in increments of 0.05 MeV.

Both the simulations and measurements were normalized to 10.5 cm depth, as normalizing to depth of dose maximum would be subject to the statistical noise and measurement uncertainty of that area. Local dose difference is calculated as a function of depth. Since depth values did not exactly line up, spline interpolation was used to determine the Monte Carlo values. The local dose differences for different energies were then plotted as a function of depth to observe any trends in the difference. Removing the data points before depth of dose maximum, a linear function was fitted to the local dose difference data. The energy associated with the local dose difference with the most constant slope was chosen to be the best fit.

3.3.2.2 Source Size

The dose distribution of a 5 x 5 cm² field was measured in SAD setup at a 10 cm depth in solid water, with 10 cm of solid water below, with EBT-2 film. 418 MUs were delivered to expose the film within the calibration range. For information on film calibration and processing, refer to section 3.1.1. Profiles approximately a half a centimeter thick were extracted from the film in both the in-plane and cross-plane directions, averaging over the pixels perpendicular to the direction of the profile. As for the Monte Carlo simulations, the phantoms dimensions matched the measurement setup. The voxel dimensions in the X and Y direction were 3 x 1 cm in the center, flanked on either side by 35 x 0.1 cm (to measure penumbra accurately) and flanked again on either side by one 10 cm voxel. The Z dimensions of the voxels were 0.2 cm centered at 10 cm depth, flanked by two voxels measuring 9.9 cm each. For source 19 in BEAMnrc, the source size can be described either with sigma (σ) or the full-width-half-maximum (FWHM), where \[ FWHM = 2\sqrt{2\ln2}\sigma \approx 2.35\sigma. \] In this work, FWHM is used to describe the source size. Many simulations were run, varying the FWHM of the source in both X and Y directions from 0.05 cm to 0.15 cm in increments of 0.01 cm. Both simulations and film measurements were normalized to the point on the central axis. The root mean square difference between simulations and film measurements were calculated for each simulated dose distribution, and the FWHM corresponding to the lowest RMS difference value was chosen.
It must be noted here that initial simulations demonstrated a discrepancy between simulated and clinical jaw definitions. According to the AAPM Task group 142, there is a 1 mm tolerance on the jaw position indicators, suggesting there could be an error of up to 1 mm in clinical jaw position. Therefore, the jaw position was varied within 1 mm until agreement was reached. The simulated jaw position only had to be moved a fraction of a millimeter. The penumbra simulations were run after the appropriate jaw positions were found.

3.3.2.3 Angular Spread

In a similar fashion to the percent depth dose curves, the 15 x 15 cm² jaw defined field profiles in water were measured with the IC-10 ionization chamber in the Wellhöfer water tank. The 15 x 15 cm² jaw defined field was chosen as it is the largest field size available in SRS mode. Although Almberg’s procedure suggested adjusting the angular spread to fit large field profiles at 1.5 cm and at 10 cm, a variety of profile sizes were measured at depths: 1.5 cm, 5 cm, 10 cm, 20 cm and 30 cm. A complete description of these measurements is presented in section 3.6.2. The simulated geometry matched the measurement geometry with 0.5 x 0.5 x 0.2 cm³ voxels. Both measurements and simulations were normalized to the central point at 1.5 cm depth. The root mean square difference between simulations and ion chamber measurements were calculated for each simulated dose distribution for both depths. The angular spread that provided the best compromise between the two RMS differences for both depths was chosen.

3.4 Reprogramming of DYNVMLC code

The focus of this work is the SRS mode of the Novalis Tx, which is equipped with a 120 leaf high-definition multi-leaf collimator, or HD120 MLC. The HD120 MLC is very similar to the Millenium MLC, but the HD120 MLC contains four types of leaves, instead of three, and the leaves are much smaller. The HD120 MLC contains two sets of target/isocenter leaf pairs: quarter leaves which project 0.25 cm at the isocenter, and half leaves which project 0.5 cm. Each leaf of these pairs has a thicker part. A leaf is called a “target” leaf when its thick part is closer to the target, and called an “isocenter” leaf when its thick part is closer to the isocenter. The HD120 MLC also has outboard leaves, which flank the entire leaf bank, but these are not considered in this work, as they are outside the scope of the largest SRS mode field size. Given that the HD120 MLC could not be modeled accurately with any available component modules, the DYNVMLC component module was reprogrammed to model the HD120 MLC, creating the HDMLC component module.

The code changes necessary to model the HD120 MLC can be summed up into three statements: 1) all full leaf information was removed, 2) all target/isocenter leaf pair code was
duplicated, changing their names to accommodate the second pair of leaves, and 3) minor sanity checks had to be modified to ensure one leaf pair could fit next to the other. The reader is encouraged to read appendix A for notes concerning specific code changes.

If one were to compare the model of the HD120 MLC with the engineering schematics, there are a few minor features in the real leaves that are not modeled. These minor features are: the support rail does not extend all the way until the leaf-end in reality, but it does in the simulation, the support rail of the quarter leaves is trapezoidal in shape (instead of rectangular in the case of the half leaves and in the simulation), the “T-nut” area of the driving screw hole is not modeled nor is the rounded driving screw end distal to it, there are small notches in the leaf-end that are not accounted for by the model, and finally, there is no “back end” to the modeled leaves where only the support rail exists. Figure 2.5 demonstrates where these differences exist.

![Figure 2.5: An approximate diagram of one of the leaves from the HD120 MLC. The red circles indicate areas where there are differences between the actual differences between the actual leaves and the leaf model.](image)

3.5 Characterization of the HDMLC component module

Having modified the DYNVMLC component module code to model the HD120 MLC, certain free parameters of the MLC model had to be adjusted before it could be used. Specifically, three free parameters of MLC were investigated: interleaf air gap, leaf density, and abutting leaf air gap. Even though initial estimates of these parameters can be made with measurements or given by the manufacturers, ultimately, the simulation must be adjusted to match measurements.

3.5.1 Interleaf air gap and leaf density

The interleaf air gap and leaf density were adjusted to attempt to match an EBT-2 film measurement of leakage through the MLC. See Figure 3.1 for an example of how such a
measurement would look like. The film was placed at a depth of 5 cm in solid water, with 10 cm solid water for backscatter, SAD setup. The jaws defined a 15 x 15 cm$^2$ field, but the MLC leaves were completely closed, blocking the entire field. 14000 MUs were delivered to sufficiently expose the film to achieve reasonable uncertainty. Such a large amount of MUs was needed as only a little over 1% of the radiation will leak through the MLC leaves, and due to the shape of the calibration curve, there is large uncertainty associated with low dose measurements with EBT-2 film. The simulated geometry matched the measurement setup, with the dimensions of the voxels at the film plane being: 2 x 0.02 x 0.4 cm$^3$, as was done by Fix et al., in their simulations of leakage.[16] Profiles were extracted from the film 2 cm away from either side of the abutment and averaged to compared with simulations 2 cm from the abutment. The rationale for averaging the measured interleaf leakage profiles on either side of the abutment will be discussed in the results section.

The interleaf air gap is the small spacing between leaves, allowing them to move independently from each other without friction. From a theoretical standpoint, increasing the simulated interleaf gap does two things: increases the overall size of the leaf bank and increases leakage between the leaves. Given that changing leaf density only has dosimetric effects on the leakage, and not any geometric effects, the interleaf air gap was adjusted to fit interleaf leakage profiles geometrically, knowing that the leaf density can be adjusted later to fit the profile dosimetrically.

Starting with the manufacturer’s recommendation of leaf density: 18.0 g/cm$^3$, the interleaf air-gap was adjusted from 0.004 cm to 0.007 cm with increments as fine as 0.0002 cm. The root mean square difference, shown in equation 2.6, was used to quantify agreement, but how the distributions agreed geometrically took precedence. As one of the simulation parameters for the HDMLC component module is the starting position of the first leaf, each time the interleaf air gap was changed, the starting position of the first leaf was adjusted to ensure that the simulated MLC leaf bank remained centered at zero. In the simulation results of Fix et al., it seems like their group kept the overall size of the leaf bank constant by adjusting the thickness of the leaves when changing the interleaf air gap. In this work, the thickness of the leaves was never varied, as this would be modifying two parameters instead of one. After finding the appropriate interleaf air gap, the density of the leaves was varied from 18 to 19 g/cm$^3$ in increments of 0.1 g/cm$^3$. In this case, the leaf density which had the lowest root mean square difference was chosen.
3.5.2 Abutting leaf gap

The measurement setup to determine the abutting leaf gap is identical to the one used to determine the leaf density and interleaf air gap. Leakage through the abutment, however, is on the order of 20%, compared to 1% in between leaves. Therefore, a second film had to be irradiated with only 1000 MU in the same setup, to deliver a dose through the abutment that would lie within the calibration curve range. The simulation setup was identical to the previous setup, but with a different voxelization: $0.02 \times 1 \times 0.4 \text{ cm}^3$ voxels centered at the film plane at 5 cm depth. Profiles 1 cm thick were extracted from the film and simulations in the cross-plane direction at the center of the quarter leaf bank. The abutting leaf gap was varied from 0 to 0.1 cm, choosing the gap that would correspond to the lowest root mean square difference with the film measurements.

3.6 Validation of Novalis Tx model

Now that the source has been characterized and the free parameters of the HD120 MLC model have been determined, the BEAMnrc model of the Novalis Tx must undergo sufficient testing to prove its capability and robustness. First, the geometry of the simulated HD120 MLC must be visualized using a “ray tracing” test. Secondly, the model must be able to reproduce measurements of profiles in water for a variety of depths and field sizes. Thirdly, the model must be able to predict accurately the output as a function of field size. And fourthly, the MLC model’s ability to model more complex distributions must be examined. Lastly, more complex tests in heterogeneous phantom geometries will be investigated, but these will be examined in the context of comparing the BEAMnrc model of the Novalis to the iPlan Monte Carlo code. Please refer to the section 3.7 for details concerning these heterogeneous slab geometry tests.

3.6.1 Ray tracing

To test the component module’s geometry, a method called “ray tracing” is used, which entails dumping a particle’s coordinates when it crosses a boundary into another medium. To accomplish this, a file is opened during the particle transport and a small line of code is added to HDMLC component module code to write the coordinates of particles to that opened file. In beamnrc.mortran, there are two lines that are added during the shower call subroutine. Before the shower call, a file is opened using the command: OPEN(UNIT=70), and then at the end of the shower call, the file is closed using: CLOSE(70); serving to open and close this file. Then, another line is written in the CM file HDMLC_cm.mortran in the HOWFAR subroutine to save the coordinates of particles as they cross boundaries. For simplicity, an accelerator is created with only the HD120 MLC model. Source 6, a rectangular beam incident from the front, is used and
the leaf media, normally tungsten, is set to air. After sufficient number of histories, the “ray
tracing” will draw the boundaries of the MLC model and illustrate the geometry.

3.6.2 Measurements in water

To test the basic performance of the Novalis Tx model, simulations of basic square fields
are compared to measurements in water. The measurements are made with the IC-10 ionization
chamber in a Wellhöfer water tank. Profiles are measured in SSD setup at depths: 1.5 cm, 5 cm,
10 cm, 20 cm, and 30 cm. Field sizes ranged from 4 x 4 cm² to 15 x 15 cm². The measured data
was corrected for point of measurement shift, as described in section 2.1.2, but no further
corrections were applied as these measurements are relative. The simulated geometry matches
the measurement geometry, with voxel sizes: 0.5 x 0.5 x 0.2 cm³. Both measurements and
simulated profiles were normalized to the central point at 1.5 cm depth. Local dose differences
between simulations and measurements were made, as described by equation 2.6, to quantify
the agreement.

![Figure 3.6: Use of the Wellhöfer water tank for profile and output factor measurements with
the Novalis Tx.](image)

3.6.3 Output Factors

While the majority of validation measurements presented here are relative, they
ultimately depend on the agreement of simulated relative dose factors with measurements. This
agreement tests the overall performance of the Monte Carlo simulation, as well as the use of a
single normalization factor to convert the simulated output: dose per particle, to dose. In the
case of small fields, these output factors also test the characterization of the source.
The relative dose factors are measured separately with both IC-10 and microLion chambers in the Wellhäuser water tank. The water tank was setup in an SAD setup, with the ion chamber positioned at 1.45 cm depth along the central axis of the beam. Both the jaws and the MLC defined the fields and are set to match a measurement setup designed by BrainLab. For square field sizes under 6 cm, the MLC defined the field size, with the jaws behind 1 mm on each side. For example, if the MLC defined a 3 x 3 cm² field, the jaws were set to 3.2 x 3.2 cm². For square field sizes above 6 cm, the MLC and jaws defined the same field size. Furthermore, for each jaw size, all MLC sizes below the jaw sizes were measured, but the reverse was not measured, as extending the MLC beyond the jaws should have no effect on the dose delivered. For example, for a 6 x 6 cm² jaw size, the MLC sizes measured were: 6 x 6 cm², 4 x 4 cm², 3 x 3 cm², 2 x 2 cm², 1 x 1 cm² and 0.5 x 0.5 cm². MLC sizes ranged from 0.5 x 0.5 cm² to 15 x 15 cm². Jaw sizes ranged from 0.7 x 0.7 cm² to 15 x 15 cm². The ion recombination corrections for the microLion chamber were applied, as described in section 2.1.2.

Simulation of these output factors was accomplished in similar geometry to the measurement setup, but with no modeling of the chambers. Because the microLion chamber was able to measure small field sizes accurately, the voxel dimensions were chosen to approximate the collecting volume of the chamber. Therefore, 0.2 x 0.2 x 0.2 cm³ voxels centered at 1.5 cm were used. Local dose differences were calculated to quantify agreement.

3.6.4 Simple MLC defined fields

To test the modeling of the MLC in the BEAMnrc model, several MLC defined fields were measured with film at a depth of 5 cm in solid water, with 10 cm solid water for backscatter. The MLC fields included: simple square MLC defined fields, and a "fence field" where either all odd or even leaves are completely retracted. More irregular MLC defined fields are described in section 3.7.1. The MUs delivered for each field were adjusted to deliver a dose to the film plane within the calibration range. Film QA was used to compare simulations with EBT-2 film measurements by extracting profiles and performing gamma analysis.

Film QA has the ability to perform dose-difference, distance-to-agreement and gamma analysis on a point by point basis for 2D dose distributions. In this work, only gamma analysis is used with the criteria of 3% dose-difference and 3 mm distance-to-agreement. In the case of the highly irregular “fence field,” no gamma analysis is performed, but instead, in-plane profiles are extracted to compare to simulations.
3.7 Comparisons with iPlan Monte Carlo code

As detailed in the introduction, the purpose of modeling the Novalis Tx was to use the model to benchmark the iPlan Monte Carlo code. Three way comparisons of both Monte Carlo algorithms and film measurements are made in numerous phantoms. More complex MLC defined fields are examined in both solid water and heterogeneous slab phantoms. Percent depth dose curves of two fields sizes are examined in three different slab phantoms containing bone, lung and solid water. And finally, cases closer to clinical reality are examined in Lucy® (see section 3.7.4) and Rando® (see section 3.7.5) phantoms. The performance of the iPlan Monte Carlo algorithm in these phantoms, with respect to film measurements and calculations done with the BEAMnrc model of the Novalis TX, will determine its suitability for clinical use.

3.7.1 Profiles in water

iPlan’s Monte Carlo code is first used to calculate similar profiles in water as described in section 3.6.2. The CT scan of a solid water phantom, on which these calculations are performed, had all of its CT number overwritten to zero, so as to perform calculations as if it was in water. Calculations were performed with the iPlan Monte Carlo algorithm. The calculation preferences are as follows: dose result type – dose-to-medium, calculation resolution – 3 mm, desired uncertainty – 0.5%, MLC modeling – accuracy optimized. A three way comparison, although possible, is not done here. Rather, iPlan’s Monte Carlo calculations are simply compared to IC-10 measurements in water. Before performing more complex comparisons, the ability of iPlan’s Monte Carlo algorithm to reproduce profiles of square fields is tested.

3.7.2 Complex MLC defined fields

Following initial validation tests described in section 3.6.4, more complex MLC fields are used to simultaneously validate the BEAMnrc model of the HD120 MLC and to test the modeling of the MLC in the iPlan system. Three fields were tested: an “S” field at 5 cm depth in solid water, a circular field at an 18 cm depth in a lung slab phantom described in Figure 3.9, and a “C” shaped field at 13 cm depth in a bone slab phantom described in Figure 3.9. Heterogeneities are introduced into this experiment as an additional complexity that the algorithms must deal with. All analysis is as described in section 3.6.4, but in this section, comparisons are drawn not only between film and the BEAMnrc model, but between iPlan’s Monte Carlo code and film, and between both Monte Carlo codes.
3.7.3 Heterogeneous slab phantoms

The three heterogeneous slab phantoms used in this work are defined in Figure 3.9. The phantoms from left to right are referred to in this work as bone, lung, and combo slab phantoms. Pieces of EBT-2 film are inserted in between each slab, as indicated by the solid black lines within the phantom. CT scans of the phantoms were taken with a Phillips Brilliance big bore scanner using a slice thickness of 2 mm. The CT scans were taken with old film pieces inserted at the appropriate locations. Each phantom was irradiated in SSD setup, with two fields: 3 x 3 cm² (MLC at 3 x 3 cm², jaws at 3.2 x 3.2 cm²) and 10 x 10 cm² (both Jaws and MLC at 10 x 10 cm²). The irradiation setups were planned on the iPlan treatment planning system, where the Monte Carlo code is used to perform the calculations. Before the calculations are performed, each slab is contoured and their Hounsfield units overwritten with an average CT value. The plan is then exported to MMCTP where it is recalculated using the BEAMnrc model of the Novalis TX. As in iPlan, in MMCTP, the physical density of each contoured slab is overwritten with the value corresponding to one used in iPlan. Equation 2.1 and 2.2 are used to convert electron densities to physical densities. Doses are extracted from the EBT-2 film measurements using Film QA. Therefore, a total of six three way comparisons are created between the film measurements and both Monte Carlo codes, as there are two fields sizes for each of the three phantoms.

3.7.4 LUCY® phantom measurement

The Lucy 3D QA phantom (Standard Imaging, Deming Way, WI) is a spherical Lucite phantom designed to aid in an overall assessment of a stereotactic system. The Lucy phantom is a versatile tool allowing for the insertion of many different types of radiation detectors, the use
of alignment pointers to perform radiation alignment QA, and the ability to lock into SRS frames.

[17] In this work, a small film cassette with a piece of EBT-2 film is inserted in the phantom. Both
the surface of the film and the Lucy sphere face the gantry with the phantom positioned at the
end of the couch as shown in Figure 3.8. A CT scan of the Lucy phantoms was taken with a
Phillips Brilliance big bore scanner using a slice thickness of 2 mm. This CT image was imported
into the iPlan treatment planning system, where a target and an organ at risk is created for
planning purposes. An irradiation plan consisting of 10 beams was created to deliver
approximately 2.5 Gy to the film plane. The plan is then exported to MMCTP to be recalculated
using the BEAMnrc model of the Novalis TX. In this work, the Lucy phantom is used to test the
end to end performance of the iPlan algorithm when multiple beams are used, similar to a
clinical case, but in a simpler spherical geometry without heterogeneities. Film QA is used to
perform gamma analysis between the film, iPlan’s Monte Carlo code and the calculation using
the BEAMnrc model, as is done in section 3.7.1.

![Figure 3.8: Setting up the Lucy Phantom. The room’s lasers are aligned with marks on the
sphere. The film cartridge containing the EBT-2 film can be seen in the Lucite sphere; it is the
black object within the sphere.](image)

3.7.5 RANDO® phantom measurement

The Rando phantom is an anthropomorphic phantom composed of materials
radiologically equivalent to soft tissue and bone.[18] The whole phantom consists of a head,
torso and pelvis, all of which are cut into slabs to allow for insertion of measuring devices. Many of the slabs include an array of holes where detectors or plugs can be inserted. Please refer to Figure 3.8 for a picture of the phantom. In this work, only the head is used. The head of the Rando phantom was scanned with a Phillips Brilliance big bore scanner using a slice thickness of 2 mm. In the iPlan treatment planning system, an irregular tumour was created near the brainstem, and a 9 beam delivery was planned, ensuring that several beams passed through the air cavities of the mouth and nasal canals. This beam arrangement would not be desirable in a clinical situation, but it is done here to add the complexity of significant heterogeneities to the experiment. CT markers placed on the head of Rando are used to define a coordinate system within the planning system. These markers are used to align the Rando head prior to irradiation. The plan is then exported to MMCTP to be recalculated using the BEAMnrc model of the Novalis TX. The tumour location was planned so as to cross the interface between two slabs of the Rando head phantom. Before irradiation, an EBT-2 film was inserted between the above slabs aforementioned. Film QA is used to perform gamma analysis between the film, iPlan’s Monte Carlo code and the calculation using the BEAMnrc model, as is done in section 3.7.1. In this work, the Rando phantom is used to test the performance of the iPlan algorithm in a realistic patient case.

Figure 3.9: The Rando head phantom with two of the slabs detached from the head to show the holes in each slab. Note the CT markers on the forehead and sides of the head in blue.
3.7.6  iPlan treatment planning system

The CT scans are imported into the iPlan treatment planning system using RT IMAGE. CT numbers are converted into electron densities using a CT calibration curve in RT DOSE created by clinical physicists here at the Montreal General Hospital. The treatment parameters: field size, gantry angle, etc. are planned using RT DOSE. For slab phantom calculations, all slabs are contoured and their CT numbers overridden with the average CT number of each slab. This is done to minimize the error introduced by the CT scanner, as the sharp edges between media are smeared out on the CT images. The densities resulting from this replacement of CT numbers were in agreement with known densities for lung, water and bone. For the LUCY® and RANDO® irradiations, no overwriting of Hounsfield units was done. For the slab phantom measurements, a small tumour contour is created in the center of the phantom in order to get the system to calculate dose, because it requires a dose prescription. For the LUCY® irradiation, a small organ at risk was created next to the tumour to introduce a little complexity into the treatment plan. For the RANDO® irradiation, an irregular tumour was created near the brainstem. Calculations in phantoms within the iPlan treatment planning system are done with its Monte Carlo algorithm. The calculation preferences are as follows: dose result type – dose-to-medium, calculation resolution – 3 mm, desired uncertainty – 0.5%, MLC modeling – accuracy optimized. The treatment information can be sent directly to the treatment machine, so the delivery can reflect the planning.

3.7.7  McGill Monte Carlo treatment planning system

The iPlan treatment plans are imported in the McGill Monte Carlo treatment planning system to allow for recalculation using the BEAMnrc model created in this work. MMCTP is able to interpret the CT images, structural contours and beam parameters created in iPlan. The CT numbers are converted to materials and densities using the same CT calibration curve manually copied from iPlan as seen in Figure 3.10. The iPlan's calibration curve maps CT number to electron density, whereas in MMCTP, CT numbers are mapped to mass densities. Therefore, mass densities from the original CT data used to generate the iPlan CT conversion curve were used to create the one used in MMCTP. Furthermore, in MMCTP, not only are CT numbers assigned to densities, but also to materials defined by PEGS files. Based on the points of the CT curve, appropriate PEGS files were associated with different areas of the CT curve. It must be noted here that due to the fundamental differences between how the Monte Carlo codes used in this work deal with materials, the CT curve used in MMCTP is only approximately equivalent to the one used in iPlan.
Interfacing with BEAMnrc and DOSxyznrc, MMCTP creates EGSPHANT files of 3 mm voxel resolution and performs simulations using the BEAMnrc model of the Novalis Tx developed in this work, according to the beam parameters as planned by the iPlan system. Sufficient histories were run to achieve a statistical uncertainty of under 0.5%. “Dose-to-medium” calculations are done using the CT curve and full material information as described above. Whereas, dose-to-water calculations are done differently than the iPlan Monte Carlo algorithm, by simply replacing all the materials in the CT conversion curve with water. Even though this may not be a realistic dose-to-water calculation, it is done simply for comparison. This calculation will be referred to as the “EGS dense water” calculation.

Figure 3.10: A screenshot of CT calibration curve used in MMCTP. This curve is based off the one used in the iPlan treatment planning system. The PEGS files associated with the CT curve can be seen on the left. However, due to the fundamental difference on how the codes handle materials, this CT curve can only be an approximation to the one in iPlan.

Before MMCTP could be used, small changes in the source code had to be made to incorporate the new HDMLC component module. Essentially, all information for the DYNVMLC component model was duplicated and renamed. The classes involved with interpreting the sample BEAMnrc input file and writing new ones had to be modified to suit the HDMLC code, as described in section 2.4.

3.8 References

1. 3cognition, *Film QA*, 2007.


Chapter 4

RESULTS AND DISCUSSION

4.1 Source Characterization

Two procedures for source characterization were attempted in this work: the results from Sheik-Bagheri and Rogers’ recommendations are seen in section 4.1.1, and the results from Almberg’s method are seen in section 4.1.2. Due to the fact that the energy determining step in both characterization procedures is similar, it is presented only once in section 4.1.1.
4.1.1 The Sheik-Bagheri and Rogers method

The beam energy that produced the best match to $5 \times 5$ cm$^2$ percent depth dose curves was 6.6 MeV. Beyond depth of dose maximum, local dose differences did not exceed 1%, as shown in the inset figure of Figure 4.1. The small difference seen near 1.5 cm could be a result of normalization and over-shifting. As Tessier pointed out in his recent paper [1], the TG-51 recommendation to shift ion chamber measurements by $0.6 r_{caV}$ upstream may be too large. His Monte Carlo simulations suggest chamber dependent shift that is generally smaller than $0.6 r_{caV}$: 0.458 for the Exradin A12 chamber, for example. [1] Although not shown here, shifting the measurements 1 mm downstream and re-normalizing to 10.5 cm can affect this difference at 1.5 cm. Positioning uncertainty of the IC-10 chamber may also factor into this problem. There is, however, no concrete basis for using a different effective point of measurement shift, as this was not studied in any thorough manner. It is simply offered as a possible explanation for the difference near 1.5 cm depth.

![Graph: IC-10 percent depth dose measurements (6.6 MeV)](image)

**Figure 4.1:** Best energy match for IC-10 percent depth dose measurements (6.6 MeV). Data before depth of dose maximum was removed as no effort was made to match the build-up region. Inset graph depicts local dose difference as a function of depth and a linear fit of those differences. The percent depth dose curves are normalized to a depth of 10.5 cm.

Although both Sheik-Bagheri and Rogers, and Almberg claim that the percent depth dose curves are not sensitive to source size [2, 3], it seems that they are affected by sufficiently small source sizes. Initial simulations performed poorly for all energies used, in that trends in local dose differences as a function of depth were observed for all energies. Furthermore,
differences greater than 2% were consistently observed. These differences were due to the 0.05 cm source size being used. Simulations using a FWHM of 0.1 cm were more successful, producing the results depicted in Figure 4.1. As the reader will see in the following sections, even though the final FWHM of the source is comparable to 0.05 cm, the final percent depth dose checks depicted in Figure 4.2 agreed with measurements to within 1% local dose difference. The introduction of angular spread may have compensated for the small source size in some way, though this was not investigated.

![Figure 4.2: Comparison of measured and simulated percent depth data for three different field sizes. All data is normalized to a depth of 1.5 cm.](image)

Having determined the energy from percent depth dose curves, the next step of the procedure was to use in-air off-axis ratios to determine source size. However, with a beam energy of 6.6 MeV, it was impossible to change the source size enough to achieve agreement with the measurements. Therefore, the energy needed to be modified to attempt to find a match. Figure 4.3 shows a survey of different energies compared to measurements. The trend in this figure of increasing off-axis ratios with decreasing energy was also reported by Sheik-Bagheri and Rogers. [3] From these results, the optimal energy was somewhere 5.8 and 6.0 MeV. It was known, however, that these energies would not agree with PDD measurements, but investigation continued. Figure 4.4 depicts a survey of different source sizes using an energy of 5.8 MeV. The trend of increasing off-axis ratios with decreasing source size described by Sheik-Bagheri and Rogers was reproduced weakly by these results, but given the size of the uncertainty, the results presented here are not conclusive. [3] From inspection of the figures
mentioned, the statistics of the simulations are not ideal. Unfortunately, even increasing the number of histories did not seem to suppress these significant variations. The graphs shown here depict the better results. Though not shown here, many of these simulations performed were inconsistent with the results of Sheik-Bagheri and Rogers. The act of fitting two variables to a single distribution also produced a range of acceptable energies and source sizes, given that both variables had similar effects. Using a low energy (~5.8 MeV) could result in agreement, but when these parameters were used to perform simulations in water, the results did not agree with measurements. Furthermore, as shown in the work of Tonkopi et al. [4] in-air off-axis ratios differ depending on the build-up caps used to perform the measurement. The approximation of using collisional kerma, described by equation 3.6, may not have been appropriate, and it was outside the scope of this work to delve into chamber modeling to find the answer. Lack of confidence in the measurements and the inability to match simulations to the measurements forced the abandonment of Sheik-Bagheri and Rogers’ procedure.

Figure 4.3: The effect of varying the energy of the electron beam incident on the target on in-air off-axis ratios. Monte Carlo simulations are depicted with error bars that represent statistical uncertainty. Measurements and simulations are normalized to the central point. In-air off-axis ratios increase with decreasing energy.
Figure 4.4: The effect of varying the FWHM of the electron beam incident on the target on in-air off-axis ratios. Monte Carlo simulations are depicted with error bars that represent statistical uncertainty. Measurements and simulations are normalized to the center. It could be said that decreasing the FWHM increases off-axis ratios, but the results here are not conclusive.

The results presented in Figure 4.5 were only performed after Almberg’s source characterization was carried out to investigate whether angular spread could have improved the agreement with in-air off-axis ratios. Increasing the angular spread clearly increased in-air off-axis ratios. This is in contrast to the results of Sheik-Bagheri and Rogers concerning angular spread. [3] Their statement that angular spread had no effect may have been specific to their accelerator model. There has been no investigation as to why angular spread should have an effect in this particular case. The small size of the SRS flattening filter may be the reason, but this has not been investigated.

Figure 4.5: The effect of varying the angular spread of the electron beam incident on the target on in-air off-axis ratios. Monte Carlo simulations are depicted with error bars that represent
statistical uncertainty. Measurements and simulations are normalized to the central point. Increasing the angular spread of the beam increases in-air off-axis ratios.

4.1.2 Almberg’s method

4.1.1.1 Energy

The determination of energy using PDDs is common to both source characterization procedures discussed in this work. Therefore, the information is presented only once in section 4.1.1.

4.1.1.2 Source size

Source size, defined by the FWHM for both the in and cross-plane directions, was adjusted to match EBT-2 film 5 x 5 cm² penumbra measurements. Percent depth dose simulations were performed using a source with the same FWHM in both the cross and in-plane directions, but here FWHMs were fitted separately. Figure 4.6 demonstrates the simulated effect of modifying the FWHM. The smaller the source size, the steeper the penumbra. The root mean square difference (RMSD) between simulations and measurements can be seen in Figure 4.7, showing both in and cross-plane directions. The in-plane FWHM is more conclusive, showing a clear minimum in RMSD at a FWHM of 0.08 cm. The FWHM in the cross-plane direction is less obvious, possibly due to noise in the film measurements. The RMSDs for FWHMs lower than 0.07 cm are quite similar to the minimum at 0.07 cm. Figure 4.8 depicts the best fit in both directions, utilizing a 0.07 cm FWHM in the cross-plane direction and a 0.08 cm FWHM in the in-plane direction.
Figure 4.6: The effect of FWHM max on simulated 5 x 5 cm² penumbra. Simulations indicate clearly that a smaller source size results in a steeper penumbra.

Figure 4.7: Root mean square differences between simulated and measured 5 x 5 cm² penumbra. The FWHMs corresponding to the lowest root mean square differences were used.

As mentioned in section 3.3.2.2, the penumbra simulations exposed the difference between simulated and clinically defined jaw sizes. The simulated jaw sizes had to be adjusted within 1 mm to match the measurements, as the clinically defined jaw sizes are never as perfect as simulated jaw sizes. The simulated jaw size was within 1 mm of the clinically defined jaw size, so it is within clinical tolerances for jaw position. [5]
Figure 4.8: Best FWHM match for in-plane and cross-plane directions for EBT-2 film measurements of a 5 x 5 cm$^2$ penumbra. Simulations and measurements are normalized to the middle point along the central axis.

4.1.1.3 Angular Spread

Figure 4.9: Effect of angular spread on simulated profiles in water at 1.5 cm depth. Increasing angular spread increases the off-axis ratio, or "horns" of the profile. The same trend is observed at 10 cm depth in water.

Having determined the source’s energy and size, the angular spread was adjusted to match 15 x 15 cm$^2$ profiles in water at depths of 1.5 and 10 cm, measured with an IC-10 ionization chamber. The effect of introducing angular spread can be seen in Figure 4.9, showing
that increasing angular spread increases the off-axis ratios, or “horns” of the profile. This trend is seen at both 1.5 and 10 cm depths, though only shown at the 1.5 cm depth. Figure 4.10 depicts the simulated distribution that best matched measurements, having an angular spread of 1.27 degrees. The odd number of 1.27 degrees was chosen as a compromise between what the RMSDs of the angular spread surveys were indicating for each depth, as slightly different angular spreads seemed to produce better fits at certain depths.

Figure 4.10: Best fit of angular spread (1.27 degrees) to match IC-10 measured profiles in water at depths of 1.5 and 10 cm. “MC” denotes Monte Carlo simulations.

4.2 Characterization of HDMLC component module

4.2.1 Interleaf air gap and Leaf Density

The interleaf air gap and leaf density were adjusted to fit an EBT-2 film measured interleaf leakage profile 2 cm from the abutment. However, the film measurements of interleaf leakage, in themselves, proved to be of interest. Considering the interlocking nature of target and isocenter leaves, one would expect a periodic trend in the leakage profile following the pattern of the leaves. Figure 4.11 and Figure 4.12 show film measurements of the leakage. While one could argue that a pattern exits in the half leaves, there is clearly no pattern in the leakage through the quarter leaf bank. The lack of patterned leakage is most likely due to minute differences in leaf dimensions and position in space. Figure 4.13 demonstrates the simulated effect of changing the interleaf air gap. The simulated results show that a small change in interleaf spacing results in large difference in leakage. Therefore, it is feasible to assume that imperfections in leaf dimensions and spacing would results in significant differences in leakage.
This can be seen particularly in the case of the quarter leaves, where imperfections would account for a larger fraction of the leaves, given their small size. One could hypothesize that this is why a patterned leakage profile is not seen in measured distributions. Figure 4.11 shows interleaf leakage profiles 2 cm on either side of the abutment, and their average. Each leaf bank, therefore, has its own unique interleaf leakage profile. The average of the two leakage profiles was used to fit the leaf density and interleaf air gap.

Other hypothesises that could explain the irregular pattern in the leakage were film noise and placement of the leaves in space. When handling the leaves to check dimensions indicated on the engineering schematics, the leaves were found to have some spatial play. In investigating other MLCs, such as the Millennium MLC on one of the Varian Clinac linear accelerators here at the Montreal General Hospital, there was clear evidence on some of the leaves that they actually rubbed together at some places. It could be feasible, then, that for a given delivery (gantry position, leaf movement) leaf positions could change slightly. Performing a similar interleaf leakage film measurement, pausing the delivery and inter-digitating the leaves several times throughout the 14000 MUs to hopefully disrupt the placement of the leaves in space, produced the same interleaf leakage profile. This measurement simultaneously ruled out both hypothesises mentioned here. Strengthening this point, Figure 4.12 shows that the irregular interleaf pattern remained stable in both SRS and normal modes, even though those measurements were performed weeks apart.

![Figure 4.11: EBT-2 film measurements of interleaf leakage profiles 2 cm above and below the abutment. Note that the leakage profile is highly irregular, especially in the quarter leaf bank. These irregularities are real, and not due to film noise.](image)
Figure 4.12 demonstrates how the leakage varies given the treatment mode selected. The SRS mode leakage is approximately 0.2% lower than in normal mode. This is likely due to the different flattening filters used in these modes. The normal mode utilizes a flattening filter capable of producing square fields with sides up to 40 cm, while the SRS flattening filter, being much smaller, is limited to 15 cm. The large normal flattening filter may, therefore, produce more beam hardening, resulting in greater leakage through the leaf banks. However, this 0.2% difference may be exaggerated due to how the transmission measurements were normalized. Tabulated output factors and TMR tables were used to calculate the dose delivered to the film plane without the MLC blocking the beam. This dose is then used to normalize the transmission measurements. It is conceivable, therefore, that a small error in output factor measurements could play a role in this 0.2% difference.

![Figure 4.12: EBT-2 film measured interleaf leakage profiles 2 cm away from the abutment for the normal and SRS modes of the Novalis Tx. There is about a 0.2% decrease in leakage through the leaves of the MLC in SRS mode, compared to normal mode.](image)

Figure 4.13 demonstrates the simulated effect of modifying the interleaf air gap. Increasing the interleaf air gap increases both the intra and interleaf leakage. Furthermore,
increasing the interleaf air gap also increases the overall size of the leaf bank. Geometric agreement was the determining factor for selecting an interleaf air gap. An interleaf air gap of 0.0047 cm was chosen. The interleaf air gap chosen by Fix et al. for their Swiss Monte Carlo planning system was 0.006 cm. [6] However, from their graphs, it is apparent that they kept the overall size of the leaf bank constant, while changing leaf thicknesses to compensate for changes in interleaf gap. This was not done in this work. Given the earlier discussion of the imperfections of the interleaf leakage profile, it is conceivable that the leaves modeled by Fix et al. are simply slightly more spaced out.

Figure 4.13: The effect of varying the interleaf air gap. Increasing the interleaf gap increases leakage and the overall size of the leaf bank, as can be seen in the offsets near 0 cm.

Figure 4.14 demonstrates the simulated effect of varying the density of the MLC leaves. Logically, increasing the density of the leaves decreases overall transmission. Two densities: 18.75 and 18.8 g/cm³, had similar root mean square differences when comparing to measurements. Therefore, a density of 18.775 g/cm³ was chosen as a compromise. For comparison, the density found by Fix et al. for their HD120 MLC leaves was 18.53g/cm³. [6] Both density values disagree with the manufacturer’s specification of 18.0 g/cm³.

With an interleaf air gap of 0.0047 cm and a leaf density of 18.775 g/cm3, Figure 4.15 depicts the best simulated match to EBT-2 film measurements. It is clear that the match is not perfect, but it would not be feasible to match the irregular leakage pattern specific to the HD120
MLC. Too many simulations would have to be run to attempt to model each individual interleaf air gap.

Another variable that could have been modified, but was not, was the size of the support rail. Given that HDMLC component module did not model the triangular cut off the quarter leaf support rail, the length of the support rail could have been varied to compensate. Reducing the size of the support rail in the quarter leaf bank would increase leakage. However, it would be inadvisable to introduce a third variable into this problem. Interleaf air gap and leaf density are already intertwined with respect to their effects on leakage, in the sense, that one could increase leaf density and interleaf air gaps and still get similar leakage. Another variable would simply confuse things, as changing one variable would necessitate adjusting the other two. Referring back to section 4.1.1, adjusting two variables to match a single distribution could produce a range of acceptable variable combinations. And finally, given that simulations were already being adjusted to fit irregular data, the support rail size was ignored in this work.

![Figure 4.14: The effect of varying the leaf density on the leakage through the MLC. Increasing the leaf density decreases the leakage.](image)

Figure 4.15: Best simulated match for the EBT-2 measured interleaf leakage profile, using an interleaf air gap of 0.0047 cm and a leaf density of 18.775 g/cm³.

4.2.2 Abutting leaf gap

Figure 4.16: Best fit for abutting leaf gap (0.004 cm) to match EBT-2 film measurements for abutting leaf leakage. Discrepancies outside the abutment are not significant given the agreement possible for the transmission through each pair of quarter leaves.

The abutting leaf gap was adjusted to match EBT-2 measurement abutting leaf leakage. According to the root mean square differences between simulations and measurements, an
abutting leaf gap of 0.004 cm was the best. Figure 4.16 depicts both the best simulated distribution and film measurements. The discrepancies away from the abutment can be expected given the agreement between simulated and measured leakage through the quarter leaf bank. Refer to Figure 4.15 to see how that agreement. Referring to Emily Heath’s modeling of the Millennium MLC with the DYNVMLC component module [7], the exact same abutting leaf gap was found, and similar discrepancies exist with film measurements away from the abutment. Heath found that the abutting leaf leakage for the Millennium MLC is approximately 27%, in comparison to just over 21% in the case of the HD120 MLC in this work.

4.3 Model validation

4.3.1 Ray trace of the HDMLC

![Ray trace of the HDMLC](image)

Figure 4.17: Ray trace of the Y-Z plane through a realistic simulation of the HD120 MLC using the HDMLC component module.

To demonstrate that the HDMLC component module was indeed modeling the complex geometric of the HD120 MLC, a “ray trace” was performed to trace out material boundaries in the MLC model. Figure 4.17 shows a ray trace through a realistic model of the HD120 MLC. The support rails, driving screw holes, leaf tips, tongues and grooves are all clearly visible. The model
also correctly simulates the focused nature of the leaf sides. Figure 4.18 shows an unrealistic model of the HD120 leaves, with exaggerated dimensions for visual clarity. Here, it is possible to see the interleaf air gaps in between the leaves. Barring the small details that are not modeled by the HDMLC component module, as mentioned in section 3.4, this module effectively simulates the complex geometric of the HD120 MLC. The reader is referred to the photograph shown in figure 1.6 in section 1.4, for a comparison to the ray trace depicted in Figure 4.17.

Figure 4.18: Ray trace of the Y-Z through a non-realistic simulation of the HD120 leaves using the HDMLC component module. Leaf dimensions and air gaps have been enlarged for illustration purposes.

4.3.2 Profiles in water

The following three figures: Figure 4.19, Figure 4.20, and Figure 4.21 show the agreement between simulations and IC-10 measured profiles for 5 x 5 cm², 10 x 10 cm² and 15 x 15 cm² field sizes, respectively. In all cases, the in-field local dose differences do not exceed 2% for all depths. Discrepancies exist outside the field edge, where Monte Carlo simulations are lower than the IC-10 measurements. The cause for these differences may be the electronics of the Wellhöfer water tank positioning device. From these in-field results, it can be concluded that the source and MLC have been properly characterized for the purposes of modeling square fields in water for a range of field sizes.
Figure 4.19: Comparison between Monte Carlo simulations and IC-10 profile measurements at several depths in water for a 4 x 4 cm² field. The black dots represent Monte Carlo simulations while the coloured lines represent IC-10 measurements.

Figure 4.20: Comparison between Monte Carlo simulations and IC-10 profile measurements at several depths in water for a 10 x 10 cm² field. The black dots represent Monte Carlo simulations while the coloured lines represent IC-10 measurements.
Figure 4.21: Comparison between Monte Carlo simulations and IC-10 profile measurements at several depths in water for a 15 x 15 cm$^2$ field. The black dots represent Monte Carlo simulations while the coloured lines represent IC-10 measurements.

4.3.3 Output factors

Figure 4.22 depicts measured output factors using both the microLion and IC-10 ionization chambers. Given the relatively large size of the IC-10 chamber, at smaller fields, volume averaging occurs. This means that there is a non-uniform dose across the sensitive volume of the chamber. In other words, the signal is averaged over a larger area of the profile including some of the field’s penumbra, which, in turn, results in a lower signal. Due to the microLion’s ability to measure output factors of small fields properly, the microLion data was used to compare to simulations. Above the jaw defined field size of 3.2 x 3.2 cm$^2$, however, the IC-10 chamber measurements agree closely with the microLion chamber measurements.
Figure 4.22: Measurement of output factors with the IC-10 and microLion chambers. The IC-10 chamber’s signal drops for small field sizes due to the size of the collecting volume. Output factors are normalized to the 10 x 10 cm² field signal.

The simulated and microLion measured output factors, along with their local differences, are plotted in Figure 4.23. All fields agree within 1.25% local dose difference, with the largest difference being the 0.7 x 0.7 cm² jaw defined field. Given the difficulty of measurement under these conditions and that the chamber itself was not modeled, the simulations match very well with the microLion measurements. From these results, it can be concluded that the BEAMnrc model of the Novalis Tx models output factors for square fields in water correctly.
Figure 4.23: Comparison of Monte Carlo simulated and microLion measured output factors. The inset graph depicts the local dose difference between the two as a function of jaw defined field size. The largest local difference is 1.25% at a jaw size of 0.7 cm.

4.3.4 Simple MLC defined fields

To evaluate the performance of the BEAMnrc model of the Novalis Tx, and specifically the modeling of the HD120 MLC, several simulated MLC defined fields were compared to EBT-2 film measurements. The results presented here confirm the validity of the BEAMnrc model of the Novalis TX. Figure 4.4.24 depicts the gamma analysis for two 2 x 2 cm² MLC defined fields, with jaws set to 15 x 15 cm², centrally (left) and 5 cm offset from the abutment (right). Figure 4.25 depicts a 4 x 4 cm² MLC defined field, with jaws set to 15 x 15 cm². In all three cases, over 99% of the pixels passed the 3%, 1 mm gamma analysis. The analysis was restricted to a region of interest surrounding the MLC defined field, shown as the red box in the figures. Discrepancies exist in the penumbra of the fields possibly due to the resolution of the Monte Carlo simulations. The profile on the right of Figure 4.25 is another way to illustrate the agreement. The profile shows that the discrepancies mostly lie in the penumbra region, and that the film measurement is quite noisy. Similar profiles were drawn for the fields depicted in Figure 4.24, but these are very similar to the one shown in Figure 4.25. Figure 4.26 shows a profile 2 cm away from the abutment for an “even fence field,” where all the even leaves are retracted. The simulations and film measurements are in excellent agreement with one another except in the peaks within the
quarter leaf bank. These differences are most likely due to the under sampling of the film by the software Film QA, as described in section 3.1.1.3.

Figure 4.24: Gamma maps for a 2 x 2 cm² MLC defined fields, with jaws set to 15 x 15 cm², centrally (left) and 5 cm offset from the abutment (right). Over 99% of the pixels passed the gamma analysis for both fields. Pixels that did not pass lay in the penumbra regions.

Figure 4.25: Gamma map (on the left) for a 4 x 4 cm² MLC defined field, with jaws set to 15 x 15 cm². Over 99% of the pixels passed the gamma analysis. Pixels that did not pass lay in the penumbra regions. A horizontal profile through the field above the abutment is shown on the right. The thick green line represents the Monte Carlo calculation, while the more noisy light green line represents the film measurement. The blue line centered on zero represents the difference between the film and Monte Carlo dose.
Figure 4.26: Even “fence” field profile comparison between the EBT-2 film measurement and Monte Carlo simulation. Profile was extracted from the film 2 cm away from the abutment using Film QA. The disagreement in the central leaf bank is a results of the sampling of the film measurement.

4.4 Comparisons with iPlan Monte Carlo code

Three way comparisons between film measurements, the BEAMnrc model of the Novalis Tx, and the iPlan Monte Carlo code, simultaneously serve to both validate the BEAMnrc model, as well as benchmark the iPlan Monte Carlo code. The three way comparisons in this work include: complex MLC defined fields, heterogeneous slab phantom measurements, one 10 beam irradiation plan on a LUCY phantom and 9 beam irradiation plan on a RANDO® phantom. The MLC fields check the algorithms’ ability to model the complexity of the MLC. The heterogeneous slab phantom experiments show how the algorithms perform in varying media. The LUCY phantom irradiation demonstrates a more clinical situation on a simplified spherical phantom. And finally, the RANDO phantom irradiation checks the overall performance of the algorithm to predict dose from many beams in the presence of heterogeneous media, without performing any additional corrections.
4.4.1 Profiles in water

The following three figures: Figure 4.27, Figure 4.28, and Figure 4.29 show comparisons of iPlan’s Monte Carlo code and IC-10 measurements for 4 x 4 cm$^2$, 10 x 10 cm$^2$, and 15 x 15 cm$^2$ square fields, respectively. In these figures, the coloured lines represent the Monte Carlo simulations, while the black dots show IC-10 measurements. Similar to the cases shown in section 4.3.2, the in-field local dose differences do not exceed 2% for all depths. Discrepancies exist outside the field edge, where Monte Carlo simulations are lower than the IC-10 measurements. The source for this discrepancy is discussed in section 4.3.2. From these in-field results, it can be concluded that the iPlan Monte Carlo algorithm is capable of predicting dose accurately for square fields in water for a range of field sizes.

Figure 4.27: Comparison between iPlan’s Monte Carlo simulations and IC-10 profile measurements at several depths in water for a 4 x 4 cm$^2$ field. The black dots represent IC-10 measurements while the coloured lines represent Monte Carlo simulations.
Figure 4.28: Comparison between iPlan’s Monte Carlo simulations and IC-10 profile measurements at several depths in water for a 10 x 10 cm² field. The black dots represent IC-10 measurements while the coloured lines represent Monte Carlo simulations.

Figure 4.29: Comparison between iPlan’s Monte Carlo simulations and IC-10 profile measurements at several depths in water for a 15 x 15 cm² field. The black dots represent IC-10 measurements while the coloured lines represent Monte Carlo simulations.
4.4.2 Complex MLC defined fields

Figure 4.30 depicts the first three way comparison between the BEAMnrc model, iPlan’s Monte Carlo algorithm and film measurements. All the Gamma analyses use the criteria: 3% / 1mm and are restricted to regions of interest shown as red squares on the figures. When film is compared to the calculations with BEAMnrc model, over 94% of pixels passed. When film is compared to iPlan’s calculations, over 91% of pixels pass the analysis. Both the BEAMnrc model and the iPlan Monte Carlo seem to have similar difficulties outside the field edges. BEAMnrc and iPlan, however, are in excellent agreement (99% of the pixels passed) when compared to one another. Small differences exist along the complex field edge. Figure 4.31 depicts a profile through this “S” field demonstrating that the primary areas of disagreement between the Monte Carlo codes and film are in penumbra regions or outside the MLC defined field. A profile comparing the Monte Carlo codes is not included because the profiles are practically indistinguishable from one another.

Figure 4.30: Three gamma analyses for the “S” MLC defined field, comparing the BEAMnrc model of the Novalis Tx to film (left), iPlan Monte Carlo to film (center) and the BEAMnrc model with iPlan (right). When film is compared to the calculations with BEAMnrc model, over 94% of pixels passed the 3 %, 1 mm gamma analysis. When film is compared to iPlan’s calculations, over 91% of pixels pass the analysis. Both the BEAMnrc model and the iPlan have similar disagreements with film outside the MLC defined field. The BEAMnrc model and the iPlan agree very well (99% pixels passed), with some discrepancies at the edge of the MLC defined field, possibly due to differences in voxel resolution.
Figure 4.31: A profile through the "S" MLC defined field, comparing the BEAMnrc model’s calculation with film (left), and iPlan’s MC calculation with the film measurement (right). The thick green line represents the Monte Carlo calculation, while the more noisy light green line represents the film measurement. The blue line centered on zero represents the difference between the film and Monte Carlo dose. This figure demonstrates that the primary areas of disagreement are in penumbra regions or outside the MLC defined field.

Figure 4.32 depicts the gamma analyses for a “C” shaped MLC defined field, where the film is placed at 13 cm depth in a bone slab phantom described in figure 3.7. Comparing the BEAMnrc calculation with film (left), only 94% of the pixels passed the gamma analysis, compared to the 98% when film is compared to iPlan (center). At the top of these two figures, there is a large area above the “C” shape that does not pass the gamma analysis. The film calibration or simply the film itself may be the cause of the problem. The calibration curve in this work is not designed for measurement of low dose areas. And it is also feasible that the film may have been bent or exposed to too much ambient light. Comparing the two Monte Carlo codes (right), 100% of the pixels pass the gamma analysis. Even though the comparisons with film are not ideal, the good agreement between the Monte Carlo codes increases confidence in the iPlan Monte Carlo dose calculation algorithm. Similarly to Figure 4.31, Figure 4.33 shows a profile through the MLC defined field. Figure 4.33 demonstrates again that the primary areas of disagreement between the Monte Carlo codes and film are in penumbra regions or outside the MLC defined field. A profile comparing the Monte Carlo codes is not included because the profiles are practically indistinguishable from one another.
Figure 4.32: Three gamma analyses for the “C” MLC defined field, comparing the BEAMnrc model of the Novalis Tx to film (left), iPlan Monte Carlo to film (center) and the BEAMnrc model with iPlan (right). Comparing the BEAMnrc calculation with film (left), only 94% of the pixels passed the gamma analysis, compared to the 98% when film is compared to iPlan (center). Areas at the edge of the MLC defined field that did not pass may be a result of the 3 mm voxel sizes used in the calculations. The comparisons with film (left, center) show a significant difference at the top of the figures. Considering, however, the comparison between Monte Carlo codes (right), over 100% of the pixels passed the analysis.

Figure 4.33: A profile through the "C" MLC defined field, comparing the BEAMnrc model's calculation with film (left), and iPlan’s MC calculation with the film measurement(right). The thick green line represents the Monte Carlo calculation, while the more noisy light green line represents the film measurement. The blue line centered on zero represents the difference between the film and Monte Carlo dose. This figure demonstrates again that the primary areas of disagreement are in penumbra regions or outside the MLC defined field.

Figure 4.34 shows the gamma analyses for a circular field with a film placed at an 18 cm depth in a lung slab phantom described in figure 3.7. Comparing the BEAMnrc
calculation with film (left), only 94% of the pixels passed the gamma analysis, compared to the 96% when film is compared to iPlan (center). As in the previous case, there is a large area of the film, on the left in this case, that does not agree well with calculations. When the Monte Carlo codes are compared, 99% of the pixels pass the gamma analysis, increasing confidence in the predictive ability of iPlan’s algorithm. Figure 3.35 demonstrates again that the primary areas of disagreement between the Monte Carlo codes and film are in penumbra regions or outside the MLC defined field. A profile comparing the Monte Carlo codes is not included because the profiles are practically indistinguishable from one another.

Figure 4.34: Three gamma analyses for the “circle” MLC defined field, comparing the BEAMnrc model of the Novalis Tx to film (left), iPlan Monte Carlo to film (center) and the BEAMnrc model with iPlan (right). Comparing the BEAMnrc calculation with film (left), only 94% of the pixels passed the gamma analysis, compared to the 96% when film is compared to iPlan (center). In the comparison between Monte Carlo codes (right), over 100% of the pixels passed the analysis. There are consistent areas of disagreement in all three comparisons, located at the edge of the circle. As in the previous three way comparison in Figure 4.32, there is an area of the film that shows a significant difference with calculations at the left of the figures.
Overall, from these complex MLC defined field comparison between the Monte Carlo codes and EBT-2 film, two conclusions can be drawn: agreement between the Monte Carlo codes and film is good, but not ideal, particularly in penumbra regions and outside the MLC defined field, and iPlan’s Monte Carlo code agrees very well with the BEAMnrc model of the Novalis TX. In particular, these experiments demonstrate that the iPlan Monte Carlo algorithm models the HD120 MLC in an accurate way, similar to the HDMLC component model created in this work.

4.4.3 Heterogeneous slab phantom measurements

Three way percent depth dose comparisons between EBT-2 film, the EGSnrc Monte Carlo calculation using the BEAMnrc model of the Novalis TX, and the iPlan Monte Carlo code, were made in heterogeneous slab phantoms. As described in the Materials and Methods section, the bone slab phantom consists of a 3 cm slab of bone in solid water, the lung slab phantom consists of 8 cm slab of lung in solid water, and the combo slab phantom consists of both 3 cm of bone followed by 8 cm of lung in solid water. Both 3 x 3 cm$^2$ and 10 x 10 cm$^2$ square fields are used for each phantom. In all of the figures presented in this section, film measurements are depicted with 2.82% error bars, given the factor of $\sqrt{2}$ introduced to the 2% absolute dose error due to normalization. The 2% dose error is predicted by Devic’s film protocol[8], as well as confirmed by the standard deviations of the measurements determined in
FILM QA. [9] EGS dose-to-medium calculations are performed in MMCTP, calculating dose-to-medium with overwritten physical densities. EGS dense water calculations are performed in MMCTP setting all media to water with densities determined using the CT curve. IPlan dose-to-medium and dose-to-water calculations simply reflect the chosen dose result type chosen in the iPlan treatment planning system as described in section 2.1.2. Please refer to section 2.1.4 for a description of the difference between the two dose result types.

For the bone slab phantom case, Figure 4.36 and Figure 4.38 show the results for the 3 x 3 cm² and 10 x 10 cm² square fields, respectively. The film measurements are normalized to the Monte Carlo calculated values at 2 cm depth, where the first film is placed. Both field sizes produce similar results, except that the agreement with film is poorer in the 3 x 3 cm² case beyond the bone. In general, the EGS dose-to-medium and iPlan dose-to-medium calculations agree. Differences between those calculations exist at boundaries between solid water and bone. These differences may reflect how the algorithms deal with boundaries, but they may also originate from how the different treatment planning systems deal with contouring structures. It is not given that a set of contoured structures, as created in iPlan, translates exactly to MMCTP. This hypothesis was based on problems that were encountered in MMCTP when certain structures overlapped, which did not occur in the iPlan system. Unfortunately, these problems were not investigated. Furthermore, the 3 mm cubed voxel size used in these calculations may not be sufficiently small to properly visualize features at boundaries. Film measurements agree much better with iPlan’s dose-to-water calculations, which is to be expected given that the film is nearly water equivalent. One could consider that the film is a small cavity of water equivalent material within bone. From these results in the bone heterogeneity, Kawrakow’s method for converting dose-to-medium to dose-to-water provides a better prediction to dose to EBT-2 film than would Siebers’ method. Kawrakow’s method uses a stopping power ratio of approximately 1.09, in this case, compared to 1.11 according to Siebers. Figure 4.37 gives another perspective on the situation, as an EGSnrc calculation is performed on a mathematical phantom with varying voxel sizes to visualize the features at boundaries. There is a sharp dose build-up before the bone and a sharp drop in dose before going back into solid water. These features are seen in all calculations and measurements to varying degrees. However, a notable difference between the iPlan dose-to-medium and dose-to-water calculations exists where there should be no difference: in the water before the bone. The dose-to-water calculation seems to model the build-up before the bone, as shown in Figure 4.37, more than the dose-to-medium calculation. The EGS dense water calculations are not shown as they do not agree well with any algorithm or the film measurements.
Figure 4.36: 3 x 3 cm$^2$ field incident on the bone slab phantom. Film measurements are normalized to the Monte Carlo calculated dose at 2 cm depth.

Figure 4.37: 3 x 3 cm$^2$ field incident on a mathematical bone slab phantom. The phantom and calculation were made using DOSxyzmrC, therefore, CT artefacts do not affect the calculation whatsoever. The voxel dimensions near boundaries have a z-dimension of half a millimeter in order to visualize the detail of the dose at boundaries.
Figure 4.38: 10 x 10 cm² field incident on the bone slab phantom. Film measurements are normalized to the Monte Carlo calculated dose at 2 cm depth.

For the lung slab phantom case, Figure 4.39 and Figure 4.41 show the results for the 3 x 3 cm² and 10 x 10 cm² square fields, respectively. Again, the film measurements are normalized to the Monte Carlo calculated values at 2 cm depth, where the first film is placed. The 3 x 3 cm² case demonstrates a significant perturbation in the dose when compared to the 10 x 10 cm² case, most likely due to the small field size. The film measurements in the 3 x 3 cm² case within the lung agree more with the EGSnrc and iPlan dose-to-medium calculations than the iPlan dose-to-water calculations. Looking beyond the lung, the difference between the EGS and iPlan calculations suggests that the density chosen for the lung in MMCTP may be have been too high. This hypothesis is further strengthened by the differences between the calculations in the lung. All calculations for the 3 x 3 cm² case demonstrate the build-up in the solid water downstream from the lung, as shown in the mathematical calculation displayed in Figure 4.39. While the perturbation that the lung introduces is noticeable in the 10 x 10 cm² case, all algorithms predict similar doses and they agree with film. The EGS dense water calculations provide similar results to the other algorithms, which is not surprising since the stopping power ratio between lung and water is not far from 1. The small differences between the iPlan calculations demonstrate the use of this stopping power ratio. In general, the iPlan and EGSnrc dose-to-medium calculations follow the same trend.
Figure 4.39: 3 x 3 cm² field incident on the lung slab phantom. Film measurements are normalized to the Monte Carlo calculated dose at 2 cm depth. The EGS dense water calculation is accomplished by setting all media to water with variable densities, as determined by the Hounsfield to density conversion curve in MMCTP.

Figure 4.40: 3 x 3 cm² field incident on a mathematical lung slab phantom. The phantom and calculation were made using DOSxyzrc, therefore, CT artefacts do not affect the calculation whatsoever. The voxel dimensions near boundaries have a z-dimension of half a millimeter in order to visualize the detail of the dose at boundaries.
Figure 4.41: 10 x 10 cm$^2$ field incident on the lung slab phantom. Film measurements are normalized to the Monte Carlo calculated dose at 2 cm depth. The EGS dense water calculation is accomplished by setting all media to water with variable densities, as determined by the Hounsfield to density conversion curve in MMCTP.

For the combo slab phantom case, Figure 4.42 and Figure 4.44 show the results for the 3 x 3 cm$^2$ and 10 x 10 cm$^2$ square fields, respectively. Due to the sharp build-up at the solid water-bone interface at 2 cm depth, the film measurements were normalized to the film at 19 cm depth, instead of at 2 cm depth. In retrospect, it was not wise to have an interface between solid water and bone where the film measurements would be normalized. Given the complexity of the measurement, the mathematical case shown in Figure 4.43 provides some context. There is a sharp build-up before the bone, a sharp drop in dose at the bone-lung interface followed by another sharp increase of dose at the beginning of the lung, and finally another build-up in the solid water following the lung. Looking back at Figure 4.42, the film measurements at the interfaces between solid water and bone (2 cm depth) and between bone and lung (5 cm) show respectively the build-up and the sharp drop in dose seen in the mathematical case. The features previously mentioned are difficult to pick out in any of the calculations. For both field sizes, a build-up can be observed in many of the calculation at the first solid water-bone interface. In the 10 x 10 cm$^2$ case, the drop in dose at the bone-lung interface can be seen in some of the calculations, especially in the EGSnrc dose-to-medium case. As in the bone slab case, similar discrepancies exist between the algorithms at boundaries, but these may not be very significant, as discussed earlier. In contrast with the bone slab case, the film measurement in the bone for
the 3 x 3 cm² case is in between iPlan’s dose-to-medium and dose-to-water calculations. In the 10 x 10 cm² case, the film measurement agrees most with iPlan’s dose-to-water calculation. The measurement within the lung agrees most with the EGSnrc dose-to-medium calculation. But as in the lung slab case, there is a divergence of the algorithms within and following the lung, hinting to a difference in lung density used for the calculations. The measurement at the lung-solid water interface agrees most with EGSnrc dose-to-medium but also the EGS dense water calculation. The film measurement beyond the lung is the normalization point, so it is no surprise that it agrees with the iPlan algorithms. Overall, the complexity of this experiment makes it difficult to draw many conclusions. The behavior of the film measurements can be rationalized with the aid of the simulations in Figure 4.43, but given the 3 mm cubed voxel size used in this experiment, it is unlikely that any of the algorithms had the ability predict what was measured by the film.

Figure 4.42: 3 x 3 cm² field incident on the combo slab phantom. Monte Carlo simulations are normalized to the dose at 1.5 cm depth. Film measurements are normalized to the Monte Carlo calculated dose at 19 cm depth, as the measurement at 2 cm depth is at an interface between two media. The EGS dense water calculation is accomplished by setting all media to water with variable densities, as determined by the Hounsfield to density conversion curve in MMCTP.
Figure 4.43: 3 x 3 cm$^2$ field incident on a mathematical combo slab phantom. The phantom and calculation were made using DOSxyznr, therefore, CT artefacts do not affect the calculation whatsoever. The voxel dimensions near boundaries have a z-dimension of half a millimeter in order to visualize the detail of the dose at boundaries.

Figure 4.44: 10 x 10 cm$^2$ field incident on the combo slab phantom. Monte Carlo simulations are normalized to the dose at 1.5 cm depth. Film measurements are normalized to the Monte Carlo dose.
Carlo calculated dose at 19 cm depth, as the measurement at 2 cm depth is at an interface between two media. The EGS dense water calculation is accomplished by setting all media to water with variable densities, as determined by the Hounsfield to density conversion curve in MMCTP.

Considering the above three-way slab phantom comparisons in general, the EGSnrc and iPlan algorithms predict similar results, which strengthens confidence in the validity of iPlan's algorithms. However, in the context of validating an algorithm, the film placement at interfaces between materials was not ideal. The large size of the voxels used in the calculations made it impossible for the algorithms to pick up the features at the interfaces measured with film. These features can only be seen when the voxel size is reduced significantly near boundaries, as is done in mathematical phantom calculations seen in Figure 4.37, Figure 4.40 and Figure 4.43. While the voxel sizes can be adjusted in both MMCTP and iPlan, MMCTP terminated abruptly when dimensions under 3 mm were used, and the iPlan treatment planning computer is used clinically, so it cannot afford a lengthy calculation that would likely take days.

Considering, as well, the use of film to measure dose in bone heterogeneity, it is not obvious whether this is appropriate. Bragg-Gray cavity theory states that if a cavity is small enough so that the absorbed dose in the cavity is solely due to charged particles crossing the cavity (and not created in the cavity) and the cavity does not disturb the fluence of charged particles in the medium, that the dose to the medium can be calculated by multiplying the dose to the cavity by a stopping power ratio as described in section 2.1.2. [10] It is unclear whether a film placed in between slabs of bone fulfills the conditions of this cavity theory to warrant a simple multiplication of stopping power ratios. If we assume Bragg-Gray conditions, then our results suggest that Kawrakow's method is more appropriate for calculating dose to film in bone in this case, as compared to Siebers' method.

4.4.4 Lucy phantom measurements

Figure 4.45 depicts three gamma analysis for the Lucy phantom irradiation. All the Gamma analyses use the criteria: 3% / 1mm and are restricted to regions of interest shown as red squares on the figures. In the comparison between EBT-2 film and the calculation using the BEAMnrc model, 98% of the pixels pass the gamma analysis. Similarly, 98% of the pixels pass the analysis when film is compared to iPlan’s calculation. The comparisons with film show discrepancies outside the “target” region. These differences could be a result of setup errors or due to not correcting for attenuation through the couch. When both Monte Carlo codes were compared, 100% of the pixels passed the gamma analysis. The extremely good agreement
between the Monte Carlo codes demonstrates that the iPlan Monte Carlo dose calculation algorithm can accurately predict dose for a clinically realistic plan in a spherical geometry. Figure 4.46 shows profiles comparing the Monte Carlo codes to film measurements.

Figure 4.45: Three gamma analyses for the Lucy phantom irradiation, comparing the BEAMnrc model of the Novalis Tx to film (left), iPlan Monte Carlo to film (center) and the BEAMnrc model with iPlan (right). In the comparison between EBT-2 film and the calculation using the BEAMnrc model, 98% of the pixels pass the gamma analysis. Similarly, 98% of the pixels pass the analysis when film is compared to iPlan’s calculation. Both comparisons to film show discrepancies outside the “target” region. When both Monte Carlo codes were compared, 100% of the pixels passed the gamma analysis.

Figure 4.46: A profile through the Lucy dose distribution, comparing the BEAMnrc model’s calculation with film (left), and iPlan’s MC calculation with the film measurement(right). The thick green line represents the Monte Carlo calculation, while the more noisy light green line represents the film measurement. The blue line centered on zero represents the difference between the film and Monte Carlo dose. Both Monte Carlo codes have similar disagreement with film in-field and around 30.0 mm.
4.4.5  **RANDO® phantom measurements**

Figure 4.47 shows three gamma analyses for the Rando phantom irradiation. All the Gamma analyses use the criteria: 3% / 1mm and are restricted to regions of interest shown as red squares on the figures. Comparing the calculation using the BEAMnrc model and the EBT-2 film measurement (left), only 80% of the pixels passed the gamma analysis. Comparing iPlan’s Monte Carlo code with film measurements, 86% of pixels passed the gamma analysis. Both comparisons to film show discrepancies outside the planned tumour area. These differences may be the result of warping of the EBT-2 film. After the irradiation plan was delivered, the film was left in the Rando phantom for 24 hours before it was scanned. When the film was removed, it was clear that the plugs within the slabs of the Rando head phantom had made many imprints on the film. It might have been more prudent to remove the film following its irradiation. Another possible reason for the discrepancy was that the resolution of the dose calculation: 3 mm, may have been too large to properly account for the air gaps between the slabs of the Rando phantom. Although the phantom is designed to fit together snugly, the air gaps between the slabs are visible. Finally, as in the Lucy irradiation, setup errors and not accounting for attenuation through the couch may also contribute to the discrepancy. When both Monte Carlo codes are compared, as in the Lucy irradiation case, 100% of the pixels pass the gamma analysis. The extremely good agreement between the Monte Carlo codes demonstrates that the iPlan Monte Carlo dose calculation algorithm can accurately predict dose for a clinically realistic plan in a realistic patient geometry. Figure 4.48 depicts profile comparisons between the Monte Carlo codes and film measurements. Similar to the Gamma analyses, the profiles demonstrate that the agreement is not ideal. However, Figure 4.49 demonstrates the agreement between the Monte Carlo codes. The dose profiles from the two Monte Carlo codes agree so well, their profiles are indistinguishable.

![Figure 4.47: Three gamma analyses for the Rando phantom irradiation, comparing the BEAMnrc model of the Novalis Tx to film (left), iPlan Monte Carlo to film (center) and the BEAMnrc model with iPlan (right). Comparing the calculation using the BEAMnrc model and](image_url)
the EBT-2 film measurement (left), only 80% of the pixels passed the gamma analysis. Comparing iPlan’s Monte Carlo code with film measurements, 86% of pixels passed the gamma analysis. Both comparisons to film show discrepancies outside the planned tumor area. When both Monte Carlo codes were compared, 100% of the pixels passed the gamma analysis. The rightmost figure showing the comparison of the Monte Carlo codes is depicted zoomed out to demonstrate the excellent agreement throughout the whole distribution in the Rando phantom.

Figure 4.48: A profile through the Rando dose distribution, comparing the BEAMnrc model’s calculation with film (left), and iPlan’s MC calculation with the film measurement (right). The thick green line represents the Monte Carlo calculation, while the more noisy light green line represents the film measurement. The blue line centered on zero represents the difference between the film and Monte Carlo dose. Both Monte Carlo codes have similar disagreement with film in-field, in the penumbra regions and around 45.0 mm.
Figure 4.49: A profile through the Rando dose distribution, comparing the Monte Carlo codes. The profiles from the respective Monte Carlo codes are indistinguishable from one another, the agreement is so good. Only minor discrepancies are indicated by the blue difference line in high gradient regions, such as near -20 mm.

4.5 References


9. 3cognition, Film QA, 2007.

5.1 Summary and Conclusions

As part of the commissioning process of the iPlan Monte Carlo dose calculation algorithm, a BEAMnrc model of the Novalis TX radiosurgery unit was created to benchmark the iPlan algorithm. To create the model, a previous BEAMnrc model of a Varian Clinac was modified to include the SRS flattening filter and the HD120 MLC of the Novalis TX. With the introduction of the new flattening filter, the source parameters were characterized according to the procedure set forth by Almberg et al. [1] The energy of the electron beam incident on the target was adjusted to 6.6 MeV to match percent depth dose measurements with an IC-10 ion chamber in water. The full-width at half-max, or FWHM, of the source was determined to be 0.07 cm and 0.08 cm in the cross and in-plane directions, respectively, by comparing 5 x 5 cm² penumbra simulations to EBT-2 film measurements. With 15 x 15 cm² profiles in water at 1.5 cm and 10 cm depths as references, simulations achieved the best fit by setting the angular spread of the source to 1.27 degrees. To model the HD120 MLC, the DYNVMLC component module was reprogrammed to account for the different leaf types present in the HD120 MLC, creating the HDMLC component module. The free parameters of interleaf air gap and leaf density were adjusted to match EBT-2 measured interleaf leakage profiles. They air gap and leaf density were determined to be 0.0047 cm and 18 g/cm³, respectively. To achieve an optimal fit for EBT-2 measured abutting leaf leakage, a 0.004 cm abutting leaf gap was introduced into the simulations. Following the creation of the BEAMnrc model of the Novalis TX, the model was validated by comparing simulations to measurements of output factors, percent depth dose curves, profiles in water at multiple depths for a range of field sizes, and also dose planes of MLC defined fields. The validated model of the Novalis TX was then used to perform comparisons with the iPlan Monte Carlo code. Three way comparisons were drawn between the model created in this work, the iPlan Monte Carlo code and EBT-2 film measurements for dose planes of MLC defined fields in solid water and heterogeneous phantoms, for percent depth dose curves of
square fields incident on heterogeneous slab phantoms, as well as for dose planes of more clinical representative irradiations delivered to Lucy and Rando phantoms.

The modeling of the HD120 MLC with the HDMLC component module was successful, as it was able to produce MLC defined dose distributions of which 99% of the pixels passed a 3%/3 mm Gamma analysis with EBT-2 film measurements. It was also able to simulate the abutting and interleaf leakage. Difficulties arose in the case of interleaf leakage, however, as the model is unable to account for hypothesized minute differences in the leaf dimensions that could result in fluctuations in the leakage. An absolute difference of 0.1% in leakage is observed between simulations and measurements, which is deemed acceptable within the context of this work.

Results from the validation experiments for the BEAMnrc model of the Novalis TX created in this work show that the model agreed within 1.25% for output factors, within 1% of percent depth dose measurements, and within 2% (in-field) or 1 mm (penumbra) of profiles at numerous depths in water. From these results, it can be concluded that Almberg’s source characterization procedure was successfully implemented in this work. Although the introduction of angular spread is dismissed by many authors, including Sheikh-Bagheri and Rogers [2], it was necessary to achieve agreement in this work. The source characterization procedure where the source size is adjusted to match in-air off-axis ratios, was not successful in this work.

Using the validated model to aid in commissioning the iPlan algorithm, the three way comparisons made in the work strengthened confidence in the accuracy of the algorithm. For dose planes of more complex MLC defined fields, over 95% of pixels passed the 3%/3 mm Gamma analyses when both Monte Carlo algorithms were compared to film measurements, and over 98% of the pixels passed when the Monte Carlo algorithms were compared to each other. For percent depth dose curves comparisons in heterogeneous slab phantoms, calculations with the BEAMnrc model of the Novalis followed those of the iPlan dose-to-medium calculations. However, the film measurements performed in these slab phantoms were not entirely appropriate for these comparisons. Although separate simulations of mathematical phantoms with varying voxel sizes elucidated the film measurements, the measured features at boundaries between different media could not have been seen in the calculations, given their 3 mm resolution. Finally, for dose planes within Lucy and Rando phantoms for more clinically realistic irradiations, over 90.5% of pixels passed the 3%/3 mm Gamma analyses when both Monte Carlo algorithms were compared to film measurements, while 100% of the pixels passed when the Monte Carlo algorithms were compared to each other. While more work would need to be done to reconcile differences between the Monte Carlo codes and film measurements, the good
agreement between the iPlan Monte Carlo algorithm and the BEAMnrc model created in this work suggest that the iPlan Monte Carlo algorithm is capable of accurately predicting radiation dose for complex fields in heterogeneous media.

5.2 Limitations of this work

Although a BEAMnrc model of the Novalis TX was used to benchmark the iPlan Monte Carlo code, it was outside the scope of this work to check every detail of the BEAMnrc simulation, given that the model was derived from a previous model of a Varian Clinac. The dimensions of the target, primary collimator, shield, mirror and monitor chamber were not investigated in this work. While validation measurements provide confidence in the BEAMnrc model of the Novalis TX, small dimensional errors may still exist that may not be detectable by the experiments performed in this work.

Concerning the HDMLC component module, it must be reiterated that the CM does not model the outboard leaves of the HD120 MLC. Given that this work focuses on the SRS mode of the Novalis TX, these outboard leaves are never present within the restricted 15 x 15 cm² field size. If the HDMLC component module were to be used in a model of the normal mode (either 6 or 18 MeV) of the Novalis TX, the code would have to be modified to include the outboard leaves. As mentioned in the previous section, there are also problems in explicitly modeling the interleaf leakage, especially through the quarter leaf bank. It is hypothesized that small differences in leaf dimensions introduce fluctuations in the leakage because they change the size of adjacent interleaf gaps.

Considering the commissioning process as a whole, this work is only involved in the final validation step of commissioning the iPlan Monte Carlo dose calculation algorithm. The input data sent to BrainLAB to tailor the Monte Carlo code, and the validity of the assumptions made by the Monte Carlo algorithm have not been checked in this work. In many of the three way comparisons, slabs of bone, lung and solid water are contoured in the iPlan treatment planning system and replaced by average CT numbers in order to eliminate CT artefacts. These structures are then exported to MMCTP to perform calculations using the BEAMnrc model. Therefore, discrepancies may exist in the three way comparisons due to differences in how iPlan and MMCTP deal with contours, and not necessarily from fundamental differences in the algorithms. Furthermore, no investigation into the effect of voxel dimensionality was conducted. This work, therefore, is to be considered as an overall third party validation. In a more thorough commissioning, more verification and validation tests would be performed to evaluate specific aspects of the algorithm.
Due to time limitations of this project, only two clinically realistic comparisons were performed in the Lucy and Rando head phantoms. The reasoning behind benchmarking the iPlan Monte Carlo code with a BEAMnrc model of the Novalis TX was to be able to draw a comparison where measurement is impossible. Unfortunately, patient cases that could have been recalculated using the model created in this work were not done. Furthermore, most of the measurements used in the three way comparisons were not repeated. As a consequence, no comments on reproducibility can be made. The discrepancies for the heterogeneous slab phantom comparisons warrant further investigation. More comparisons of small fields in more realistic lung densities would be required to commission this algorithm for SBRT calculations.

5.3 Future work

Future work would include applying both the iPlan Monte Carlo dose calculation algorithm and the BEAMnrc model of the Novalis TX to many more cases. Patient comparisons, including analysis of dose-volume-histograms of the tumour and organs at risk, would be the next logical step to the validation process presented in this work. Realistic plans such as IMRT and Rapid Arc should be calculated by both algorithms. It would also be informative to perform calculations for a variety of the sites, to ascertain where Monte Carlo dose calculations are needed, and where they are not, to confirm the recommendations made in the AAPM report number 85 on inhomogeneity corrections. [3] More complex phantom testing would also be useful to investigate further the limitations of iPlan’s algorithm. In particular, the size of the voxels for the iPlan dose calculation could be reduced to produce more detailed comparisons at boundaries between the two Monte Carlo algorithms in this work.

An issue not fully investigated in this work was the dose-to-water vs. dose-to-medium debate concerning dose to bone. Kawrakow and Siebers have different methods for converting dose-to-medium to dose-to-water as discussed in section 2.1.4. From the percent depth dose comparisons in heterogeneous slab phantoms, the film measurements made in this work would have agreed more with Kawrakow’s method, but no investigation has been performed. More experiments including patient geometry and heterogeneous slab phantoms would need to be performed to fully ascertain which method is more accurate. It is also unclear whether these methods are even appropriate to apply to EBT-2 film measurements.

Finally, the BEAMnrc model of the Novalis could be further modified to produce models for all the photon modes, not just SRS mode. The flattening filter dimensions for each mode are present in the schematic drawings supplied by Varian Medical Systems, Inc. After the inclusion of the new filters, Almberg’s source characterization procedure would need to be repeated to
determine appropriate source parameters for each photon mode. [1] These models could then be used to perform other comparisons for non-stereotactic cases.

This work represents a novel method of validating a dose calculation algorithm for use in stereotactic body radiation therapy, given the difficulties of taking measurements of small field in heterogeneous conditions. Even though Monte Carlo dose calculation algorithms may be considered the gold standard, they must nevertheless be validated. In this work, the validation of iPlan’s Monte Carlo algorithm was accomplished by benchmarking to an open source, trusted Monte Carlo code system: EGSnrc. Extensive work was performed to produce and validate a model of the Novalis TX’s SRS mode, utilizing a new source characterization procedure and modeling a complex MLC. Comparisons of square fields in water, complex MLC defined fields under various conditions, and percent depth dose curves in heterogeneous media were drawn between the Monte Carlo codes with the aid of ion chamber and film measurements. While there many discrepancies, the calculations made with the iPlan Monte Carlo algorithm agreed with calculations made with the BEAMnrc model. This work, therefore, bolsters confidence in the iPlan Monte Carlo algorithm.

5.4 References


APPENDIX A

Code modifications of the DYNVMLC component module

In the interest of simulating the new HDMLC that is used by the Novalis TX radiosurgery machine, the DYNVMLC component module was modified slightly. To understand the modifications made to DYNVMLC, one needs to understand the original code by Emily Heath. There are two files of interest: DYNVMLC_cm.mortran and DYNVMLC_macros.mortran, and both were modified to produce the HDMLC component module.

(***s indicate the changes to the DYNVMLC code to produce the HDMLC component module.

***Note that the both DYNVMLC files, all instances of “DYNVMLC” have been replaced by “HDMLC.”

DYNVMLC_cm.mortran has the following structure:

-Beginning documentation, which describes the capabilities of the CM, all the input parameters, and provides an example of an input file.

***deleted full leaf description. Added two new target/isocenter leaves.
LEAFTYPE_$HDMLC (1) is a half isocenter leaf, LEAFTYPE_$HDMLC(2) is a quarter target leaf, LEAFTYPE_$HDMLC (3) is a quarter isocenter leaf, and LEAFTYPE_$HDMLC (4) is a half target leaf.

***added new example

-HOWFAR subroutine
- type declarations
- prepares local variables
  - sets region numbers and coordinate system
- boundary crossing check
  - z direction
    - air gap check
      - forward/backward
    - in MLC leaves
      - forward/backward
    - calls to WHERE_AM_I macro and to $DYNVMLC_FIND
- calls $DYNVMLC_FIND to determine region #
- air gap check
- lateral edge of MLC check
-calls $DYNVMLC_MINDISTANCE macro which gives the distance to the nearest boundary along the particle’s direction.
-redefines USTEP if needed

***there are no real changes to the HOWFAR subroutine in DYNVMLC_cm.mortran. All the necessary changes are in the macros file, where DYNVMLC_FIND is located. The WHERE_AM_I macro calls the DYNVMLC_FIND macro.

-WHERE_AM_I subroutine
Whenever a particle is transported to a boundary, the subroutine WHERE_AM_I is called, which determines which component module the particle is in. It then calls the WHERE_AM_I macro specific to the CM, ie, WHERE_AM_I_$DYNVMLC, to determine the region number the particle is in.
-type declarations
-defining coordinate system
-particle entering front face/ back face
-both call $DYNVMLC_FIND to determine region #

***No changes here. All changes are located in the macros file specific to the DYNVMLC_FIND macro.

-INPUT_$DYNVMLC subroutine
INPUT macro parses through input file, performs sanity checks and defines variables used elsewhere in both mortran files.
-type declarations
-initializing parameters
-title
-MLC orientation
-does region #’s (1=airgap, 2=leaves, 3=driving screw hole)
-distance to the reference plane
-geometric information
-leaf thickness
-Input geometries for FULL, ISO and TARGET leaf types
-each has its own sanity checks

***deleted full leaf information, and added two new target/isocenter leaves. See note above in the HOWFAR description for what each leaf type number refers to. Note that for the new half leaves, two new variables are created that are defined in DYNVMLC_macros.mortran at the beginning: type declarations. The sanity checks for the new leaves are identical to the corresponding quarter leaves.
-sanity checks between leaf types
***Checks tongue/groove will fit for both quarter and target leaves, as well as between half and quarter leaves.

-parses through each group of leaves, defining leaf type for each leaf

***If LEAFTYPE = 1, we have half leaves, if LEAFTYPE=2, we have quarter leaves. Note the difference between LEAFTYPE, which represents the group types from 1 to 2, and LEAFTYPE_$HDMLC, which represents the individual leaf types from 1 to 4./

-target/isocenter groups must have even # of leaves

***Repeated check for even number of leaves for half leaf types.

-sanity checks for amount of leaves
-start position of leaves
-interleaf air gap

-sanity checks involving tongue and groove + interleaf gap

***To ensure that the interleaf gap is smaller than the tongue, replaced full leaf check with half isocenter leaf check and added a new check for half target leaf. To ensure that GROOVE+LEAFGAP-TONGUE > 0,

-type of leaf ends
-sanity checks for focused/curved leaf ends

***Removed check corresponding to FULL leaf and added two new ones for half leaves. The checks utilized the HOLEPOS variables specific to each individual leaf type.

Z-AXIS PARAMETERS

-defining ZREG_${DYNVMLC}(leaf number, 1-10) for each leaf type
-ZREG is a 2D array that associates the Z inputs for each leaf type to specific leaves
-defines HOLEPOS, the distance from the end of the driving screw hole to the leaf end

***Removed FULL leaf and added two new half leaf types, defined the same way as the corresponding quarter leaves, with the exception of new HOLEPOS variables for the half leaves.

PERPENDICULAR TO LEAF ORIENTATION PARAMETERS
- defining YREG_\$DYNVMLC(leaf number, 1-7) for each leaf type
- YREG is a 2D array that associates leaf thicknesses input to specific leaves
- starts from START_\$DYNVMLC, and loops over every leaf...

***Modified IF statement to include half leaves when defining
YREG_\$HDMLC(I,1), where I is the leaf number. For the other YREGs, I simply
repeated the code for the target/isocenter leaves to account for the new half
leaves.

-defining SURPARA1_\$DYNVMLC(leaf number, 1-7) which is a parameter used to
scale the thicknesses from ZMIN to their appropriate z positions. It is defined as:
SURPARA1_\$DYNVMLC = YREG_\$DYNVMLC / (ZMIN – ZFOCUS)
-sanity check to see if CM goes beyond RMAX

LEAF COORDINATES

- dynamic or step and shoot mode
- reads individual leaf positions
- sanity checks for leaf positions
- defines leaf positions
- defines SURPARA2 which is used for focused leaf ends

-ZFRONT
-ZMAX
- ECUT, PCUT, dose scoring and material in each region
- set up air gap to previous CM
- set up region numbers
- establish CM boundary
- dose scoring zone

- sets all regions to air
- set leaf regions to leaf medium
- sets driving screw holes to air

***Modified IF statements to account for half leaves when setting regions to
certain media.

-ISUMRY_\$DYNVMLC subroutine
Summarizes input, writes graphic file for EGS_Windows and sets parameters that
require medium information for HATCH call.
- type declarations
-calculates mass of dose zone

***Removed the “DATA” arrays for full leaves, and simply utilized the same arrays used for the quarter leaves for the half leaves

-writes geometric information in list file

***When listing leaf dimensions, simply changed “3” to “4” to list dimensions for all four leaf types. Also, added HOLEPOS variables associated with the half leaves: HOLEPOS_HTAR_$HDMLC and HOLEPOS_HISO_$HDMLC. Modified listing format to simply state the existence of four leaves. Removed FULL leaf info.

-output graphics file if requested

-HOWNEAR_$DYNVMLC subroutine
Calculates minimum perpendicular distance to a boundary. Ultimately used to determine how much to call HOWFAR.

- types declarations
- define coordinate system
- using SURPARA array, determines which leaf the particle is in
  - determines the Y and Z region of the particle
  - checks if in groove of a leaf and moves to adjacent leaf

***Changed conditional to include half leaves. Removed FULL leaf info.

- if moved to air gap, move back to previous leaf

***removed FULL leaf information. Changed conditional to include both types of target leaves after coming from isocenter leaf.

- recheck Z, Y regions
- determines which X region we are in
- sets min and max Z boundaries
  - determines min distance to those boundaries
- sets min and max Y boundaries
  - determines min distance to them
- sets min and max X boundaries

***Changed conditional to include half leaves for Z and Y cases. No change for X direction. Removed FULL leaf info.

- determines overall minimum distance

DYNVMLC_macros.mortran has the following structure:
-Beginning documentation, which describes all the variables used
-miscellaneous replacement macros, such as max number of leaves, number of fields...
-DYNVMLC COMMONs
-type declarations

***added new variables: HOLEPOS_HTAR_$HDMLC, and
HOLEPOS_HISO_$HDMLC. In $REAL variables, changes the number of leaf types
from 3 to 4. Removed FULL leaf info.

-macro called upon each history to determine the field number and set the
opening coordinates of the leaves

$DYNVMLC_FIND(region, distance)
Macro used to determine the local region number
-determines which leaf the particle is in
-determine Z (using ZREGs), Y (using SURPARA1) and X region of particle

***Changed conditional to include half leaves for Z and Y cases. No change for X
direction. Removed FULL leaf info.

$DYNVMLC_MINDISTANCE
Macro that will give the nearest distance the particle can travel along its given
direction before it strikes any boundary. Called in the HOWFAR subroutine
-like in the HOWNEAR subroutine, the min and max Z boundaries are set up
according to region numbers (using ZREGs)
-using z direction cosine: W(NP) to determine ZDIST

-like in the HOWNEAR subroutine, the min and max Y boundaries are set using
SURPARA1
-using y direction cosine UVL(1) to determine YDIST

***Changed conditional to include half leaves. Removed FULL leaf info.

-for a given X region, determines min distance (dependant on region number)
-XDIST calculated with x direction cosine: UVL(2)

-compares Z, Y, and XDIST to find minimum distance
Bibliography

*Note that the square brackets at the end each reference denote the pages cited where applicable.


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