TESTING THE EFFECT OF DIFFERENT CALCULATION GRIDS ON THE ACCURACY OF INTENSITY MODULATED RADIATION THERAPY PLANS

A RESEARCH PAPER (3 SEMESTER HOURS) SUBMITTED TO THE GRADUATE SCHOOL IN PARTIAL FULFILLMENT OF THE REQUIREMENTS FOR THE DEGREE MASTER OF ARTS BY JUSTIN D. GAGNEUR

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Abstract

The goal of treatment planning for cancer radiation therapy is to adequately treat the patient’s tumor volume while minimizing dose to the surrounding healthy tissue. In the case of intensity modulated radiation therapy (IMRT), this is accomplished by using complex computer models to calculate the dose to a given volume. These volumes and their resolution are defined by the calculation grid, which defines the space where the dose calculation models are applied and the resolution of that space. The resolution is determined by the size of the voxels (a 3-D pixel). IMRT procedures require a great deal of accuracy and much work has been done at Ball Memorial Hospital (BMH) to verify the accuracy of these treatments. However, the calculation grid has been generally left at a default value to minimize the amount of time that the treatment planning system needs to perform the dose calculations. The intent of the project is, therefore, to test the effect of very fine calculation grid resolutions on the accuracy of IMRT plans.
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Chapter 1

Introduction

The use of ionizing radiation for the treatment of cancer is a relatively new concept in the history of medicine dating back to the dawn of the twentieth century. The first report of radiation therapy to cure a patient came in 1899. Unfortunately for patients being treated with radiation therapy during this time, little was understood about the biological effects of ionizing radiation and why it sometimes cured cancer. Therefore, more harm than good generally came out of those treatments. Throughout the twentieth century though, volumes worth of research and experimentation have been performed to provide a strong basis for its therapeutic relevance today. Ionizing radiation may be either electromagnetic, such as X-rays and gamma rays, or particulate, such as electrons, protons, neutrons. X-rays and electrons are used most often in radiation therapy although there are a limited number of proton and neutron facilities around the world. The goal of radiation therapy is not to destroy the cancerous cell outright, but, cause enough damage to the DNA so that the cells die during mitosis. This is referred to as a mitotic death (Hall, 1994). Ionizing radiation acts to break the chemical bonds found within DNA strands. By causing enough breaks along the DNA, the cell loses its ability to replicate properly and dies while attempting to undergo mitotic division, hence a mitotic death.
This is beneficial for the patient because the amount of radiation needed to cause these breaks is significantly less than what is needed to destroy the cell outright. The goal of treatment planning software is to create a plan that accurately delivers the prescribed dose to the targeted tumor volume while sparing the surrounding healthy anatomy. That is, however, the perfect world and healthy surrounding tissue will always take some fraction of the prescribed dose. In all but the most superficial cancers the radiation delivered to the patient from an external source will go through healthy tissue to reach the prescribed tumor volume. The goal, nevertheless, is to limit the dose to healthy anatomy to reduce the probability that the patient will suffer any side effects due to the radiation therapy.

As would be expected, accuracy is paramount when designing and implementing a radiation therapy plan. The obvious problem is how to accurately image the patient. This includes the normal healthy anatomy and the tumor volume. Today there are various options available to doctors and physicists to achieve this goal. The current gold standard for imaging in radiation oncology is the Computed Tomography (CT) scanner. The images acquired from the CT scanner, however, do not always offer the best quality for certain types of tumors. When this is the case, a patient can undergo a Positron Emission Tomography (PET) or a Magnetic Resonance Imaging (MRI) scan. The PET uses a molecule that is very similar to glucose that is tagged with a positron emitter. Since cancerous cells have a much higher metabolic rate than normal tissue, they uptake more of the molecule than surrounding tissue making it “light up” during the scan. MRI involves using an extremely strong magnet (5000 to 30000 gauss) to polarize the hydrogen atoms in the body. These polarized atoms have a resonant frequency. A Radio Frequency signal at the resonant frequency is shot into the patient. When the atoms
depolarize the RF energy from the initial signal is released and recorded by a very sensitive sensor. This energy fluency map creates the image (Haacke, 1999). The images from these other two modalities can be fused to the original CT data set. This fusion creates a data set that is superior in quality to either of the separate parts. Now that an accurate set of images has been obtained the treatment planning process can begin. Since each patient is unique, every plan is also unique. Consequently, much of the planning process involves some trial and error to optimize dose coverage within the targeted volume while minimizing exposure of healthy tissue.

The model that the treatment planning system uses to calculate dose are created from beam measurements, implying that accurate dosimetry is essential for building a viable model. These models are used in every facet of the dose calculations, and mistakes or errors in the model would become evident very quickly during the patient Quality Assurance (QA) process or, more seriously, if a patient undergoing treatment suffered from acute side effects related to radiation. Therefore, the validity of these models is rigorously tested to ensure patient safety. The resolution that these models are applied at, however, is generally set at a standard value and not changed. This is because very high resolution calculation grids take and an enormous amount of time to calculate the dose compared to the standard resolution grids.

The plans produced using the standard calculation grid are considered to be favorable. Because of this, no tests have been performed to ascertain whether or not a finer calculation grid would produce a significantly better plan to warrant its use even with the added time needed for the calculations. The intent of this project is to test the effect of different calculation grids on the accuracy of Intensity Modulated Radiation
Therapy. To do so, plans for multiple anatomical sites will need to be created, and the dose will need to be calculated using the standard and multiple fine resolution calculation grids. In this paper, recommendations will be made on when it is useful and viable to use a finer calculation grid based on the measured dose data, plan data, and the time that the computer took to perform the dose calculations for each plan.
Chapter 2

Intensity Modulated Radiation Therapy Planning

Treatment planning systems are one of the single most important machines in the modern radiation oncology department. Intensity Modulated Radiation Therapy (IMRT) plans involve extremely accurate dose calculations and planning. Ideally human patient data would be used as the basis for the creation for the plan. Unfortunately due to complications with the Ball Memorial Hospital (BMH) Institutional Review Board (IRB) and the Ball State University IRB, using patient data was not feasible for this project. However, BMH does have a number of anthropomorphic phantoms that approximate the human body very well.

Figure 2.1: Acrylic Pelvis Phantom with imbedded bony anatomy used for this study.
The initial difficulty with using phantoms, however, is that they only have bony anatomy and not the various organs that a human patient would have inherently. The solution to this problem is to use a reference manual that has anatomy identified in various CT slices. By using this manual and the bony anatomy imbedded in the phantom, it was possible to accurately contour the organs using the CMS XiO treatment planning system (CMS, 2006). Once these contours were complete, XiO could accurately calculate the dose given to those contoured volumes, whether they are Organs At Risk (OAR) or targets. In this chapter the techniques used to create the IMRT plans will be examined in detail.

2.1 Dose Calculation Methods

The dose calculation model at the core of the treatment planning system is what drives the entire operation. Without an accurate, rigorous, and reliable model, radiation therapy can easily cause more harm than good. Currently there are three prevailing algorithms that can be used in radiation therapy treatment planning: the Modified Clarkson Sector Integration Algorithm, Fast-Fourier Transform Convolution, and Multigrid Superposition. These three algorithms are used in most treatment planning systems because they are considered to be rigorous and reliable. Ball Memorial Hospital uses CMS XiO, which is capable of using all three algorithms, and it was used to create all of the plans for this study.

First is the Modified Clarkson Sector Integration Algorithm. The Clarkson algorithm is by far the oldest of the three and has begun to show its age when compared to the Convolution and Superposition algorithms. This is because the Clarkson was developed before the massive computing power that we take for granted, was available to
model the dose distributions. In essence the Clarkson Algorithm uses real world measured data, instead of modeling the physics, to predict the dose distributions. Therefore, the Clarkson will not accurately model the X-ray scatter throughout the patient or scatter due to intensity modifiers in the path of the incident beam. Also, it will not accurately model dose when the patient data are missing. These obvious shortcomings and the availability of the Convolution and Superposition Algorithms have led to the very limited use of the Clarkson Algorithm (Butts, 2000).

The Convolution and the Superposition algorithms are similar in that they both compute the dose by convolving the total energy released in the patient with Monte Carlo-generated energy deposition kernels computed by Mackie (1988). The Fast Fourier Transform (FFT) convolution and the Superposition algorithms are both commissioned using a similar process. They both use the same beam parameters and the same Total Energy Released per unit Mass or TERMA data. The TERMA data allow the treatment planning program to calculate very accurately the amount of energy released into the patient. Without this accurately calculating dose would be exceedingly difficult. A number of other measurements are taken to ensure the accuracy and reliability of the calculation. For a full listing of these measurements and steps please see Appendix A.

At this point all off the steps have been common for both FFT convolution and Superposition. For FFT convolution the data are transformed into a Fourier frequency domain that the Fast Fourier Transform can be used on. For Superposition there is no Fourier transformation done, rather, a multigrid approach is taken (CMS, 2006).

The Convolution algorithm is much faster than the Superposition algorithm, but it does not calculate dose as accurately in the presence of tissue inhomogeneities. This
does not necessarily mean that Superposition is better than Convolution. It simply means that Convolution should be used in certain circumstances that play to its strengths. The Convolution algorithm is useful for beam commissioning and for other computations in homogeneous media. In heterogeneous media, however, it may be less accurate than the older Clarkson algorithm. For patient calculations in which tissue inhomogeneities are present, the user should avoid Convolution and use the Superposition algorithm instead (Khan, 1994).

2.2 IMRT Plan Creation

In its loosest definition, IMRT has been employed since the beginning of radiation therapy. Simply putting a block of metal in between the X-ray source and the patient modulates the intensity of the incident beam. Today, however, IMRT is considered to be modulation of the X-ray beams by using a Multi-Leaf Collimator (MLC) or a solid compensator. MLC involves interlocked tungsten leaves that move during the treatment to properly attenuate the beam. Solid compensators are specially milled metal pieces that sit in between the patient and the X-ray source and attenuate the incident beam according to the treatment plan.

Figure 2.2: Examples of MLC (left) and Solid Compensators (right).
IMRT plans are extremely complex and would be very difficult for a person to replicate in any feasible amount of time. This is because the treatment planning system uses an inverse planning scheme. In 3-D conformal planning, beams are placed and manipulated to try to achieve the prescribed doses. IMRT, however, asks for dose constraints and prescriptions to be inputted. These values can then be assigned a power ranging from two to three. These power values are on a logarithmic scale so small input changes represent a large change when it is applied to the plan. The higher the power value the harder the treatment planning program will try to meet that constraint. It is possible, unfortunately, to set the power setting high enough that the program will essentially ignore all other parameters. This makes the plan unusable so the highest the power option is generally set is 2.3 to 2.5.

![Figure 2.3: The menu in CMS XiO that controls the IMRT loop setup.](image)

Once the constraints have been inputted the optimization process can begin. The convergence criteria for the optimization loop can be altered at this point. This will affect at what point the loop will decide that it is close enough to the correct answer to stop, i.e.,
it has met its convergence criteria. The loop runs until it meets its convergence criteria. When the loop stops and the final dose calculations are performed, Radiation Oncology staff can review the isodose lines and the dose volume histogram to determine if the plan meets the criterion. The inverse planning process is not perfect and will take multiple refinements to the loop to achieve the desired dose coverage of the tumor all the while sparing the healthy surrounding tissue. As shown below, with the initial run on the left and the final run on the right, dramatic changes can be achieved by refining the constraints on the IMRT process.

Figure 2.4: On the left is the initial run. On the right is the final optimized run

The point of this study, however, is to test the effect that the different calculation grids have on the dose. Therefore, plans must be run for each and every calculation grid that is to be tested. Once the plan is deemed satisfactory, the calculation grid must be changed and the plan rerun. This involves restarting the optimization loop once the new calculation grid has been inputted into the treatment planning system. If a finer grid was chosen, then the computer will have to perform many more calculations; this is because
the treatment planning system uses the calculation grid to define a three-dimensional pixel or voxel. Dose calculations are performed on each voxel, therefore, the finer the calculation grid the more calculations that will be performed. As a consequence, the times needed to perform the dose calculations are much greater for a finer grid than a coarser one. Whether or not the increase in time for dose calculations is clinically viable will be discussed later in this paper.
Chapter 3

Delivery and Measurement of the Plan

Creation of the treatment plan in CMS XiO is a preliminary and essential step to this project, but without the delivery and accurate measurement of that planned data, no conclusions can be drawn about the effect that a finer calculation grid would have on the plan. There are many ways to accurately measure the dose which is outputted by a LINAC using the plan prior to delivering the dose to a patient. First are ion chambers, which come in both the Farmer type cylindrical chamber and the parallel plate type chamber. When properly calibrated these chambers provide very accurate dose measurements for a certain point in the treatment field.

Figure 3.1: Examples of ion chambers. On the right a parallel plate ion chamber. On the left a Farmer ion chamber.
However, they do not provide any information about the dose in any other part of the treatment field. Therefore, using ion chambers for this particular set of measurements would be very tedious and yield minimal information. Film that is specially designed to be exposed to high energy X-rays could also potentially be used. With film the big picture is taken into account and once it is developed the film can be analyzed with a film scanner. A computer program that can interpret the scanned film can then determine what dose was delivered to any part of the film. There are however major drawbacks to working with film. First, every piece of film is slightly different than the last; this small variability must be taken into account when propagating errors and is often difficult to determine. Second, film inherently has noise associated with it. Whether through small defects in the film or effects that take place during the developing process, there is always noise associated with film measurements. This noise affects the accuracy of dose measurements that can be taken with film. Since a high degree of accuracy is needed for this project using film to take the measurements is not desirable.

3.1 Mapcheck

The BMH Cancer Center however does have a device that is specifically designed to fill this particular niche of seeing the big picture and being able to have very accurate dose measurements. The Sun Nuclear Mapcheck device is an array of hardened diodes that can measure up to a 22-cm x 22-cm field size.
This device has 445 diodes with 221 diodes in the critical 10-cm x 10-cm center area. All of the treatment fields to be delivered are in the 10-cm x 10-cm area thereby using the maximum density of diodes. The hardened diodes are capable of detecting single photons that are incident on the array. The amount of photon flux incident on the array can be determined relatively simply with some mathematical manipulation.

\[
\text{Dose Rate} \quad 600 \frac{cGy}{\text{min}} = 6.0 \frac{Gy}{\text{min}} = 6.0 \frac{J}{\text{min}} = 0.10 \frac{J}{\text{sec}}
\]

2.0 MeV

\[
\bar{E} = 2.0 \text{MeV}
\]

\[
\therefore 1.25 \times 10^{13} \frac{MeV}{J} \times 0.10 \frac{J}{\text{sec}} = 1.25 \times 10^{12} \frac{MeV}{\text{sec}}
\]

\[
1.25 \times 10^{12} \frac{MeV}{\text{sec}} = 6.25 \times 10^{11} \frac{\text{photon}}{\text{sec}}
\]
As shown above the number of incident photons to the detector array is enormous. Even at a diode efficiency of 1 percent there are 6.25 billion incident photons per second that are incident and detectable on the array at the assumed dose rate.

According to Létourneau et al. and Ahluwalia et al. the Mapcheck dose response is very linear as well. Up to 295 centi-Grey (cGy) the response is very linear; this allows for a very predictable and robust measurement.

![Graph showing dose response](image)

Figure 3.3: From IMRT QA with 2-dimensional Diode Array of Detectors (Ahluwalia et al.)
For fields that would exceed this 295-cGy threshold, the effects should be taken into account and using the Mapcheck device for those fields exceeding 295 cGy may be a contraindication. The fields used in this project are well below this threshold ranging from 50 to 200 cGy.

When using diodes temperature effects must also be taken into account. The temperature of the device relative to the room around it can affect the measured dose. This can vary as much as 0.57 % per degree Centigrade (Létourneau, 2003). To get rid of this factor multiple precautions can be taken. Store the Mapcheck in the same room as which the measurements will take place. If this cannot be done because of concerns over the safety of the device or the room is not secure, the Mapcheck should be allowed to stabilize to room temperature before measurements are taken. If all of these factors are taken into account, then the Mapcheck will provide extremely accurate and reproducible results that can then be compared to the plan data.

3.2 Delivery and Measurement

Delivery of the desired fields is a relatively simple task when using the Mapcheck. Certain factors much be taken into account, however, to achieve the desired accuracy with the measurements. First of all the LINAC does not have a constant output. The output is allowed to vary up to one percent from the yearly calibration. The treatment planning system assumes a constant output, however, so the machine output must be determined before any measurements can be taken to take the machine variability out of the equation. To do this multiple measurements are taken with a parallel plate ion chamber at varying depths of build up. These readings are compared to the annual calibration to get the output factor. This factor can then be applied to
minimize the error involved with the system. It is important to note that the output is heavily dependent on the temperature and pressure of the environment. If either of these two change significantly, then the output be retaken for the remaining measurements or they should be allowed to stabilize back to their original numbers. With the output measured the next step is to setup the Mapcheck and allow it to measure the background radiation in the room. This step allows the software package to subtract out the background radiation from the actual measurement when it is taken. The Mapcheck is attached to an isocentric mount that then attaches to the LINAC treatment head.

Figure 3.4: The Isocentric mount used at Ball Memorial Hospital for QA.
By using the isocentric mount the Mapcheck is in the exact same position for every measurement. The mount also keeps the Mapcheck from shifting during actual delivery of the field eliminating any error due to small movements of the device. Once all of these setup steps are complete, the delivery of the desired field can begin. The Mapcheck constantly updates the displayed information during the delivery of the field. This allows for spot checks that the correct field is being delivered. When the LINAC stops, delivery the treatment the measurement can be stopped and saved as a text file for later analysis. At this point you can also make sure that there is nothing grossly incorrect about the field.

Figure 3.5: The finished output for an individual field
The process is repeated until all of the fields are delivered. Depending on the dose rate of the machine, which is how fast the machine can output a certain amount of radiation, and the complexity of the field it can take thirty to ninety seconds to deliver the field to the Mapcheck.
Chapter 4
Evaluation and Results

As described above, the Mapcheck device easily gathers massive amounts of data and compares it in such a way as to make interpretation simple and significant. Therefore, the intent of this chapter is to describe the methodology used to compile and analyze the data acquired from the Mapcheck.

4.1 Distance to Agreement Analysis

Clinically what is looked at during a normal analysis of a treatment field is the overall pass/fail rate for the diodes in the array. To determine whether a field has passed or failed, a Distance To Agreement (DTA) analysis is used. DTA entails three inputted factors from the user that is uses as parameters to determine what diodes have failed (Ahluwalia, 2004). First is the percent difference; this value simply determines the limit of how different the delivered field can be from the master field. Next is the distance; this value is a distance in millimeters. Lastly is the threshold value this value sets a threshold for what points are used. It defaults to ten which means that a data point lower than ten percent of the maximum is not used in the analysis. For the purpose of this study the threshold value was kept at the clinical level of ten percent.
Once these three values have been inputted the DTA analysis can begin. The software analyzes every diode and compares it to a master field, which is acquired from the treatment planning system. At each diode the program draws a circle using the distance input as the radius of the circle. It then compares the measured diode reading to the all of the planned values inside of the drawn circle. If any of the values inside the circle meet the percent difference criteria, then the diode has passed. However, if the criteria are not met inside of the circle, then diode has failed. Diodes which have more dose than planned are colored red, while diodes with less than the planned dose are colored blue.

Figure 4.1: The finished comparison for a field with DTA setting of 2% difference in a 2-mm circle
Clinically DTA setting of 3% difference in a 3-mm circle are used. This, however, is not a rule set in stone and varies from clinic to clinic (Nelms, 2007). Therefore, analysis of all fields where done with DTA settings of 5% difference in a 5-mm circle, 3% difference in a 3-mm circle, and 2% difference in a 2-mm circle.

Obviously the 2% in 2-mm circle is the most rigorous DTA analysis, allowing for the least drift in the delivery of the fields. As an operator one can have any combination of DTA settings. Some are better than others which are why the 5, 3, and 2 DTA settings are considered to be loose guidelines to setting up a QA regiment at a clinic. There is also variability in deciding whether or not the measured field is correct and therefore ready to be treated. The lowest clinically used is a 90 percent to 95 percent pass rate. Ideally the pass rate would always be 100 percent, but, that rarely happens if the DTA analysis was properly restrained with its three inputs and performed correctly.

4.2 Results

With the DTA analysis completed and the results recorded, the overall results can be analyzed in detail. The motivation for this study was to determine if a change in the calculation grid of the treatment planning system would affect overall accuracy of the plan. The easiest way to determine if a delivered plan is more accurate than another plan is to look at the diode pass/fail rates using a DTA analysis. As stated before DTA looks at percent difference and distance. The shorter the distance and the lower the percent difference mean a more accurately delivered field and a higher passing percentage for the diodes. With this in mind, a DTA analysis was done on every delivered field for each calculation grid. Two anatomical sites where used to create the fields. Approximately half of the fields were created in case one, which was a prostate. The other half of the
fields were created in case two, which was a highly involved lung tumor. The reason for this was to determine if the results of different calculation grids were similar over simple (case one) and complex (case two) cases.

The calculation grids can be customized in all three dimensions. The amount of information in these three dimensions is not equal however. The CT scanner takes helical slices through the patient that can vary in thickness. The CT scanner at BMH on a standard scan uses a 2.5-mm slice thickness. Therefore, when the calculation grids were altered, the Z direction was kept at a constant of 2.5 mm and the X and Y directions were altered from 4.0 mm to 1.0 mm in size. This was done to minimize the amount of interpolation that the treatment planning system has to perform. A value for the calculation grid in the Z direction that is not equal to the CT slice thickness means that some amount of interpolation between the slices has to be done. However, it is important to test if altering the Z direction gives the same general trends as keeping the Z direction constant. A quick run was performed for a 1.0 by 1.0 by 1.0 mm grid. There was a slight increase in time from the 1.0 by 1.0 by 2.5 mm grid. Going from 795 to 894 seconds in time needed to perform one iterative run. The pass rates for a DTA setting of 2% at 2 mm were nearly identical however. The original 1.0- by 1.0- by 2.5-mm grid had an average pass rate of 95.18 while the tested 1.0- by 1.0- by 1.0-mm grid has a pass rate of 95.42. There is an increase in time but no appreciable increase in the pass rate.

As expected the pass rates for 5% at 5 mm in Fig. 4.2 is very high, with the lowest being 97.6 percent passed. The data also show a trend pointing to the superior accuracy of the 2.0-, 1.5-, and 1.0-mm calculation grids. The 4.0- and 2.5-mm calc grids were even at an average 99.02 percent pass rate. On average the 2.0-mm calc grid had a 0.4
percent higher pass rate, while the 1.5-mm calc gird had a 0.7 percent higher pass rate and the 1.0-mm calc grid had a 0.83 percent higher pass rate. These trends become more obvious when looking at the more restrained DTA analyses.

The pass rates for 3% at 3 mm shown in Fig. 4.3 again show trends that point towards the superior accuracy of the 2.0-, 1.5-, and 1.0-mm calc grids. The 4.0- and 2.5-mm calc grids were nearly even with the 4.0 mm calc grid slightly higher at 96.18 percent passed, while the 2.5-mm calc gird was at 95.86 percent passed. On average the 2.0-mm calc grid had a 96.88 percent pass rate, while the 1.5-mm calc gird had a 97.82 percent pass rate and the 1.0-mm calc grid had a 98.08 percent pass rate. The further the DTA analysis is constrained the more the disparity there is between the finer and coarser calculation grid results.
The pass rates for 2% at 2 mm are shown in Fig. 4.4. The 4.0- and 2.5-mm calc grids were again nearly even with the 4.0-mm calc grid slightly higher at 90.13 percent passed, while the 2.5-mm calc grid was at 89.40 percent passed. On average the 2.0-mm calc grid had a 90.98 percent pass rate, while the 1.5-mm calc grid had a 93.89 percent pass rate and the 1.0-mm calc grid had a 95.18-percent pass rate.
In case two there are similar trends in the data. However, for case two only four different calculation grids were used. The reason for this is because the time needed to perform one iteration of calculations was over twenty minutes. This is not clinically viable when time is always an issue. This will be discussed in greater detail later. The pass rates for 5% at 5-mm for case two are shown in Fig. 4.5. The 4.0- and 2.5-mm calc grids were again nearly even with the 4.0-mm calc grid slightly lower at 99.82 percent passed, while the 2.5-mm calc grid was at 99.88 percent passed. On average the 2.0-mm calc grid had a 99.82 percent pass rate, while the 1.5-mm calc grid had a 99.76 percent pass rate.

The pass rates for 3% at 3 mm for case two are shown in Fig. 4.6. The 4.0- and 2.5-mm calc grids were again nearly even with the 4.0-mm calc grid slightly higher at 98.94 percent passed, while the 2.5-mm calc grid was at 98.77 percent passed. On average the
2.0-mm calc grid had a 99.01 percent pass rate, while the 1.5-mm calc grid had a 98.94 percent pass rate.

The pass rates for 2% at 2 mm for case two are shown in Fig. 4.7. The 4.0- and 2.5-mm calc grids were again nearly even with the 4.0-mm calc grid slightly higher at 95.32 percent passed, while the 2.5-mm calc grid was at 95.20 percent passed. On average the 2.0-mm calc grid had a 96.73 percent pass rate, while the 1.5-mm calc grid had a 96.52 percent pass rate.
One of the major portions of this study was to determine whether or not finer calculation grids were feasible to use with respect to time issues. Looking at the pass rates the finer calculation grids obviously perform better than the coarser ones. Now the time aspect must be taken into account as well. Figure 4.7 shows that time does indeed play a large factor in deciding which calculation grid to use. The physics staff cannot afford to wait large amounts of time for a plan to finish if they do not have to. As shown in the table the amount of time taken by successively finer calculation grids is not linear. The 4.0- and 2.5-mm calculation grids are comparable in time used and results gained. When moving from 2.5- to a 2.0-mm calculation grid, however, there is a significant jump in time. On average however approximately 1 percent is gained on the pass fail ratio. Moving still to the finer grids of 1.5 and 1.0 mm the time needed dramatically increases. In case one the increase in the pass/fail ratio is quite large 3.76 and 5.05 percent. This is
offset, however, by a large increase in time needed. Case two offers opposite results with a slight 0.21 percent decrease in the pass/fail ratio for a cost of 393 seconds.

<table>
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<th>Calc Grid</th>
<th>TIME(s)</th>
<th>DTA 2% at 2mm</th>
<th>Calc Grid</th>
<th>TIME(s)</th>
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<td>210</td>
<td>90.98</td>
<td>2.0</td>
<td>484</td>
<td>96.73</td>
</tr>
<tr>
<td>1.5</td>
<td>382</td>
<td>93.89</td>
<td>1.5</td>
<td>877</td>
<td>96.52</td>
</tr>
<tr>
<td>1.0</td>
<td>795</td>
<td>95.18</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Figure 4.8: Time need to complete one iterative loop compared to the pass rate at DTA settings of 2% at 2 mm

Up until this point all of the test cases have been using high energy photons. It was assumed early on in the study that low energy (6 MV) photons and high energy (18 MV) photons would behave in a similar fashion when the different calculation grids were used. This however was unsubstantiated and is an important question to answer. Therefore a four-field test case was devised to test the validity of the assumption. Figure 4.8 shows that the behavior in the low energy case is identical to that of the high energy case. Therefore, the assumption made early in the study appears to be valid.

Figure 4.9: Low Energy (6 MV) photons test case DTA settings 2% at 2 mm
Finally, after speaking with various physicians at BMH, the question was brought up of the change in dose per field for the different calculation grids. This is an interesting question and one that is very important to address because of the possibility to deliver less dose to healthy surrounding tissue. The treatment planning system has tools which easily display this information. As shown in Fig. 4.9 there is a slight change in overall dose for case one and only a 1-cGy change in dose for case two. The case two results are interesting. They show the same total dose but a significant restructuring of how much dose each field is assigned.

<table>
<thead>
<tr>
<th>Units are in CentiGray (cGy)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Calculation Grid Case one</td>
</tr>
<tr>
<td>Field #</td>
</tr>
<tr>
<td>1</td>
</tr>
<tr>
<td>2</td>
</tr>
<tr>
<td>3</td>
</tr>
<tr>
<td>4</td>
</tr>
<tr>
<td>5</td>
</tr>
<tr>
<td>6</td>
</tr>
<tr>
<td>7</td>
</tr>
<tr>
<td>Total Dose</td>
</tr>
</tbody>
</table>

Figure 4.10: Dose given for each field in each case for various calculation grids

This happened because the treatment planning system decided that for the finer calculation grids the dose should be spread out differently than the coarser calculation grids so that it could achieve the convergence criteria. Given this information there is not a significant change in dose when using a finer or coarser calculation grid.
Chapter 5

Summary and Conclusion

It has been determined that calculation grids can dramatically affect the output of a radiation therapy treatment plan from the CMS XiO treatment planning system. This was determined by using the Sun Nuclear Mapcheck device and its supporting analytical software. It was determined that the standard 2.5-mm calculation grid performs nearly identically to the 4.0-mm calculation grid, while finer grids have up to a 5 percent higher diode pass/fail ratio. The pass/fail ratios were not the only determining factor to decide whether or not a calculation grid was superior to the standard 2.5-mm grid.

The non-linear time response to the use of finer calculation grids it is not clinically feasible to use a calculation grid lower than 2.0 mm. Because the standard calculation grid offers accurate and acceptable results to start with, waiting an extra 669 seconds per iterative run for an increase of 5 percent is not feasible. Case two results highlight that blanket generalizations cannot be made about the calculation grids affects on the plan. There is an obvious field dependency that limits or enhances the effect of certain grids. This dependency along with the time needed to perform the calculations is reason not to use 1.5- and 1.0-mm calculation grids.
The 2.0-mm calculation grid offers a consistent 1 percent gain over the coarser 4.0- and 2.5-mm calculation grids for a much smaller increase in time needed to perform the calculations. Thus, experimental results suggest that the 2.0-mm calculation grid should be used on complex cases where any increase in accuracy of the dose delivered will benefit the patient.
References


