Influence on different Monte-Carlo variances at the dose calculation and the impact to the commissioning

By

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Thesis

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Declaration

I hereby affirm that I have written this thesis independently and without the use of resources other than those quoted. Cited and copied works are marked as such.

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Jailan Alshaikhi Date

Dedication

To my parents

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1. Introduction

The goal of radiation therapy is to kill tumor cells while reducing the injury to the normal tissues. The physician prescribe the target dose based on the experience gained over several years through research on the radiation dose that is adequate for the control of the particular tumor type, while the physicists ensure that the prescribed dose is delivered to the target with a great degree of accuracy. While small deviations from the prescribed dose may be acceptable, large deviations are not since this can result in either poor tumor control or increase the possibility of the adverse effects of radiation. Source of error in delivering the prescribed dose to the target include error in the positioning of patient and the errors due to the uncertainty of calculated dose. The primary method of delivering the radiation dose to the target volume with external beam radiation therapy is with the use of linear accelerators. The dose that will be achieved in the target from a linear accelerator is often calculated and verified afterwards through measurements before the planned treatment is delivered to the patient. As would be expected, there are always some errors which exist between the calculated dose and the measured dose. The aim however, is to keep such discrepancies to a minimum, and the analysis of discrepancies between the calculated and measured dose is necessary in order to implement programs aimed at reducing such errors.

The desirable characteristics of dose calculation methods in radiation therapy are that the calculation be fast enough so that the treatment planning process can be completed in a clinically acceptable time frame; and secondly that the result of the

dose calculation be sufficiently accurate (Oelfke and Scholz, 2006). Intensity modulated radiation therapy is considered as the state of the art method of achieving greater tumor control with a minimum of normal tissue injury. The intensity-modulated fields are used to deliver highly conformal dose distributions in radiotherapy, which leads to better sparing of the normal tissue. Intensity modulated fields delivered with multi-leaf collimators are routinely used in the treatment of prostate and head and neck cancers and the verification of the dose distribution with this method of radiotherapy treatment poses new challenges for quality assurance (Jenghwa Chang et al., 2000). Some authors have reported discrepancies between calculated and measured dose of greater than 5 percent while other authors have reported even higher values in excess of 10 percent. The uncertainty in the dose calculated by a conventional dose calculation algorithm has been reported to be between 5 to 10 percent in the presence of heterogeneities. Similar error values were also reported for dose calculated using Monte Carlo methods (Ma et al., 2000).

Medical physicists strive to achieve an accuracy of better than 5 percent during the course of delivering the prescribed target dose. This objective can only be achieved if the dose calculation accuracy is better than 2 percent (Fippel, 2006), since there are other contributing sources of dose delivery errors such as patient positioning, motion, etc. As stated by Oekfke and Scholz in their work (Oelfke and Scholz, 2006), the calculation of distribution within the patient form the only reliable and verifiable link between the chosen treatment parameters and the observed clinical outcome for a specified treatment technique. It is thus necessary that the dose

calculation method be accurate. The accuracy can only be quantified comparing the calculated dose with the measured dose. There are many methods of dose calculations. It is the objective of this research work to investigate the accuracy of one of such methods of patient dose calculation, which is gradually being introduced for clinical dose calculation.

1.1 Dose calculation methods

The purpose of dose calculation is to predict the dose at any point in a given medium using information of the makeup of the medium and the dose delivery device. In radiation therapy treatment planning, the dose distribution in a patient for any given beam set up is first calculated/predicted before the treatment is delivered. Computerized tomography images of the patient are usually used to provide information on the material makeup of the patient, and hence the interactions of the treatment beams in the patient, while measurements made in a water phantom usually provide the information used to model the beam delivery device and the quality of the beam. The physician prescribes the dose necessary to control the tumor based on prior experience on the dose which is adequate to control the tumor, it is thus important that the calculated dose distribution be accurate to a high degree, since large deviations may result in adverse effects. Evidence suggests that dose difference of about 7 percent is clinically evident and researchers have also shown that a 5 percent dose difference can result in 10 – 20 percent changes in tumor control probability or up to 20 - 30 percent changes in normal tissue complication probabilities (Indrin et. al., 2007). Another important

desirable feature of any dose computation method for clinical use is that the computation method be fast enough to be clinically feasible. A wide range of methods for computing dose exists with varying degrees of complexities and no general classification consensus (Rosenwald, 2007); however, some authors has classified dose computation as either 'indirect/correction based' or 'direct/ model based' dose calculation methods (Podgorsak, 2005, Fippel, 2006)). The dose calculated by any of these methods is often verified experimentally through measurements before the treatment is delivered (Podgorsak, 2005).

1.1.1 Indirect/ correction based dose computation method

This method of dose computation was the first to be developed (Oelfke and Scholz, 2006). The indirect first methods measure the physical characteristics of the radiation beam such as depth dose, tissue-air ratio, output factors, tissue phantom ratios, etc. in a homogenous water phantom. To calculate the dose distribution in a patient, the dose measurements made in the water phantom are extrapolated and adapted to the patient by correcting for the differences between the makeup of the water phantom and the patient. The corrections that are necessary to adapt or convert the dose distribution in the phantom to that of the patient include corrections for tissue inhomogeneities, since some irradiated tissues in the patient such as the bones and lung have different electron densities that are considerably different from the homogenous electron density of the water phantom; correction for beam modifiers that may be required for the treatment of the patient and

corrections for irregular patient surface (Mackie et al., 2007, Podgorsak, 2005). The correction based methods have the advantage of being fast (Oelfke and Scholz, 2006), the accuracy of this method is however low in the presence of inhomogeneities.

1.1.2 Direct / model based dose computation methods

The direct method of dose computation is a more complex method of predicting the dose distribution within a patient. Unlike the indirect method which measures dose distribution from the beam in a water phantom and then adapt the measurement to the desired medium (the patient), the direct dose computation methods model the interaction and deposition of energy by the beam as it transverses the patient. Although direct method also requires that measurement be made in a phantom, they are used to set the parameters for the model and for verification. This method of dose computation is computationally more expensive relative to the indirect method, however they are more accurate. The model based methods include the pencil beam, collapsed cone and Monte Carlo dose computation methods (Fippel, 2006). The pencil beam is the simplest while the Monte Carlo method is the most complex. A rule of thumb in the dose calculation methods is that the simpler methods are often the fastest while the more complex methods are the most accurate. The method of interest in this work is the Monte Carlo method of dose computation, which will be discussed in better details.

1.2 Monte Carlo dose calculation

The Monte Carlo method is the most complex of the dose calculation methods and also the most accurate and thus has a clear preference relative to other dose calculation methods in the quest for a dose delivery accuracy of 5 percent or better (Fippel, 2006). Research has shown Monte Carlo dose calculation to be particularly accurate in a heterogeneous medium clinically represented by the patient, where other dose calculation methods yield poor results due to failure to accurately model electron transport in such medium and various levels of approximations they employ. This method has however until recently been considered impractical for clinical patient dose calculation due to the often long calculation time required. However, the development of faster computers and Monte Carlo codes has lead to Monte Carlo codes being increasingly clinically available for patient dose calculation. The Monte Carlo method of dose calculation simulates the transport and interactions of photons and electrons as they traverse through a medium by using current physical knowledge of the probability of interactions of individual photons and electrons as they traverse the medium of interest. The kind of interactions simulated for radiation therapy includes photoelectric absorption, Raleigh scattering, Compton scattering and pair production. The macroscopic features (physical manifestation of interactions) of the radiation beam are computed as an average of many simulated interactions of particles or histories. If the true average of the particles' interactions exists and the individual particle interactions has a variance of σ^2 from the average value, then the *Central limit theorem* stipulates that that the estimate of the average

interactions gets closer to the true value as the number of simulated particle histories/ interactions is increased. The theorem also predicts that as the number of simulated histories tends to infinity, the statistical variance tends to zero. The number of simulated particles (histories), N, that has to be directed toward a target volume in a Monte Carlo simulation is approximately given by:

$$N = \frac{A}{\mu\sigma^2\ell^3} \qquad eqn\ 1.1$$

Where A is the exposed beam area, σ is the percent relative error (deviation) being sought, μ is the attenuation coefficient, and ℓ^3 is a typical voxel dimension. Thus for a given field dimension and medium of given attenuation coefficient, the relationship between voxel size and number of particle histories is inverse. The greater the number of histories, the smaller is the uncertainty (Mackie et al., 2007, Nahum, A, 2007, Bielajew, A, 2007, Fippel, 2006).

The efficiency, ε, of Monte Carlo dose calculation is expressed as

$$\varepsilon = \frac{1}{s^2T} \qquad eqn\ 1.2$$

With s and T as the estimate of the variance and computation time required to obtain the variance respectively. There are two ways in which the efficiency of a given Monte Carlo dose calculation dose calculation can be improved: either decrease s^2 (variance) for a given computation time or decrease T (computation time) for a given particle history while not changing the variance. Techniques which improve the efficiency of the dose calculation by changing the variance for a given particle history while not biasing the results are known as variance reduction techniques. Widely used variance reduction techniques include

Bremsstrahlung splitting and Russian roulette. (Indrin et. al., 2007, Kawrakow & Fippel, 2000)

1.3 Beam measurement and verification

Measurement of the calculated dose can be done using ionization chambers, thermoluminescent devices (TLDs), radiochromic or radiographic films, or electronic portal imaging devices (EPIDs) together with specially designed verification phantoms. The measured dose is compared to the calculated dose using verification software. There are many commercially available software that can be used to compare the computed and measured dose distributions. The verification software read in the calculated dose from the treatment planning system and that measured using the measuring device and then analyze both data sets for agreements and quantify the error therein. Standard evaluation tools are the overlay of both isodoses and profiles of the dose data (Rhein and Haring, 2006).

The gamma index is a mathematical tool that enables two dose distributions to be quantitatively compared for similarity and is widely used in IMRT verification software tools that was proposed by Low (Rhein and Haring, 2006, Hrbacek et. al., 2007). According to an article cited in the work by Hrbacek et. al. (Hrbacek et. al., 2007), when gamma evaluation is being performed, one dose distribution is referred to as the reference while the other is referred to as the evaluated. The gamma index is computed for each point of the reference dose distribution using

the entire evaluated dose distribution. The work further stated that the gamma evaluation is not symmetric with two dose distributions and that care should be taken on the dose distribution to be used as a reference as it could have influence the gamma evaluation result.

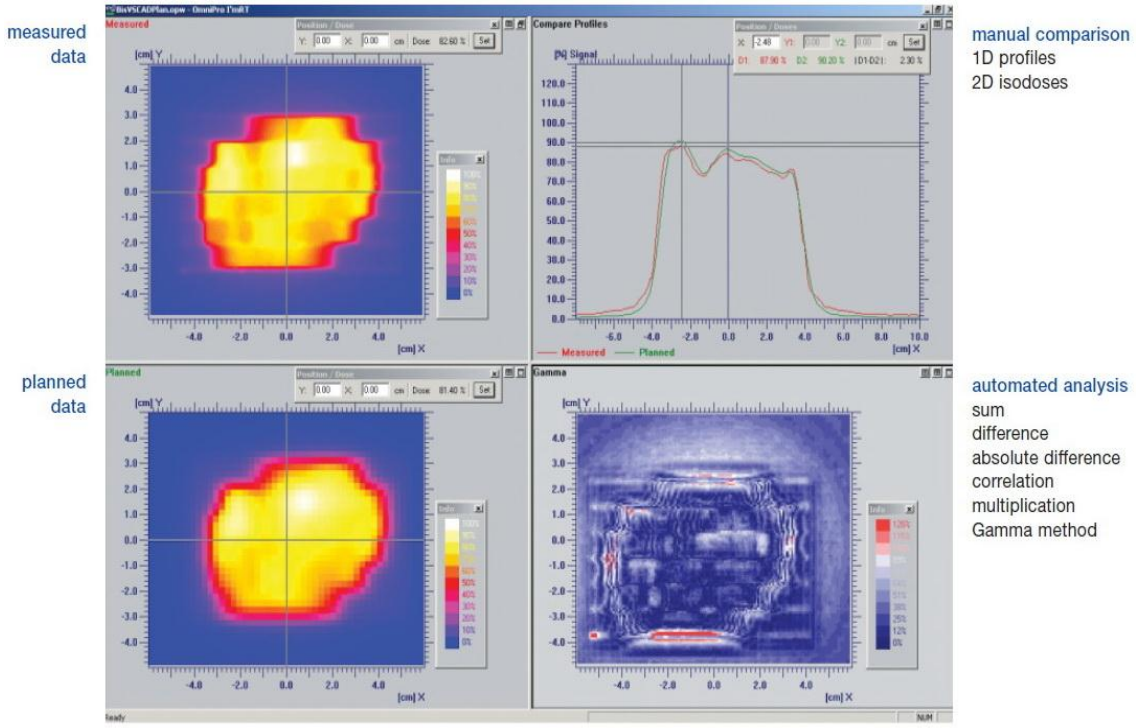


Figure 1: Screenshot of a dose comparison program.

The two images to the left are the measured and calculated doses; the right upper plot contains the profile of the doses while the right lower plot is the gamma comparison image. Red highlights region of disagreement between the two dataset. Source: OmniPro I'mRT documentation

The gamma index is calculated based on a dose difference and distance to agreement criteria and the measured dose data is usually used as the reference. The gamma value, γ, for the measurement point r_m is defined as

$$\gamma = \min[\Gamma(r_m, r_c)] \; \forall \; [r_c] \qquad eqn\ 1.3$$

Where

$$\Gamma(r_m, r_c) = \sqrt{\frac{r^2(r_m, r_c)}{\Delta d^2 M} + \frac{\delta^2(r_m, r_c)}{\Delta D^2 M}} \qquad eqn\ 1.4$$

$$r(r_m, r_c) = |r_c - r_m|$$

and

$$\delta^2(r_m, r_c) = D_c(r_c) - D_m(r_m) \qquad eqn\ 1.5$$

Where r_c is the dose calculated for a given point, r_m is the dose measured for the calculated point which is often taken as the reference dose, ΔdM is the defined passing distance between isodose points (the calculated and the measured), ΔDM is the dose difference value between the calculated and measured which is accepted as pass. For each point in the evaluated data points, the gamma is computed as specified by *equation 1.3* and the evaluation is scored. This scoring could be said to be Boolean and is specified as

$$\gamma(r_m) = \begin{cases} 1, & \gamma \leq 1 \\ 0, & \gamma > 1 \end{cases}$$

Where 1 and 0 specifies passed and failed evaluation points respectively. The gamma calculation is performed for all r_m (Low et al., 1998, Rhein and Haring, 2006). In other words, if the dose calculated and that measured does not agree based on the defined evaluation criteria (distance to agreement and acceptable

dose difference values), the calculation for that point fails and vice versa. The result of gamma analysis is usually presented in terms of the percentage agreement between calculated and measured points. Hrbacek et. al. reported very good gamma evaluation results using the OmniPro I'mRT® software in their work (Hrbacek et. al., 2007), which shows that even at a stringent gamma evaluation criteria of 1 mm and 1 % for the distance to agreement and dose difference values respectively, that 70 % of the evaluated profiles had gamma values of less than or equal to one.

2. Materials and methods

2.1 Dose calculation and delivery

Calculation for dose distribution was made using Monaco treatment planning system (Elekta, Monaco Version 2.03.00). The algorithm employs Monte Carlo simulation for dose computation. Multiple calculations were made using variances (dose calculation uncertainty) of 0.5, 1, 2, 3, 5 and 10%. The Monaco® training guide recommends using variance ranging from 1 – 3% although the full available variance treatment planning system ranges from 0.5% - 10%. Calculations were made for various regular square and rectangular open fields, irregular shaped fields and IMRT clinical cases. The calculation times at different variances for some field shapes and sizes were recorded and analyzed.

The calculated dose was exported to the control software of the linear accelerator, Mosaiq®, and ultimately to the accelerator itself. The linear accelerator is an Elekta Synergy (Elekta Oncology Systems Ltd, Sweden). A beam energy of 6 MeV was used for calculation and the same beam energy was used for the exposure of the dosimetric film. The films used are GAFCHROMIC EBT2 dosimetric films that were exposed in a homogenous solid water phantom. The phantom has 29 slabs each of dimension 30 X 30 X 1 cm and the films were exposed at a chosen depth corresponding to that specified when the treatment plan was exported from the treatment planning system to the machine. 100 cm SSD was used for the film

exposure. Figures 1 and 2 show the linear accelerator and the solid water phantom used for the dosimetric studies respectively.

Figure 2: An Elekta Synergy linear accelerator

2.2 Film calibration, comparison of calculated and measured dose

Prior to exposing the dosimetric films, the films were carefully labeled for proper identification and scanned using EPSON Expression® 10000 XL Photo Scanner and the pre-exposed film images were saved.

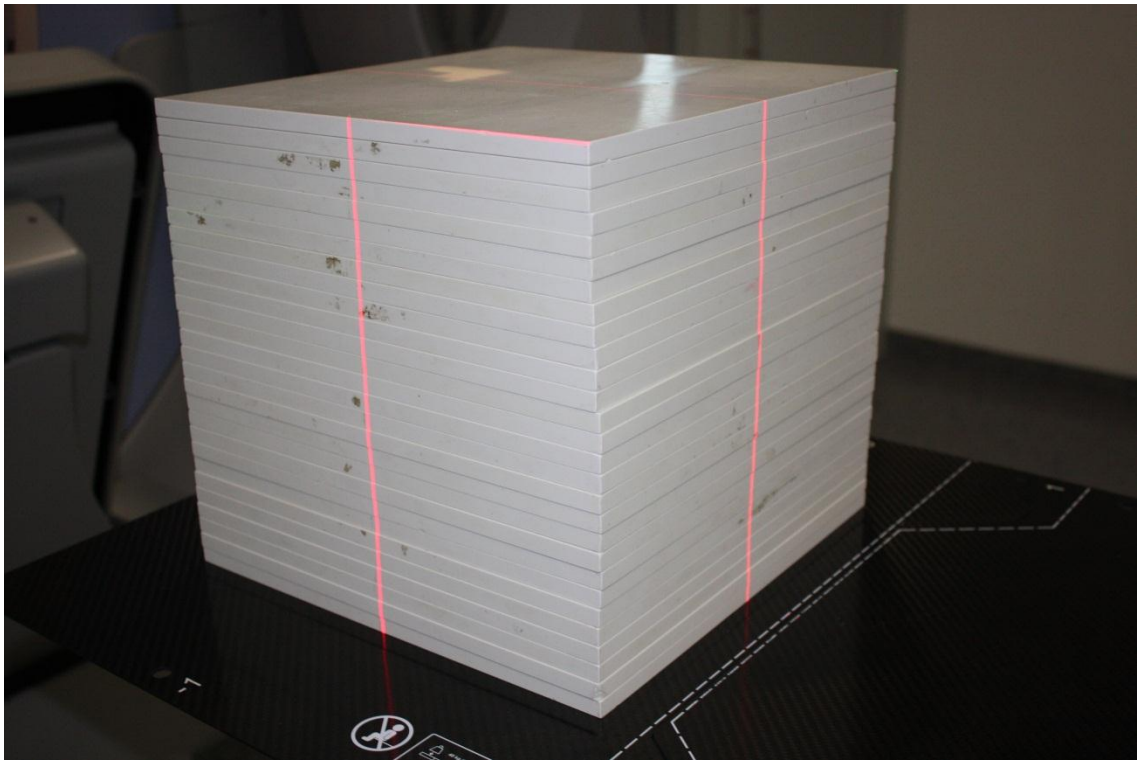


Figure 3: The solid water phantom used for film exposure

The films were exposed in the homogenous water phantom and subsequently processed. The processed films, which contain the measured dosimetric information were scanned using the same scanner settings and format as was used for the scanning of the pre-exposed film. The scanning set up employed also ensured that the pre-exposed and exposed films were scanned using the same

orientations. The difference between the pre-exposed and exposed film was subtracted using an algorithm to give the true value of the optical density – the correct measured dose. The subtraction is necessary because the full optical density of the exposed film also include background density due to film fog and the density of the film materials, such as the film base and emulsion layer. An algorithm was used to convert the measured optical density to dose values. The calculated dose was compared with the measured dose using OmniPro I'mRT® software. The software employs the Gamma evaluation method described in earlier section. Four dose difference (DD) and distance-to-agreement (DTA) values were used for the comparison. The values used are 5 % and 5 mm, 4 % and 3 mm, 3 % and 3mm, and 2 % and 2 mm for the DD and DTA pairs respectively.

3. Result

3.1 Gamma analysis and computation time

Figure 4 shows the calculated dose distribution for square fields of 10 cm^2 using variances of 0.5, 1, 2, 3, 5 and 10%. At low variances, the calculated dose distribution appears smooth with little noise, whereas at higher variances, the distribution appears to contain a lot of noise which appears as increase in the graininess of the image as the variance increases.

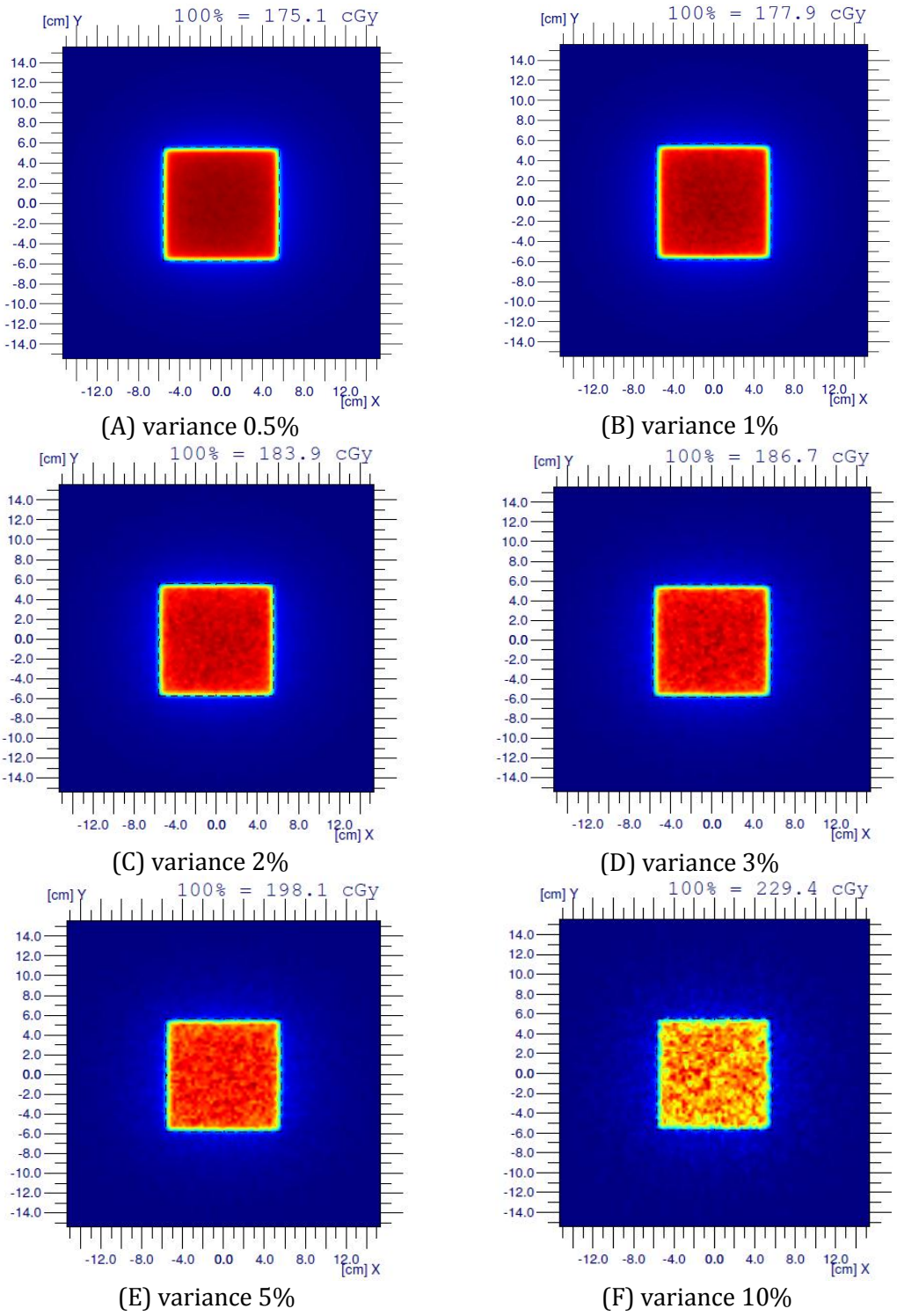
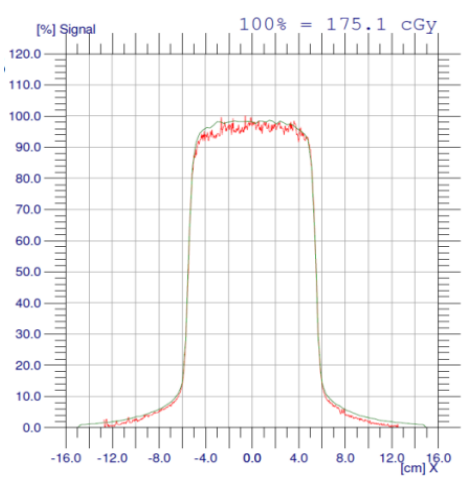


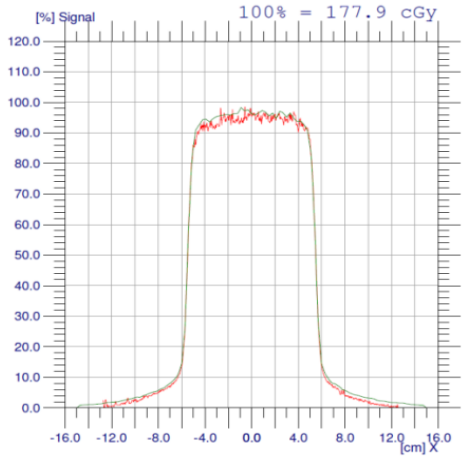
Figure 4: Calculated dose distribution

The figure shows 6 images shows the dose distribution calculated for a square field using different variance. Note how the graininess (noise) in the images increases with increasing variance.

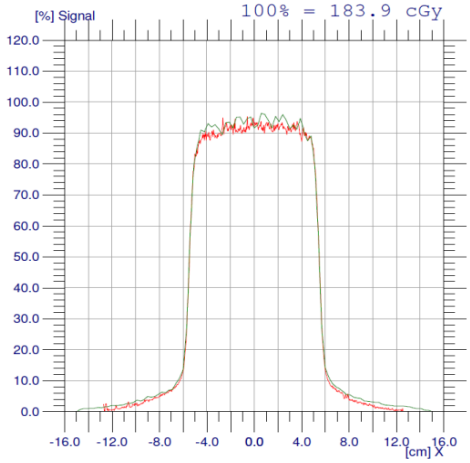
Figure 5 shows the profile of the calculated and the measured dose distribution at different variances for a square 10 x 10 cm field. The red plots in the figures represent the profile of the measured dose while the green ones show the profile of the calculated dose. At low variances, the measured and calculated dose distributions are mostly similar, whereas at higher variances, the noise in the calculated dose is increased, evident as the spikes present in the profile.



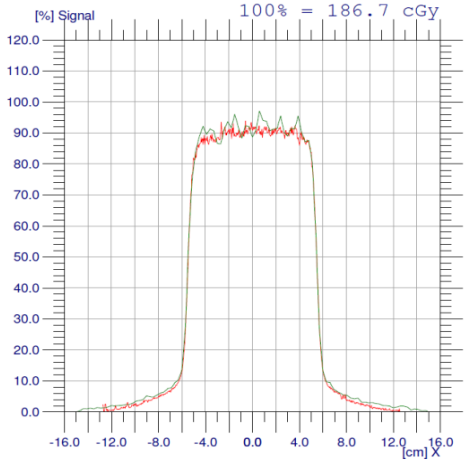
(A) variance 0.5%



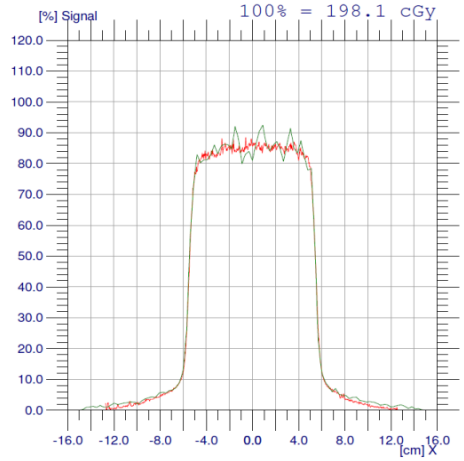
(B) variance 1%



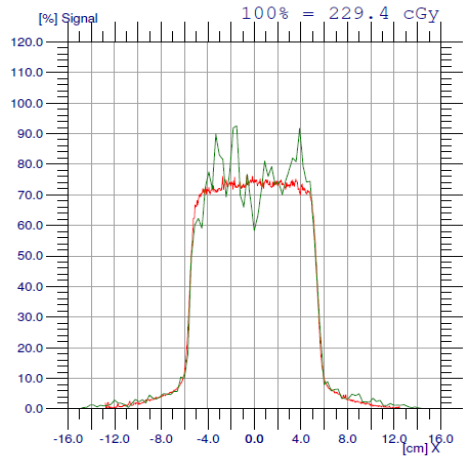
(C) variance 2%



(D) variance 3%



(E) variance 5

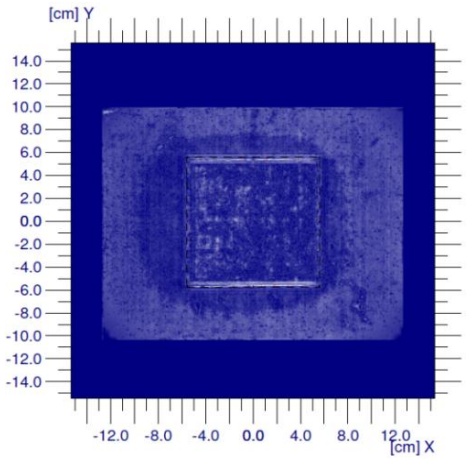


(F) variance 10

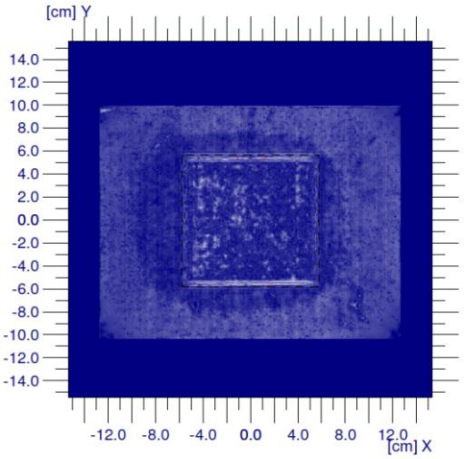
Figure 5: Profile of calculated and measured dose distribution

Note how the noise which is evident as the spikes in the calculated dose distribution increases with increasing variance.

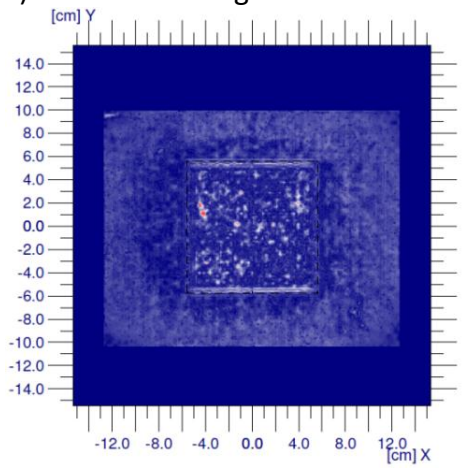
Figure 6 show the gamma evaluation results for dose calculated using different variance. The gamma comparison was made with DD and DTA of 4 % and 3 mm respectively. The values of the variance used in calculating the dose are included in the figure. It can be seen that the gamma result decreases (grainy red regions) as the variance (uncertainty in calculation) increases due to increasing disagreement of the data pairs as the variance is increased.



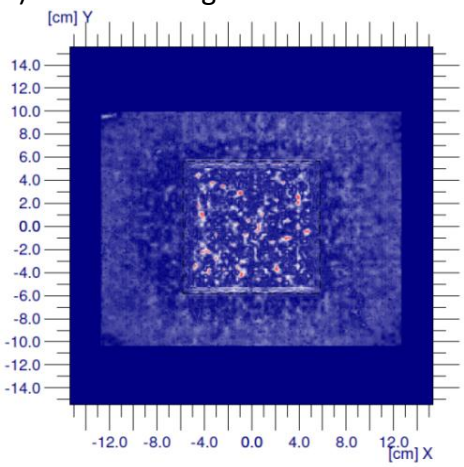
(A) variance 0.5%- gamma result = 99.90%



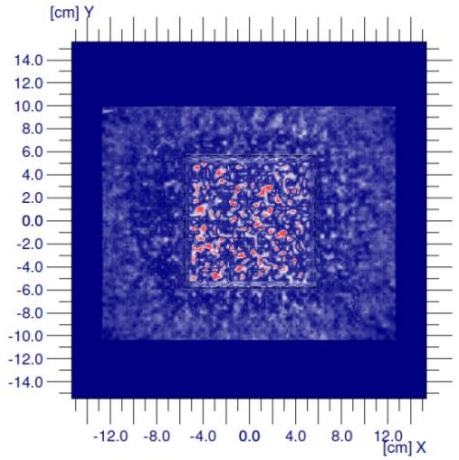
(B) variance 1%-gamma result = 99.86%



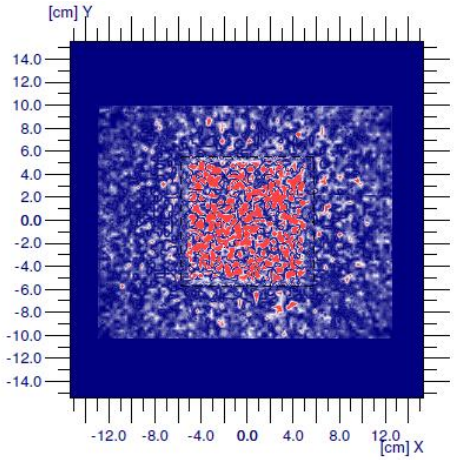
(C) variance 2%- gamma result = 99.72%



(D) variance 3%- gamma result = 98.62%



(E) variance 5- gamma result = 93.50%



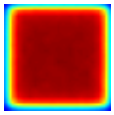
(F) variance 10- gamma result = 82.16 %

Figure 6: Gamma evaluation result

The gamma evaluation result for the square field shows the decrease in agreement between the calculated and measured dose distribution as the variance is increased.

Table 1 shows the field sizes, approximate shape, irradiation parameters (depth, monitor units (MU)), variances, gamma evaluation criteria and results for square fields. For all the fields, variances of 0.5, 1, 2, 3 and 5% were used in calculating the dose distribution while DD and DTA value pairs of 5 % and 5 mm, 4% and 3 mm, 3 % and 3 mm and 2 % and 2 mm were chosen as the pass criteria and used for evaluations. Gamma scores of 95 – 100% are considered as having passed the evaluation whereas lower scores are considered as a failure of agreement between the computed and measured dose distribution. It can be seen that at DD and DTA of 5 % and 5 mm respectively, nearly all the gamma evaluation have results of 100 % and only the evaluation result for the 15 X 15 field made with a variance of 5% failed the gamma evaluation at the stated DD and DTA values. It can also be seen that generally, the passed results generally reduce as the field size increases. For instance, at DD and DTA of 3 % and 3 mm and field sizes of 1.1 cm^2 to 3 cm^2 except for the calculated dose made with a variance of 5%, all gamma evaluation results exceeded the pass value. However, at larger field sizes, the failed points increase. At low variance up until the 3% recommended and gamma criteria $\geq$ 3 % and 3 mm for the DD and DTA respectively, almost all the calculated dose agrees with the measured dose, however at higher variance values, the level of agreement is reduced.

Table 1 shows the gamma evaluation result for the square fields studied.

Field Size W cm x L cm	The Shape	Depth (cm)	MU	Variance (%)	The Result Of Gamma Index: Dose Comparison Tools: The Dose Difference (DD) & The Distance-To-Agreement (DTA)				
					DD	5%	4%	3%	2%
					DTA	5mm	3mm	3mm	2mm
1.1 x 1.1		5	250	0.5	100%		100%	100%	100%
				1	100%		100%	100%	100%
				2	100%		100%	100%	100%
				3	100%		100%	100%	100%
				5	100%		100%	100%	100%
2 x 2		5	250	0.5	100%		100%	100%	100%
				1	100%		100%	100%	100%
				2	100%		99.17%	98.15%	93.87%
				3	100%		98.27%	97.15%	91.71%
				5	100%		98.00%	97.00%	89.60%
3 x 3		5	250	0.5	100%		99.71%	98.65%	94.59%
				1	100%		99.70%	98.83%	93.94%
				2	100%		99.93%	98.61%	92.62%
				3	100%		99.57%	98.01%	88.94%
				5	98.75%		96.71%	94.41%	86.65%
5 x 5		5	250	0.5	100%		100%	99.66%	91.69%
				1	100%		99.58%	97.94%	88.02%
				2	99.92%		97.29%	94.12%	81.67%
				3	99.50%		96.50%	92.88%	78.16%
				5	97.93%		93.22%	89.17%	77.25%

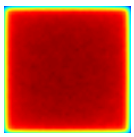
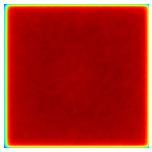
10 x 10		10	250	0.5	100%	99.91%	99.54%	98.64%
				1	100%	99.87%	99.49%	98.20%
				2	99.98%	99.77%	99.26%	95.92%
				3	99.84%	98.84%	97.38%	91.89%
				5	98.32%	95.43%	92.82%	82.53%
				10	82.16%	72.07%	65.95	52.59%
15 x 15		4	250	0.5	99.98%	99.34%	97.50%	84.06%
				1	99.97%	98.95%	96.43%	82.94%
				2	99.72%	97.23%	93.32%	80.61%
				3	98.61%	94.42%	89.62%	77.96%
				5	94.34%	86.04%	80.96%	69.17%
20 x 20 Half Field		2	200	0.5	99.90%	99.40%	98.09%	91.18%
				1	99.88%	99.28%	97.78%	90.34%
				2	99.76%	98.36%	96.00%	87.45%
				3	99.09%	96.54%	93.21%	83.76%
				5	96.33%	91.67%	86.55%	75.56%

Table 2 shows the results for the rectangular fields. For the 10 X 2 cm and 15 X 5 cm fields, note that two measurements were made with the films oriented horizontally and vertically relative to the positions of the MLCs and the interesting pattern of the comparison results observed for the two types of orientations. The gamma results for the 2 X 10 cm films are generally better than the results for the 10 X 2 cm film, whereas the results for the 15 X 5 cm films are generally better than the results for the 5 X 15 cm film.

Table 2 shows the gamma result for rectangular fields

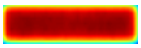
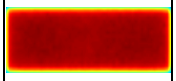
Field Size W cm x L cm	The Shape	Depth (cm)	MU	Variance (%)	The Result Of Gamma Index Dose Comparison Tools: The Dose Difference (DD) (%) & The Distance-To-Agreement(DTA) (mm)				
					DD	5%	4%	3%	2%
					DTA	5 mm	3 mm	3 mm	2 mm
0.6 x 1.1 Smallest Field in Monaco		5	250	0.5	100%		100%	100%	100%
				1	100%		100%	100%	100%
				2	100%		100%	100%	100%
				3	100%		100%	100%	100%
				5	100%		100%	100%	100%
10 x 2		5	250	0.5	100%		97.55%	94.16%	76.95%
				1	99.97%		97.17%	94.05%	76.35%
				2	99.92%		95.51%	91.30%	71.98%
				3	99.80%		95.18%	89.87%	70.77%
				5	98.40%		90.38%	86.62%	69.57%
2 x 10		5	250	0.5	100%		99.60%	98.81%	89.74%
				1	100%		99.58%	98.74%	89.30%
				2	99.94%		98.94%	97.10%	87.93%
				3	99.51%		97.94%	96.27%	87.77%
				5	99.48%		97.61%	96.15%	87.20%
15 x 5		5	250	0.5	99.85%		96.55%	91.59%	74.31%
				1	99.62%		96.39%	91.18%	74.00%
				2	99.01%		94.60%	89.07%	72.41%
				3	98.01%		91.87%	86.67%	71.31%
				5	93.58%		85.89%	80.38%	64.19%
5 x 15		5	250	0.5	99.57%		90.49%	78.35%	53.57%
				1	98.67%		89.24%	78.08%	52.66%
				2	96.67%		85.03%	75.68%	56.69%
				3	94.48%		82.79%	74.99%	54.63%
				5	88.04%		76.48%	70.88%	57.51%

Table 3 shows the gamma evaluation results for irregular fields obtained at various dose difference (DD) and distance to agreement (DTA) values. As should be expected, it can be seen that as the variance increases, the gamma evaluation result (proportion of pass) decreases. At DD and DTA of 3 % and 3 mm respectively, almost all dose calculation made with variance of less than 5 % passed the gamma evaluation, whereas most of the calculation made with variances equal to or higher than this value mostly failed (gamma ≤ 95%) .

Table 3 shows the field shape, exposure parameters, gamma criteria and results for irregular fields

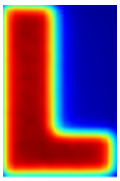
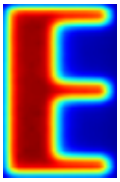
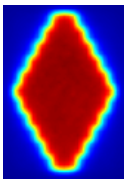
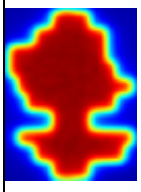
The Field	The Shape	Depth (cm)	MU	Variance (%)	The Result Of Gamma Index Dose Comparison Tools: The Dose Difference (DD) (%) & The Distance-To-Agreement(DTA) (mm)				
					DD	5%	4%	3%	2%
					DTA	5mm	3mm	3mm	2mm
Letter L		5	250	0.5	100%		99.37%	98.54%	90.84%
				1	100%		99.21%	98.23%	90.38%
				2	100%		98.89%	98.01%	89.84%
				3	99.52%		97.75%	96.63%	89.55%
				5	99.21%		96.02%	94.38%	83.06%
				10	95.91%		89.48%	86.83%	74.50%
Letter E		5	250	0.5	100%		99.00%	97.75%	84.82%
				1	100%		98.62%	97.22%	84.31%
				2	99.96%		97.20%	95.51%	81.76%
				3	99.58%		96.38%	94.33%	82.47%
				5	98.31%		93.78%	92.13%	79.18%
				10	96.14%		90.63%	88.00%	73.74%
Diamond Shape		5	250	0.5	98.71%		93.54%	90.65%	78.64%
				1	97.95%		93.37%	89.85%	76.92%
				2	96.92%		92.24%	89.14%	76.95%
				3	96.15%		92.04%	89.30%	78.41%
				5	95.91%		90.72%	87.78%	76.01%
				10	91.82%		85.76%	81.88%	68.16%
Random Shape		5	250	0.5	100%		99.12%	98.05%	84.77%
				1	100%		99%	98.8%	84.95%
				2	99.88%		98.32%	96.19%	82.43%
				3	99.30%		96.42%	94.33%	81.35%
				5	97.75%		94.26%	91.85%	78.84%
				10	92.60%		85.70%	82.26%	67.56%

Table 4 presents the Monte Carlo dose calculation time at different variances. The table shows a huge increase in the calculation time as the uncertainty is reduced. The calculation time appears to increase exponentially as the variance is reduced.

Table 4 Calculation times for square fields at different variances

	Calculation time (s)						
Variance (%)	**2cm²**	**3cm²**	**5cm²**	**7cm²**	**10cm²**	**15cm²**	**20cm²**
0.5 %	41,5	69	163	306	621	1363	2412
1 %	25,7	34	57	94	174	368	632,5
2 %	22,5	25	31,5	39,5	62,2	119	192
3 %	21,5	23,5	25,5	31	40,5	71,5	113
5 %	20	22,5	23,5	25	30,7	47.5	69

Figure 7 shows calculation time plotted against variance for some field size which shows that the time increase with reduction in variance does indeed appear to be exponential. The calculation time increased as the variance is reduced and as the field size is increased.

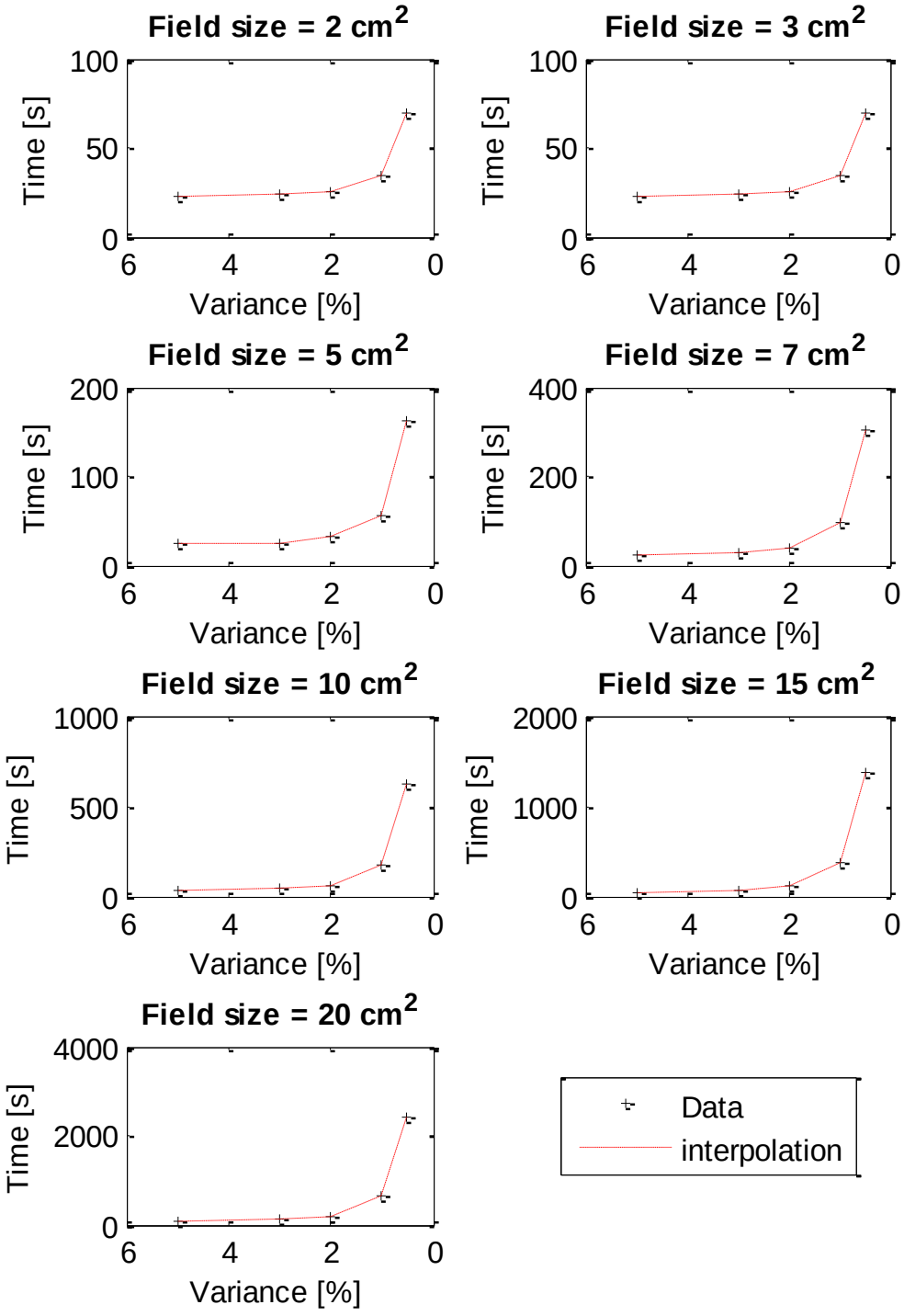


Figure 7: Plot showing Monte Carlo calculation time as a function of variance and field size

3.2 Dosimetric effect of variance on DVH – IMRT clinical cases

Figures 8 and 9 are used to show the effect of using different variance when computing the dose distribution in a patient. Considering figure 8, the calculation made with a variance of 10 % appears to agree more with that made with a variance of 0.5 % (assumed to be the most correct), than the calculation made with a variance of 5 %. In the region of the dose to about 10 % of the structures (high dose region), the effect of varying variance seems to be most evident with about 1000 cGy difference in the dose to the structure in this region being projected with the dose calculations made with variance of 0.5, 1 and 3 % and that made with a variance of 5 %.

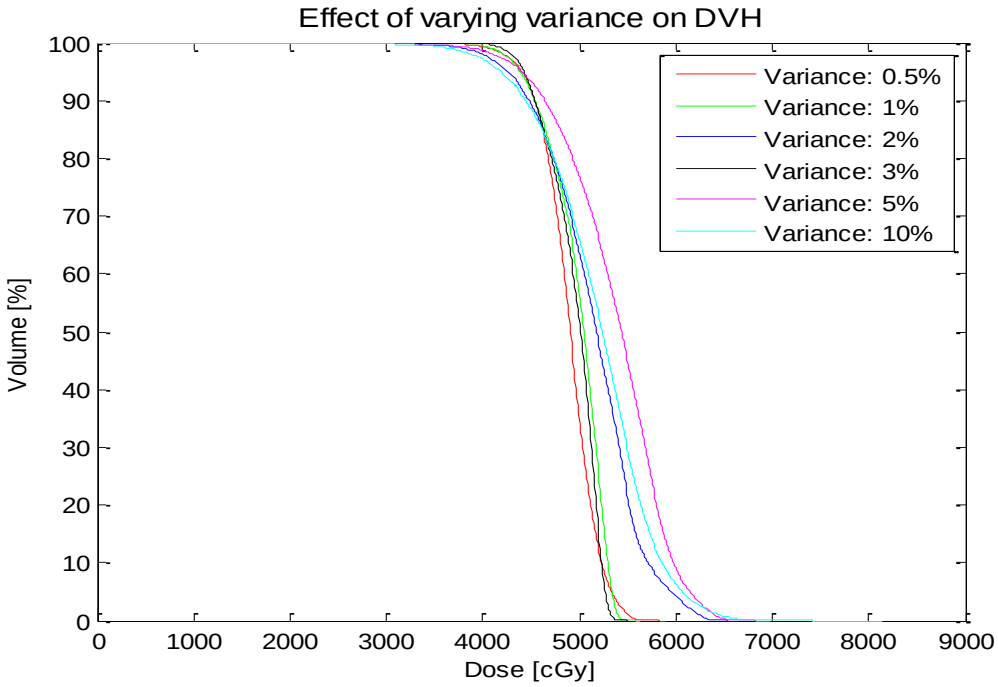


Figure 8 shows the effect of using different variance in computing the dose on the dose volume histogram (DVH) of a patient (case 1)

Considering figure 9, if the DVH computed with a variance of 0.5 % is assumed to be the most accurate, then the observed pattern is mostly in agreement with the expectation since that made with a variance of 10 % deviated the most from this value as is expected, and that computed with a variance of 1 % is probably the second most accurate (again, assuming that that made with a variance of 0.5 % will agree more with the measured. No measurements were made in the case of the patient dose calculation data.

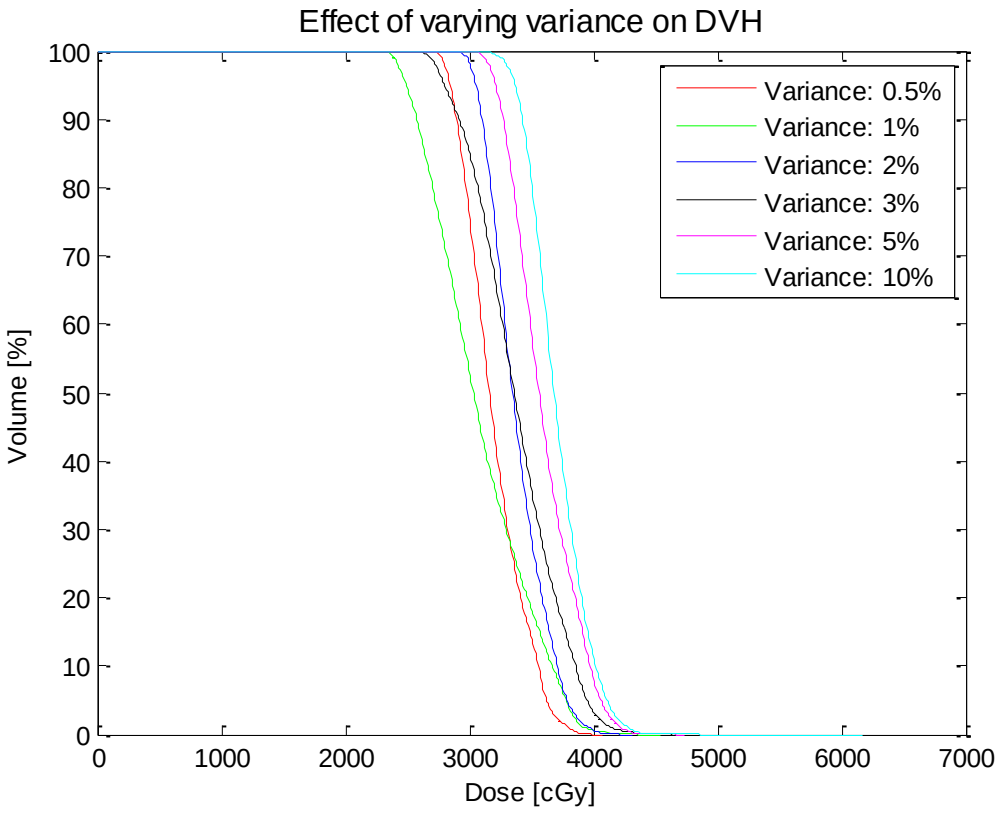


Figure 9: shows the effect of using different variance in computing the dose on the DVH of a patient (case 2).

4. Discussion

Table 3 shows the dose calculation time for a square 10 X 10 cm field while tables 1 and 2 show the gamma evaluation results. From table 1, it can be seen that the gamma evaluation results for the calculations made with variances of 0.5 and 1 % are quite similar. Table 3 however show the dose calculation time with a variance of 0.5% is more than 3 times the time required for dose calculation with an uncertainty of 1%. It can thus be concluded that there is no significant gain in going lower than a variance of 1% since there is no significant improvement in the accuracy of the calculated dose when compared with the measured. Figure 7, which is a plot of the computation time against variance show that it will be impossible to achieve a Monte Carlo calculation dose distribution calculation of absolute accuracy, i.e. one with no variance. This is so because as the plots clearly show, as the variance tends to zero, the computation time tends to infinity.

The recommended upper and lower values for variance to be used for calculation of dose distribution are 3 % and 1 % respectively. The findings of this research also agrees with the suggestion since at higher variances, the result of the gamma result is decreased due to increase failure of agreement between the calculated and the measured dose distribution. The results of this research is in agreement with the recommended variance values of 1% to 3% since the range of values appear to be very good compromise between reducing calculation time and having an accurate calculation of dose distribution. Based on the gamma result, at

DD of 4% and DTA of 3 mm, there is agreement between the calculated and measured dose distribution in almost all the cases. This value is also the recommended value that is used in clinical routine for IMRT quality assurance. The purpose of the dose computation in the first place is to accurately determine the dose that will be deposited in the patient from the beam geometry employed for the treatment. To simulate the dosimetric impact due to the use of different variance values in the dose computation, patient dose was computed for IMRT cases using different variance values. The results showed substantial difference between the plans made with the different variances (Figures 8 and 9). Dose computed with variances of 1 and 10 % for a subject showed up to 1000 cGy difference in the dose to 90 % of the structure's volume, while the dose calculated with variance of 0.5 and 5 % for another subject showed up to 1000 cGy difference in the dose to approximately 10 % volume of the structure. Such large variation could lead to adverse effects (Indrin et. al., 2007)..

The results also show that at small field sizes, the calculated and measured dose distribution always agrees irrespective of the variance values used. In other words, for small field size, even when high values of variance were used during the dose calculation, the gamma evaluation passed. Thus for a small field, large uncertainties in calculation still yielded accurate result. This means that when the field is small, the large variances can be employed to reduce the calculation time. At DD and DTA of 5% and 5 mm respectively, the evaluation result is nearly passed in all the cases. However, this is not a good evaluation criterion since the pass criterion is fairly loose and a dose difference value of 5% between the calculated

and predicted is substantial. Lower values such as the 4% and 3 mm values commonly employed in QA programs are more stringent criteria that should be used.

A unique pattern is observed for the gamma evaluation result of the rectangular field (Table 2). The 10 x 2 cm and 2 x 10 cm field sizes are of the same dimension; however both shapes are defined by different number of MLC leaf pairs that has to be in the open position during the irradiation of the film. For the 10 x 2 cm field size, only2 leaf pairs are open, whereas the 2 x 10 cm fields requires 10 MLC leaf pairs to be in the open position (leaf width = 1 cm). The 10 x 2 cm field will be referred to as the horizontally oriented field and the 2 x 10 cm field will be referred to as the vertically oriented field. Both field dimensions will be referred to as *case a*. The same convention will be used in describing the 15 x 5 cm and 5 x 15 cm fields, i.e. 15 x 5 cm field being horizontally oriented and 5 x 15 cm field being vertically oriented. Both fields will be referred to as *case b*. It was observed that the gamma results for the vertically oriented field was better than that of the horizontally oriented field for all observations in *case a*, with a reversal of the pattern in *case b*, i.e. the horizontally oriented field having more pass values than the vertically oriented one. The reason for such discrepancies and reversal of pattern is not easily understandable. This observation is however, not statistical since this pattern was true for all 40 comparison points made for the two field sizes.

The results of this research are also mostly in agreement with results reported by earlier researchers. At a variance of 3 % and gamma evaluation criteria of 3 % and 3 mm for the DD and DTA respectively, the results are mostly similar with that reported by Wang (Wang et al., 1996). However, with an increase in the acceptable variance in dose calculation to 5 %, the result agree more with that reported in the work of Ma (Ma et al., 2000, Oldham and Webb, 1997). From the results, it can thus be possible to predict the calculation time (for a square field in this case) for any given field size and variance with fair degree of accuracy with increase with increasing field size. Since the data shows calculated time to tend more towards agreeing with the modeling function as the field size increases. The results also show that the calculation times tend to infinity as the variance approaches zero.

5. Conclusion

In conclusion, the result of this work showed the Monte Carlo method to achieve good dose calculation results. The results are similar to that reported in previous literature. However, the accuracy of the dose calculation comes at the expense of computation time whose increase appears exponential with increasing dose calculation accuracy. The dosimetric effect of using different variances on the DVH pattern was studied and the result showed substantial dose difference between dose computations made with low and high variances. This huge dose difference due to variance selection could result in serious therapeutic consequences.

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