DETEMINATION OF OUTPUT FACTOR FOR 6 MV SMALL PHOTON BEAM: COMPARISON BETWEEN MONTE CARLO SIMULATION TECHNIQUE AND microDiamond DETECTOR

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ABSTRACT

In order to improve the quality of life for cancer patient, the radiation techniques are constantly evolving. As a result, the two modern techniques which are intensity modulated radiation therapy (IMRT) and volumetric modulated arc therapy (VMAT) are quite promising. They comprises of many small beam sizes (beamlets) with various intensities to achieve a higher radiation dose to the irregular tumor and lesser dose to the nearby normal tissue. The study aimed to prove that whether or not the microDiamond detector (PTW Freiburg, Germany, PTW-60019), a synthetic single crystal diamond detector, is suitable for small field output factor measurement. The output factor for the collimator field size of 1×1 and 5×5 cm^2 of 6 MV photon beams were applied. The percentage depth dose and beam profile at 5 and 10 cm depth were also analyzed. The results were compared with those measured by the stereotactic field detector (SFD: Scanditronix IBA) and the Monte Carlo simulation (EGSnrc / BEAMnrc / DOSXYZ). By the calibration of the Monte Carlo, the percentage depth dose and dose profile measured by the stereotactic field detector (SFD) for field size of 10×10 cm^2 at a distance of 100 cm SSD were applied to adjust the suitable parameters of the energy of initial electrons beam and radial intensity distribution width. It was found that those parameters were 6.3 MeV and 0.6 cm, respectively. Comparison of the values obtained from the calculations and measurements are consistent, no more than 1% difference. When comparing PDD and beam profile at a depth of 5 cm and 10 cm in field size of 1×1 cm^2 and 5×5 cm^2, it was found that the SFD detector, microdiamond detector and Monte Carlo simulation are different less than 2%. At 5 cm depth the penumbra width for field size of 1×1 cm^2 of SFD and microDiamond are 2.74 mm and 3.52 mm, respectively and for field size of 5×5 cm^2 the penumbra widths of the former and the latter are 3.36 mm and 4.68 mm, respectively. For the output factor comparison of microdiamond with SFD detector and microdiamond detector with Monte Carlo simulation, the results demonstrate that the percentage differences are 1.32% and 1.47% for field size of 1×1 cm^2 and 1.17% and 1.25% for field size of 5×5 cm^2, respectively. This study found that the difference of microDiamond in small field dosimetry compared with the SFD detector and Monte Carlo simulation is within 2%. The microDiamond detector can be considered as one of the suitable detectors for small field output factor measurement.

KEY WORDS : SECOND CANCER / RISK ASSESSMENT / ORGAN EQUIVALENT DOSE / IMRT

63 pages

กำหนดค่าเอาท์พุทแฟคเตอร์สำหรับลำรังสีโฟตอนพลังงาน 6 MV ในพื้นที่ลำรังสีขนาดเล็กโดยเปรียบเทียบระหว่างการจำลองด้วยมอนติคาร์โลเทคนิคและไมโครไดมอนต์ดีเทคเตอร์

DETERMINATION OF OUTPUT FACTOR FOR 6 MV SMALL PHOTON BEAM: COMPARISON BETWEEN MONTE CARLO SIMULATION TECHNIQUE AND microDiamond DETECTOR

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บทคัดย่อ

การรักษาด้วยรังสีมีการพัฒนาอย่างต่อเนื่องเพื่อให้ผู้ป่วยมีคุณภาพชีวิตที่ดียิ่งขึ้นเป็นผลให้เกิดเทคนิคการฉายรังสีแบบปรับความเข้ม (IMRT) และปรับความเข้มหมุนรอบตัวผู้ป่วย (VMAT) ซึ่งเป็นเทคนิคที่มีประสิทธิภาพทำให้มีแนวโน้มในการใช้งานมากยิ่งขึ้น โดยลำรังสีที่ใช้จะมีลักษณะเป็นลำรังสีขนาดเล็ก (beamlet) ที่มีการปรับเปลี่ยนความเข้มของลำรังสีตามรูปร่างของก้อนมะเร็ง การวัดปริมาณรังสีในพื้นที่ลำรังสีขนาดเล็กจำเป็นต้องอาศัยความรู้ ความเข้าใจตลอดจนเครื่องมือในการวัดที่เหมาะสม ในการศึกษานี้ได้ทำการวัดปริมาณรังสีในพื้นที่ลำรังสีขนาด 1×1, 2×2, 3×3 และ 5×5 ซม^2 เพื่อหาค่าเอ้าท์พุทแฟคเตอร์สัมพัทธ์ในโฟตอนพลังงาน 6 MV ด้วยหัววัดรังสี microDiamond (PTW Freiburg, Germany, PTW-60019) และหัววัดรังสี stereotactic field diode (SFD: Scanditronix IBA) เปรียบเทียบกับการจำลองด้วยเทคนิคมอนติคาร์โล (EGSnrc / BEAMnrc / DOSXYZ) ที่ทำการจำลองด้วยข้อมูลการเปรียบเทียบจากการวัดปริมาณรังสีในแนวลึกและปริมาณรังสีในแนวระนาบที่ระดับความลึก 5 และ 10 ซมในพื้นที่ลำรังสี 10×10 ซม^2 ด้วยหัววัด SFD โดยใช้ค่าเปอร์เซ็นต์ความแตกต่างระหว่างการวัดและการคำนวณน้อยกว่า 1% ซึ่งเมื่อทำการเปรียบเทียบการวัดปริมาณรังสีด้วยหัววัด microDiamond และ SFD พบว่ามีค่าเอ้าท์พุทแฟคเตอร์สัมพัทธ์แตกต่างจากค่าที่ได้จากการคำนวณด้วยมอนติคาร์โลน้อยกว่า 2% โดยหัววัดรังสี microDiamond มีค่าเปอร์เซ็นต์ความแตกต่างเมื่อเทียบกับการคำนวณและ SFDเท่ากับ 1.47% และ 1.32% ตามลำดับสำหรับพื้นที่ลำรังสี 1×1ซม^2 และ 1.25% และ 1.17% สำหรับพื้นที่ลำรังสี 5×5ซม^2 นอกจากนี้เมื่อพิจารณาความกว้างของเงามัวจากการวัดด้วย microDiamond และ SFD ในพื้นที่ลำรังสี 1×1ซม^2 พบว่ามีค่าความกว้างของเงามัวเท่ากับ 3.25 และ 2.74 ม.ม.ตามลำดับ และพื้นที่ลำรังสี 5×5 ซม^2 มีค่าความกว้างของเงามัวเท่ากับ 4.68 และ 3.36 ม.ม. ตามลำดับ ซึ่งเห็นได้ว่าการวัดปริมาณรังสีเพื่อหาค่าเอาท์พุทแฟคเตอร์สัมพัทธ์ในพื้นที่ลำรังสีขนาดเล็กด้วยหัววัดรังสีทั้งสองให้ค่าความแตกต่างอยู่ภายใน ±2%

ผลการวิจัยนี้มีข้อสรุปคือ microDiamond detector ให้ผลการวัดค่าเอาท์พุทแฟคเตอร์สัมพัทธ์ใกล้เคียงและสอดคล้องกับค่าที่คำนวณได้และวัดด้วยหัววัด stereotactic field diode โดยมีค่าเปอร์เซ็นต์ความแตกต่างมีอยู่ภายในเกณฑ์ที่กำหนด ($\leq$2%)

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LIST OF ABBREVIATIONS

ABBREVIATIONS	Term
2D	Two Dimentional
3D-CRT	Three Dimentional conformal radiation therapy
AAPM	American Association of Physics in Medicine
CBCT	Cone beam computed tomography
cm	Centimeter
cm^2	Centimeter Square
CPE	Charge Particle Equivalent
CT	Computed Tomography
EPID	Electronic Portal Imaging Device
Gy	Gray
IC	Ionization Chamber
IGRT	Image Guide Radiation Therapy
IMRT	Intensity Modulated Radiation Therapy
kV	Kilo Voltage
LINAC	Linear Accelerator
MLC	Multi-Leaf Collimator
OBI	On Board Imager
OAR	Organ at risk
PDD	Percentage Depth Dose
SD	Standard Deviation
SFD	Stereotactic Field Detector
TPS	Treatment Planning System
VMAT	Volumetric Modulated Arc Therapy

CHAPTER I
INTRODUCTION

Radiation therapy or Radiotherapy is a method of treatment aimed towards the curing of cancer patients through high radiation energy that causes the ions to split apart and destroy or inhibit the growth of tumor cells. Normally, 50-60 percent of cancer patients are treated with radiation therapy [1]. The treatment may only involve radiation therapy alone or in combination with other methods and alternatives to surgery, chemotherapy, or hormonal treatment to obtain the best treatment results. However, radiation used in the treatment of cancer will destroy and inhibit the growth of both the tumor cells and tissues within the radiation spectrum and those that surrounds the tumor cells. Therefore, the treatment of cancer with radiation therapy also has limitations that make it necessary and thus, crucial to search for alternatives and ways to enable the tumor cells to receive a high amount of radiation in order to destroy and inhibit growth, not allowing the cells to repair. The surrounding tissues within the radiation spectrum and those that surrounds the tumor cell, though, must receive the lowest amount of radiation to be able to repair and function well after the radiation therapy; radiation for the treatment of cancer patients will also involve passing through the tissues that surround the tumor cells. As tumor cells are often located deep in the dermal layers of the skin, treatment via radiation will require radiation of over 7,000 cGy to destroy the tumor cells [2]. The amount is considered high compared to the amount of radiation received by the tissues surrounding the tumor cells that can still go through repairing processes and are not damaged. Therefore the International Commission on Radiological Units and Measurement: ICRP sets the radiation dislocation for the entire radiation therapy process of not over 5 percent of the radiation amount (dose) set or required. The dose or the amount of radiation the patients receive are relatively high; if there are even slight or minor errors in the calculation of the radiation, the projection of high radiation can be done upon the surrounding tissues instead of the tumor cells specifically when the tumor cells are located closely to important organs, resulting in detrimental complications for patients and may lead to death.

Treating patients with cancer involve numerous radiation techniques including the Conventional Technique (2D), 3DCRT, the development of a computerized radiation technique: "Multi-Leaf Collimator : MLC" that enables radiation to be sheltered and shielded through a computerized program, introducing new techniques like the Intensity Modulated Radiation Therapy (IMRT) and the Volumetric Modulated Arc Therapy (VMAT), following the increasing demands to treat patients. Aimed towards the projection of high radiation upon the tumor and the tissues surrounding the tumor least affected, the projection of high radiation towards the tumor is limited by the tolerance dose or the amount of radiation the surrounding tissues received and are affected. However, with radiotherapies like those of the IMRT and the VMAT, treatment extents the range from the ability to control radiation intensity according to the tumor, conformal radiation technique enabling the affected tissues in the surrounding area to be controlled whilst the radiation upon the tumor can be increased. Therefore, these two techniques are widely accepted as treatments for patients as shown in Figure. 1.1.

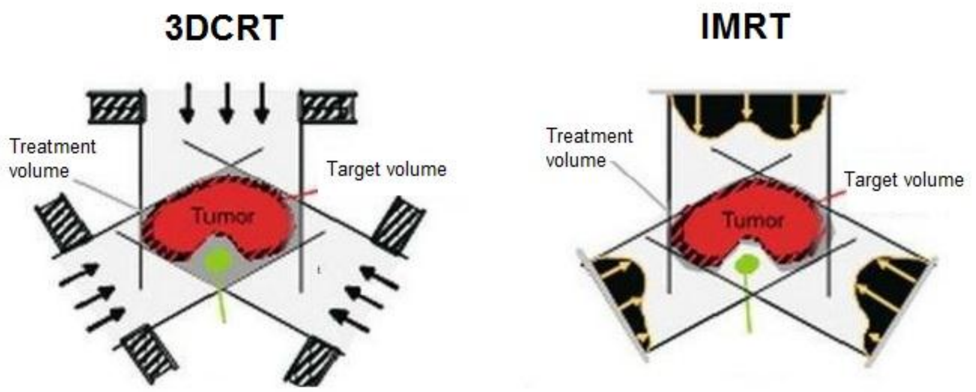


Figure 1.1 Comparison radiation technique between 3DCRT and IMRT [3]

Through the continuous movement of the MLC, radiation intensity can be controlled in both IMRT and VMAT. As radiation is performed dynamically to fully cover the tumor cell according to its Figure 1.2, 'beamlets' are produced where its sizes depend upon the sizes of MLC's leaf and its movement. Therefore, it is observed that IMRT and VMAT have smaller fields compared to that of the traditional radiotherapy.

Treatment involves accuracy, that is, the amount of radiation the tumor receives must be as stated in the treatment plan. Thus, the study aims to reduce errors resulting from radiation in which involves an experiment to determine the specific beam characteristic and is a first step to control the errors of radiation and beam data collection. As beams are used as a standard in the calculation, accurate percentage depth dose (PDD), beam profile, and output factor will determine patient's treatment plans.

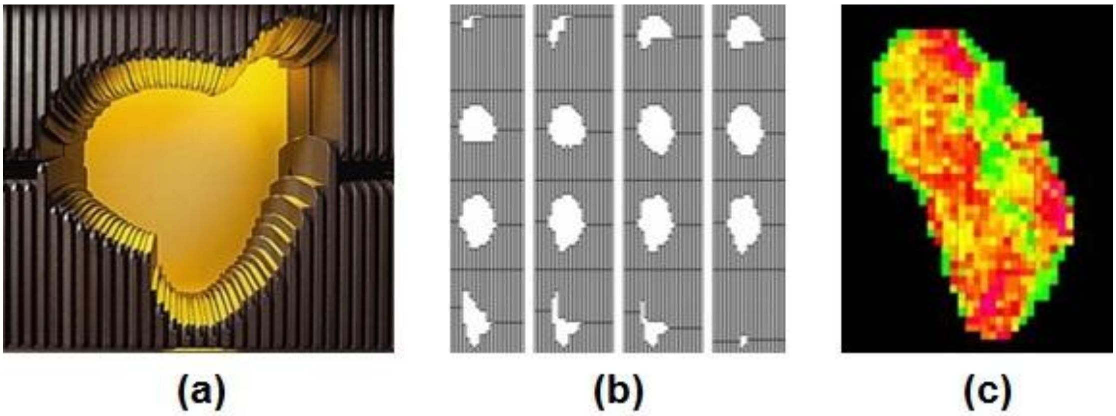


Figure 1.2 (a) MLC from images courtesy of Varian Medical Systems of Palo Alto, California. Copy right 2013, Varian Medical Systems.

(b) MLC movement each beam eye view by computer control [4].

(c) Feature of beamlet from intensity profile of an IMRT.

To accurately measure a small field area, it is crucial to understand the characteristics of the field and the measuring process. The measurement of radiation within a small field may result in errors in measurement due to the lack of charged particle equilibrium (Non-CPE) and the errors in the measurement of field width that overlaps with the edges, or the penumbra, and perturbation that results in volume averaging measurement. Thenceforth, the measurement of radiation in a small field requires a suitable detector in which is dose rate independent, energy independent, angular independent, high in resolution, high in sensitivity, active in linearity, small and less in perturbation, and lastly, is tissue equivalent.

Radiation detectors used in small fields are those of ionization chambers (IC) and semiconductors. Although the development of smaller ionization chambers is in progress, the detectors are not small enough to reduce perturbation, which then leads to volume averaging that results in underestimation. As the detectors become smaller in size, it also leads to lower sensitivity. In contrast to the latter, semiconductors like diode have higher sensitivity than IC, is smaller in size, and higher in measurement resolution. However, the disadvantage of this radiation detector is its sensitivity that depends upon the dose rate (dose rate dependent), is angular dependent, and is non-water equivalent, which leads to the possibility of an increase in a secondary electron that then results in overestimation. As a result, there is yet a suitable detector to measure the amount of radiation projected over a small field without any disadvantages. However, Ramathibodi hospital uses the stereotactic field detector (Scanditronix IBA SFD) with a diameter of 0.6 mm and sensitive volume of 0.017 mm^3 in the radiotherapy process to measure small fields. Currently, detectors are still in development to suit small fields, shifting from PTW's diamond detector to microDiamond detector (PTW-60019) with a diameter of 2.2 mm and a sensitive volume of 0.004 mm^3, which is four times smaller than that of SFD's. Moreover, the sensitivity of energy dependence and dose rate dependence are lowered along with the characteristics of being tissue equivalent, marking it a beneficial detector towards the small field, but a costly tool. Furthermore, as the measurement of small fields may result in errors and lacks the standard conditions in measurement compared to large fields, the evaluation of right measurement processes and controls are as well difficult. Monte Carlo Simulation is a study of the interaction between radiation fields and mediums in the random or stochastic process. With random numbers, probability distribution, the fundamentals of physics, and the absorbing of center photon radiation as interaction forecasting factors, the study of complex problems can be simpler and will further lead to standard acceptance.

This study aims to compare the relative output factor obtained from a measurement of the stereotactic field detector (Scanditronix IBA SFD), which is used as a standard for small fields in radiotherapy used in Ramathibodi hospital and microDiamond detector (PTW Freiburg, Germany, PTW-60019), with the outcome from Monte Carlo simulation.

CHAPTER II
OBJECTIVES

The objective of this study was:

To compare the relative output factor obtained from a measurement of the stereotactic field detector (Scanditronix IBA SFD), which is used as a standard for small fields in radiotherapy used in Ramathibodi hospital and microDiamond detector (PTW Freiburg,Germany, PTW-60019), with the outcome from Monte Carlo simulation.

CHAPTER III
LITERATURE REVIEWS

3.1 Overview

3.1.1 Monte Carlo Simulation

The evolution of radiation therapy has been continuously developed as it enters the phase in becoming the "Advance Radiation Therapy" that will soon replace the traditional 2D radiation therapy. The latest radiation technique aims to limit and set the radiation spectrum, forcing it to take up similar shapes and sizes of the lesions that needed treatment to reduce the area that receives high amount of radiation, resulting in a smaller radiation spectrum or a small field.

Intensity Modulated Radiation Therapy (IMRT) and Volumetric Modulated Arc Therapy (VMAT) are both beamlet based intensity modulated techniques. Therefore, this results in a difference in radiation measurement of radiation field of over or equal to 4×4 cm^2, as radiation measurement in a field area smaller than 4×4 cm^2 may lead to the lack of charge particle equilibrium (non-CPE) and the overlapping of radiation spectrum and penumbra resulting from oversized radiation spectrum that leads to perturbation of detector. Although there are various detectors to choose from, ranging from sizes (mini to micro), shapes (timber, spherical, and plane parallel), and types (ionization chamber, semiconductor, chemical, film), there are still no definite specifications that can serve as a standard to the selection of a radiation detector to measure the amount of radiation in small fields.
The simulation and adhesion of particles, also known as Monte Carlo Simulation, is the simulation of linear accelerators (LINAC) that relies upon the details of the internal components of the particle accelerators, as they are products of the manufacturer. It is considered as one of the most effective methods of measurement that gives accurate results, as Monte Carlo Simulation deals with the simulation of the movement of particles in a medium that relies on statistical evaluation or probability

distributions of each particle and the interactions in intermediaries; data obtained will be used for the findings of the radiation dose through program evaluation or developed codes like EGSnrc, MCNP, PENELOPE, and FLUKA.

Monte Carlo method is widely used in fields of Medical Physics for LINAC model simulation and the calculation of dose for treatment of cancer patients in the field of radiation therapy. As it is acknowledged as an effective method that accurately measures the amount of radiation or the dose to treat the patients, continuous studies have been witnessed to develop the Monte Carlo method for clinical radiotherapy dose calculation. Monte Carlo codes will be introduced, where each code will own various functions for varying uses: Electron Gamma Shower Version 4 (EGS4) is a code known within the Medical Physics field for the simulation of electron and photon transport, DOSXYZ is a code for the calculation of dose distribution in Voxel Phantom (a continuous development of the EGS4), and the MCDOSE (developed from the EGS4) simulates the beam modifier and calculates the dose for radiotherapy treatment planning. Various developments of codes are all aimed towards model accuracy, including being able to simulate under a short period of time and obtain accurate results. Therefore, Monte Carlo method is among the most effective techniques to calculate through random sampling and solve indefinite problems through probability distribution.

3.1.2 Monte Carlo Code

Monte Carlo simulation for radiation therapy involves the use of a widely known program or code often known as the Electron Gamma Shower (EGS), which is a code for the simulation of the interactions between photons and electrons. EGS owns codes for the simulation of particle accelerators which includes various modules (components module) to choose from: BEAMnrc and DOSXYZnrc.

. 3.1.2.1 BEAMnrc Code is a code used in EGS4/EGSnrc to simulate the radiation from particle accelerators, including both photons and electrons, that relies upon the accurate entry of the details of the particle accelerator and its internal components. As BEAMnrc simulates the radiation from the latter information, data obtained will appear in the form of Listing Output file, Phase Space file, and Graphic data for the calculation of radiation dose.

3.1.2.2 DOSXYZnrc Code is among Monte Carlo's codes for the calculation of radiation dose in a medium, represented in a 3D form of a rectangular voxel. The code can set the required characteristics needed for each voxel, for instance, setting up a voxel that imitates the characteristics of tissue, where the thickness of each voxel can also be specified. DOSXYZnrc uses data obtained from the simulation through BEAMnrc or Phase space to calculate radiation dose and displays the results in a 3D format with unit as Gy per incident particle. Furthermore, the values of uncertainty of each voxel will also be displayed.

3.1.3 Monte Carlo Dose Calculation Process Basic Principle

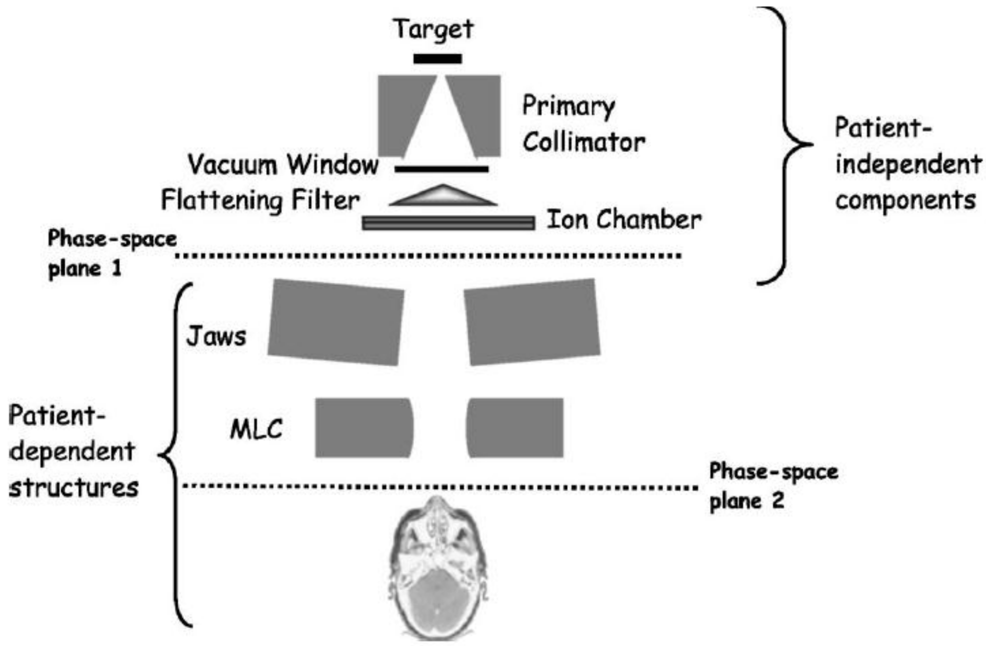


Figure 3.1 Components of Linear Accelerators for Monte Carlo simulation

3.1.3.1 PSD Generation (Pulse-shape discrimination) is a process to pass on the particles from the target, the collimator, the flattening filter, until the ion chamber. Therefore, this process will provide the details of the accelerator geometry and simulates in the form of a Phase Space file, which provides information of the particles including the location, the direction, and the electric charge. The latter information provided in the Phase Space file will be used for the calculation of the radiation dose (Figure3.1).

3.1.3.2 Dose Calculation is the process where the particles pass through the internal components of the particle accelerator, like the jaw, MLC, Block, and the wedge to the location where the calculation of radiation dose is required. The medium for calculation can be selected, for instance, radiation dose calculation in water phantom, solid phantom, or radiation dose calculation in patients. Nevertheless, calculation through the Monte Carlo technique is in the form of voxel.

3.1.4 Monte Carlo Optimization

Although the Monte Carlo Simulation technique provides accurate information, its main limitation is the period of time required for simulation. To obtain accurate results upon the calculation, high amount of particles are required. Once there is an increase in the number of particles, the accuracy of the calculation also increases accordingly. However, the CPU time will also increase as influenced by the latter. The amount of particles required to provide accurate information upon simulation, once compared to the radiation dose of 2 Gy/ fraction, is at least 100 million. Therefore, time needed for the calculation process then takes longer. With this reason, it is crucial to search for Monte Carlo optimization possibilities, to accelerate the simulation process along with providing the most accurate results with reliance upon the following equation:

$$\varepsilon = \frac{1}{\sigma^2 T}$$

When	ε	is efficiency
	σ^2	is Variance on the quality of interest
	T	is computing time to obtain a variance σ^2

From the equation, it is seen that decreasing σ^2 and T to their lowest values can enable the optimization of Monte Carlo Simulation as follows:

3.1.4.1 Decrease the value of σ^2 and let T stay constant, this method is known as "Variance reduction technique", where decreasing the values of σ^2 can be done as: Splitting and Russian Roulette, Photon forcing, Electron track repeating technique.

3.1.4.1.1 Splitting and Russian Roulette is a method that derives from the principle states the photons are unevenly weighted.

Therefore, heavy weighted photons will split apart, therefore decreasing the weight and divides into a group of N Photon (where each photon weighs 1/N of the initial weight) where they transport to selected areas and interact. Once interaction occurs, this will result in high numbers of scattering photons that are less in weight, resulting in slight contribution at the target dose distribution. Moreover, optimization can be successfully enhanced through the method of Russian Roulette to gather the light weight particles and together add up to the weight.

3.1.4.1.2 Photon Forcing controls and forces the particles to the locations marked.

3.1.4.1.3 Electron Track Repeating technique uses the electron track that has initially been developed in the uniform tissue phantom that has one density and will be repeated to use in the new location or new geometry that varies, with varying densities. This will result in higher optimization and accuracy. The electron track not only minimizes or enlarges according to the material density, but also alters the track according to the energy deposit and electron multiple scattering through stopping power ratios and scatter ration that is crucial to the optimization of calculation and to decreasing the time required for sampling.

3.1.4.2 Decrease the value of T and let σ^2 remain constant through the method of Electron Range Rejection technique and Transport Cutoff

3.1.4.2.1 Electron Range Rejection technique uses residual range of electron to determine whether the electron's movements should be stopped and whether energy should be stored in a particular region. This method works by looking at the residual range value of electron, whether it has lower values than the distance from the location of the electron at a particular time as compared to the edge of an area closest to the electron location.

3.1.4.2.2 Transport Cutoff is a method that works similarly to the threshold specification that includes two types of parameters which controls the transport and the building of secondary electron. The parameters used for the control of transport are the ECUT and PCUT technology that sets the energy value of particles. Therefore, if the energy of the particles are lower than the values that has initially been set, the next transport will not be possible. In the same way, AE and AP will be used in the production of secondary electron through the specification of the energy values of the particles. If the energy of particles are lower than that of the initially specified ones, the production of secondary electrons will not be possible.

3.1.5 Small Field Dosimetry

To ensure accurate measurement of radiation dose in a small field (FS $\leq 4\times4$ cm^2) similar to that of the measurement of radiation dose in a large field (FS $> 4\times4$ cm^2) that has been through the standard procedure of measurement and can be verified, it is crucial to know that measuring radiation dose in a small field is in the state where there is a lack of lateral CPE. The lack of CPE results from a higher lateral range of secondary electron that exceeds the field size. When measured, the radiation dose in sensitive volume is not equal to the radiation dose from lateral direction electron. Its small field size also results in the overlapping of penumbra, this then requires a small detector that does not exceed or disrupts the radiation spectrum. If the detector size exceeds the field size, the measurement value will be lower than it should, as affected by the process of volume averaging. With this reason, ionization chamber then becomes the golden standard in the measurement of radiation dose in a large field, and is not suitable for the measurement of dose radiation in a small field. Therefore, to measure the radiation dose in a small field and get the least amount of errors, it is crucial to rely upon small detectors or probes, small sensitive volume, high sensitivity, tissue equivalence, energy, and dose rate independence. However, currently there are still no detectors that have all the qualifications in one model and there are as well still no definite procedures to measure radiation dose in a small field. Small fields have been used in radiotherapy through various radiation techniques, where the field size has been specified as smaller than 4×4 cm^2 and non-equilibrium condition. The three factors required to consider or classify radiation field as small fields are:

3.1.5.1 The radiation field source can only be seen from the detector's location that passes the radiation limiter in some of the parts, resulting in lesser radiation dose value than it should be, due to overlapping at the penumbra area

3.1.5.2 The size of the detector used in measuring radiation dose is large compared to radiation field, which resulted in volume averaging effect.

3.1.5.3 The distance of electron in the control radiated, where the field is lesser than the lateral range of the particles with charges and does not result in charged particle equilibrium (CPE).

3.1.5.4 Measuring radiation dose in radiotherapy often uses ionization chamber, which is a major type of detector deemed large in size compared to field size of small fields, making it not a very appropriate alternative to measure in areas with high dose gradient, for time-dose variance, and non-uniform beam distribution that requires the estimation of detector volume and is in a non-electronic equilibrium state which relates to the distance of secondary particles that depends upon energy [30]. Therefore, the selection of detectors in small fields will require high consideration regarding the appropriateness, where AAPM TG-106 [5] has divided the detector by using active volume as a standard as follows:

3.1.5.4.1 Standard Chamber ($\leq 10^{-1}$ cm^3) is a detector with large volume, for example: Farmer-type ionization chamber, CC13, and IC 10.

3.1.5.4.2 Mini Chamber ($\leq 10^{-2}$ cm^3) is a detector with an average volume of 0.05 cm^3, for example: CC01, PinPoint, and CC04.

3.1.5.4.3 Microchamber ($\leq 10^{-3}$ cm^3) is a detector with an average volume of 0.007 cm^3 and is an appropriate detector to measure radiation dose in small fields. For example: microDiamond, SFD, EDGE, SRS, PFD, and microLion.

3.2 Review of related Literatures

In 2007, Indra J. Das [6] and crew studied radiation dose measurement under the non- equilibrium state, gathering all problems in measuring radiation dose in a small field. Due to the indefinite specification upon what the field size should be, the field size is then considered as a small field. Indra J. Das has set the size of the field smaller than 3×3 cm^2, which is thought-out as a small field. Furthermore, from the findings of the value of full width at half maximum (FWHM), it is found that the radiation source size affects the specification of the radiation spectrum. The field dose profile is measured, where small field can result in overestimated field size due to the overlapping of the penumbra.

Wolfram U. Laub and crew [7] studied the impacts from the volume effect of the detector in small field through the IMRT technique. The study applied the usage of the PTW diode type 60008 p-type Si diode, PTW Diamond detector type 60003, PTW Pinpoint ionization chamber type 31006, 0.125 Semiflex tube type 31002, and Farmer Chamber where the output factor has been measured with a field size of 1×1 to 15×15 cm^2, energy of 6 MV (Figure3.2). It is found that in the field size of over 10×10 cm^2, the Pinpoint chamber gives off higher output factor than necessary as it highly responses to the low energy Compton scatter. On the other hand, once measured in a small field, the non-water equivalent diode detector will increase the value of the secondary electron, making the output factor higher than it necessary. However, measuring with the ionization chamber will obtain lower values than it should be, as a result of the increase in lateral electron disequilibrium and volume of the detector, as stated by Wolfram U. Laub. Once the values of the output factor obtained from various detectors are compared, it is found that the volume of the detector with photon energy of 6 MV is more effective than photon energy value of 15 MV. Therefore, measuring the output factor in a small field, the spatial resolution, and water equivalent detectors are important factors. According to Wolfram U. Laub's study, the Diamond detector is the most suitable detector to measure radiation dose in a small field.

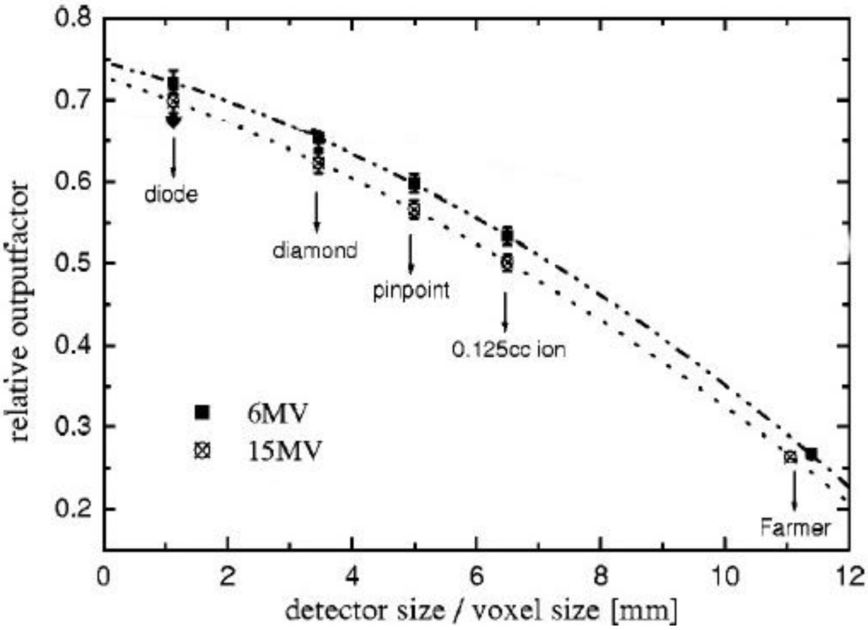


Figure 3.2 Output Factor form variable detector at field size 1×1 cm^2

Nevertheless, the advantages and disadvantages of detectors in measuring the output factor in a small field have been studied, where both De Vlamynck et al 1999, Heydarian et al 1996, Martens et al 2000, and Zhu et al 2000 have found that the Diamond and the Diode detector are both reliable in measuring the output factor for small fields. The Diamond detector is water equivalent and has high spatial resolution despite its high cost. Its reading is also based upon the dose rate (Laub et al 1997), in which the increase in dose rate will influence the diamond detector to have lower response. On the other hand, the Diode detector has high spatial resolution and linear dose response, although it is a non-water equilibrium detector that responses according to the energy.

In 2002, F Haryanto [8] and crew have conducted the study to examine the measurement of output factor for conformal radiation therapy, comparing between the measurement and the calculation with Monte Carlo technique. The measurement involves 4 types of detectors: PTW type 31002 with the largest sensitive volume of 0.125 cm^3 in this study, PTW type 60003 (Diode detector) with the smallest sensitive volume of 1 mm^2 × 2.5 μm, PTW type 60003 (Diamond detector) with a sensitive volume of 5.6 mm^2 × 0.25 mm, and PTW type 31006 (Pinpoint detector) with a sensitive volume of 2 mm diameter × 5 mm. The output factor, profile, and depth dose distribution are measured through Elekta Sli Plus with photon energy of 6 MV. The MP3 Water Phantom is used for the profile and depth dose, measuring with a field size of 10 × 10 cm^2 and depth of 10 cm, where the detector is placed to obtain high spatial resolution. In other words, the measurement of profile will rely upon the detector's location of the X axis, and the measurement of the depth dose distribution will require the detector placed on the Z axis. The measurement of the output factor will be done in a field size of 5 × 5 cm^2, as it is a size of a reference field and is closest to the conditions of a small field, where the SSD technique with a depth of 10 cm is used.

Monte Carlo Simulation technique, BEAM/EGS4 code (Nelson et al 1985, Roger et al 1995) was used with the mean energy of 6.8 MeV, ECUT of 700 keV, and PCUT of 10 keV; the DOSXYZ code in the simulation of particle transport in water phantom is also used. The calculation of depth dose distribution will require a field size of 1 × 1 cm^2 along the central axis, a depth of 0.2 cm at the build-up region, and 1 cm after the build-up area is surpassed. Furthermore, the profile nearby the area of the penumbra will require voxel resolution of 0.1 cm, while the calculation of the output factor will require voxel size of 0.1 cm and 0.5 cm, and a field size that equals to the one needed for real measurement.

Measuring the output factors with various detectors showed that field sizes of 2 × 2 cm^2 to 15 × 15 cm^2 has a relationship of not over 3% (Figure 3.3). However, field size of 1 × 1 cm^2 has an output factor that highly fluctuates, where the difference of the output factor is highest at the ionization chamber and the Diode detector according to the difference of the detectors and the water equivalence characteristic it owns. Once the ionization chamber and the pinpoint has been compared, it is seen that smaller detector size will also result in an increase lateral electronic disequilibrium. With the detector's capability of being tissue equivalent in consideration, the results from radiation dose measurement with the Diode detector and Diamond detector can prove the latter. Once the two detectors are compared, the Diode detector is less water equivalent compared to the Diamond detector. Measuring the output factor showed that the Diode detector gives off higher output factor value than the Diamond detector, due to the surrounding components of the Diode detector that lessens the electronic disequilibrium that results from the increase of lateral scattering in which is clearly seen when the field size is lesser than 2 × 2 cm^2.

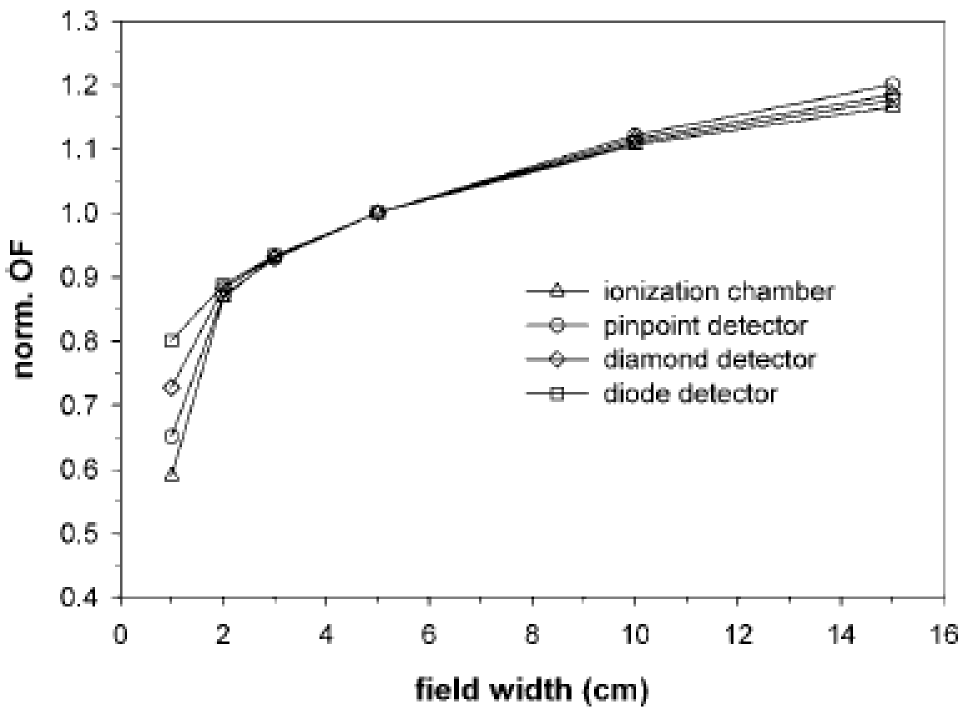


Figure 3.3 Measurement output factor by IC, pinpoint, diamond and diode detector

The evolution of radiotherapy has led to a systematic technology known as "Advance radiation therapy", where the 2D radiation therapy has been replaced. The latest radiation therapy focuses upon specifying radiation field to obtain the highest possibilities in matching the size, shape, and location of lesions to be treated in order to ensure there is least effect upon other areas surrounding the lesion, therefore resulting in smaller radiation fields.

Intensity modulated radiation therapy (IMRT) and volumetric modulated arc therapy (VMAT) are both radiation techniques that combine the usage of both small fields together with beamlet based intensity modulation [9] which differs from common radiation therapy techniques with larger fields or field sizes of 4×4 cm^2 as there is a lack of charged particle equilibrium (non-CPE), overlapping of fields over areas of penumbra, and perturbation of detector [6] as shown in Figure 3.4.

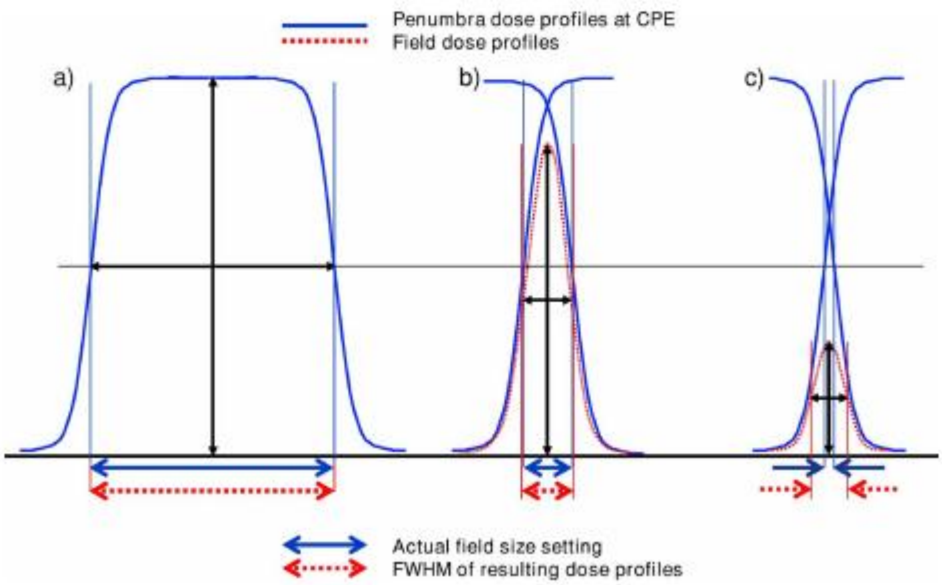


Figure 3.4 Dose profile between a) large field and b) and c) small field

Radiation detectors have been developed and modified to measure radiation dose in small fields. The characteristics have been modified to resemble human body tissues while the field size has been minimized to ensure lesser perturbation of radiation field during radiation. The study to measure radiation dose in small field relies upon the varying characteristics a detector owns (as shown in Table 3.1) and depends upon the manufacturing process of the manufacturer, therefore making the guidelines in radiation dose measurement, both in the process and the selection of an appropriate detector, uncertain. The process that serves as a guideline to measure radiation dose in small fields refers to Monte Carlo simulation in which predicts or calculates radiation dose the particle accelerators release.

Table 3.1 Comparison of the specification of detector in radiotherapy [10]

	Manufacturer	Sensitive Dimensions/mm	Potential problems
RK Chamber	Scanditronix	$\varnothing=4$, $L=10$	Size
Pinpoint chamber	PTW	$\varnothing=L=2.9$	Energy dependence—Al anode? Size?
Diamond detector	PTW	$\varnothing=2.96$, $L=0.26$	Dose-rate dependence? Size?
Shielded diode	Scanditronix	$\varnothing=2$, $L=0.06$	Energy dependence—Si? Energy dependence—W backing?
Unshielded diode	Scanditronix	$\varnothing=2$, $L=0.06$	Energy dependence—Si?
X-OMAT film	Kodak	Scanner resolution set to 0.5 mm	Energy dependence, supralinearity

Alan E. Nahum and crew [10] have studied Monte Carlo simulation to predict the characteristics of photon with small field sizes. Simulation has been done upon Varian Clinac 2100C particle accelerator with BEAMnrc which specified the PCUT value as 0.01 MeV and ECUT as 0.7 MeV, where DOSXYZnrc has been used to simulate water phantom to calculate the radiation dose that specified Voxel lateral in which gives off the best resolution along with the least time needed to calculate. Alan E. Nahum has specified the penumbra area and build up region to use voxel size of 0.5 mm × 0.5 mm × 2 mm and 1 cm × 1 cm×1cm in the central section with a depth from dmax, where the specification of the latter voxel resulted in an uncertainty value of lesser than 1%. Furthermore, examination of the particle accelerator simulation with Monte Carlo has compared measuring with field size of 3×3 and 40×40 cm^2 where small fields (0.5, 1.0, 1.5, 2.0 cm) will have the output factor calculated with voxel changed along with the detector size (1-4 mm) in which larger voxel size resulted in lesser output factor due to volume averaging as shown in Table 3.2

Table 3.2 Measurement and calculation of output factor for small fields

Field size (cm)	Monte Carlo (2 mm)	Diamond detector (3 mm)	Pinpoint chamber (3 mm)	Unshielded diode (2 mm)	Unshielded diode corrected	Shielded diode (2 mm)	X-OMAT film (0.5 mm)
0.45[a]	0.446	0.440	0.416	0.479	0.467	0.525	0.506
0.5	0.488						
1.0	0.740	0.747	0.724	0.752	0.749	0.772	0.781
1.5	0.862	0.864	0.854	0.869	0.871	0.880	
2.0	0.918	0.931	0.926	0.934	0.941	0.939	0.911
Approximate uncertainty	0.010	0.001	0.001	0.002	0.003	0.001	0.018

[a]Values are calculated with the jaw separation in the Monte Carlo calculation reduced by 0.5 mm in order to match 0.5 cm field profile measurements.

IMRT and VMAT radiation techniques work with both small fields and beamlet as earlier stated to calculate radiation dose for patients, in which requires specific information obtained from measurement during the commission. It has been found that measuring radiation dose for beam data will result in various changes depending on the detector used, for example: the differences of PDD measured with a varying detector as shown in Figure 3.5 and Figure 3.6 where it can be observed that radiation measurement under similar condition but different detector can result in various outcomes in some areas or might have different results in the entire measuring process.

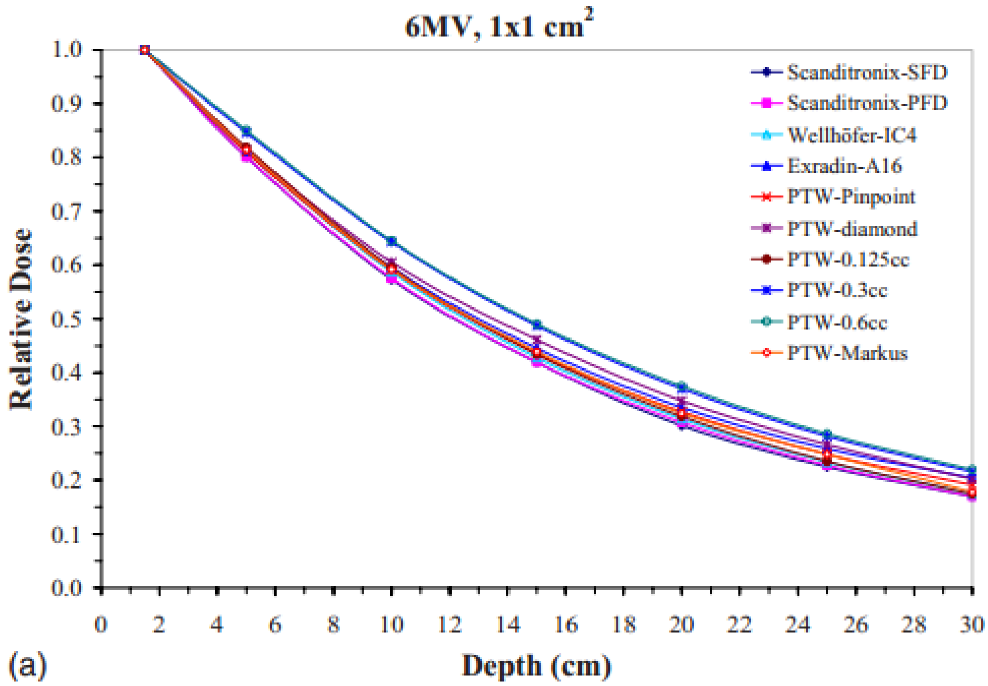


Figure 3.5 Photon 6 MV depth dose curve using different detector in field size of 1×1cm^2

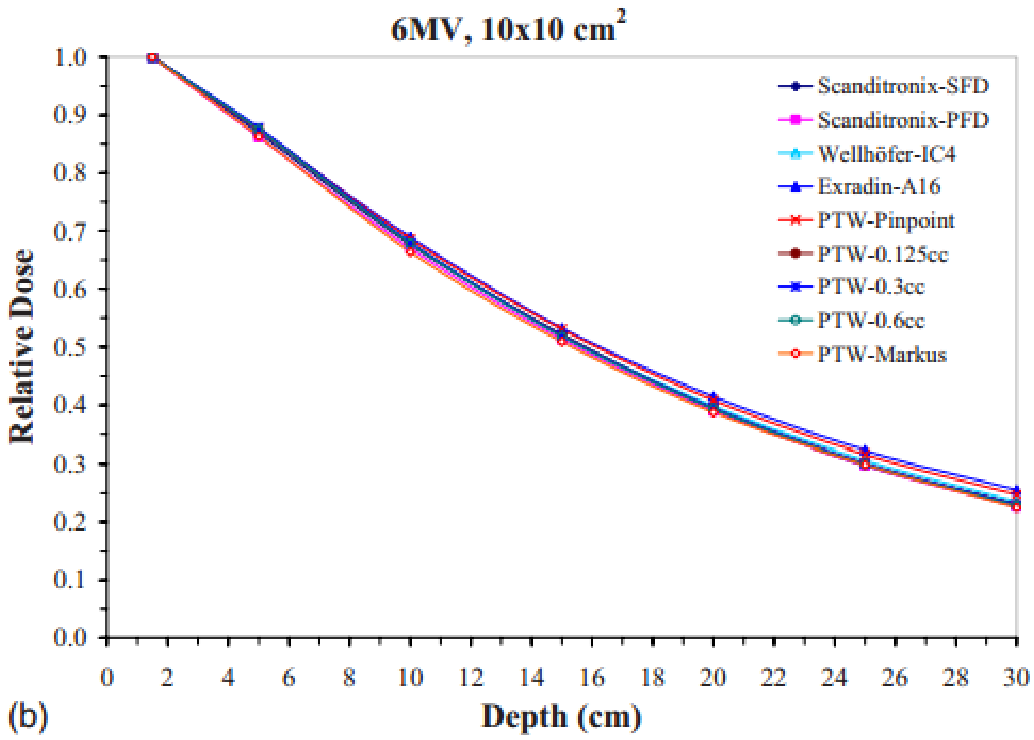


Figure 3.6 Photon 6 MV depth dose curve using different detector in field size of 10×10 cm^2

Measuring radiation dose for photon beam data normally requires IMRT to measure field size, including traditional field until 1×1 cm^2, and field size smaller than 4×4 cm^2 is considered a small field according to TG-106. Das et al's [6] study that measures radiation dose in lack of lateral electronic equilibrium small field that as well overlaps on geometric penumbra due to detector size as shown in Figure 3.7.

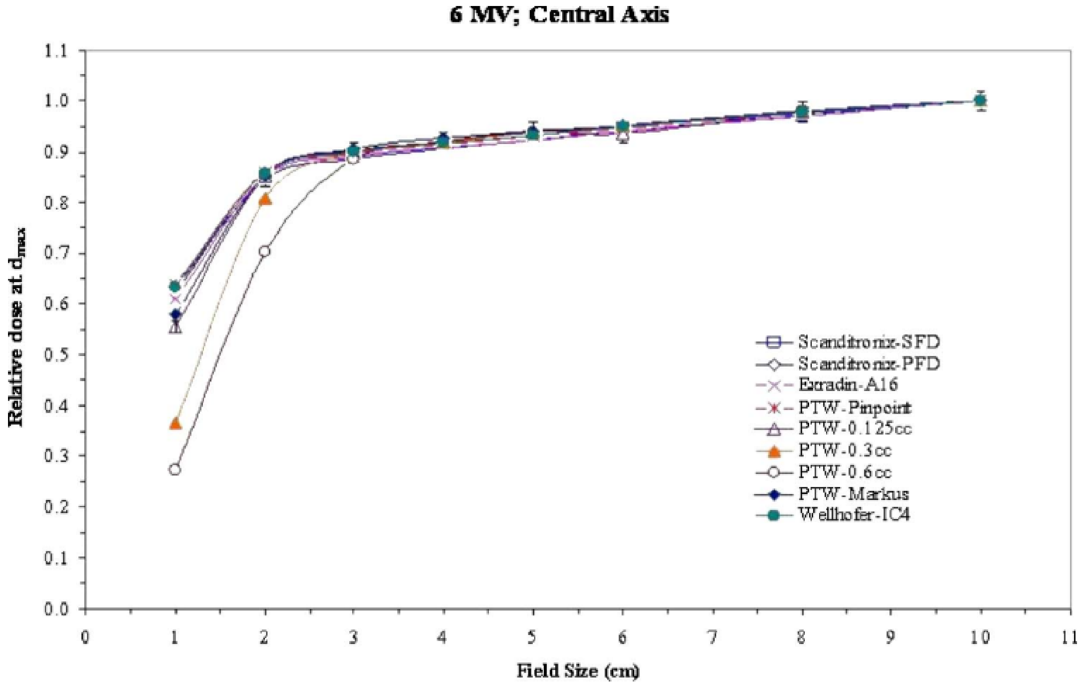


Figure 3.7 Output factor versus field size measured with various detector for 6 MV

Intra J. Das has conducted a study to measure radiation dose in small field where measuring the scatter factor (Scp) in photon beam 6MV and 15 MV has been compared with various detectors like Scanitronix SFD, PFD, Extradin A16, PTW Pinpoint, PTW 0.125, PTW 0.3, PTW 0.6, PTW Markus, and Wellhofer IC4, results as shown in Figure 3.7. It is observed that when the field size decreases in size, the measuring value decreases quickly where the semiconductor diode detector (SFD) has a sensitive volume and is considered as a micro detector, has a good spatial resolution as well as high measuring speed. Furthermore, it does not require external bias while stopping power ratios have values that almost resembles energy independence. However, problem with measuring in low energy photon has been detected due to the increase in photoelectron in the cross-sectional area of silicon when compared with water. Saini and Zhu's study 8-9 has proved that the response of the diode detector depends upon temperature, dose rate, energy, and the detector design which may affect angular dependence. Although SFD is among the interesting alternatives in measuring radiation dose in small fields, the latter has shown that it is not a golden standard material for small field dosimetry. However, a diamond detector, which is a solid state detector with high sensitivity, large signal, but small sensitive volume, along with a diamond detector with human-like tissues and properties as earlier stated, will serve as an appropriate detector pick for measuring small field dosimetry. Nevertheless, radiation dose measurement in small field still has no standard reference field size which results in differences in measurement. Intra J.Das has suggested small volume detectors, both ion chamber and diode that are used for small field dosimetry, should have the least perturbation upon radiation field and least response upon energy and dose rate. Examining the accuracy in small field dosimetry might reply upon Monte Carlo simulation to compare the measurement results and might be used to find the correction for each detector to enhance value accuracy. The study conducted by Intra J.Das have shown that a diamond detector as properties that are ideal for small field dosimetry and has been well developed to serve processes in small field dosimetry under the name of microDiamond detector.

Wolfgang and crew [11, 12] compared the output factor photon of 6 and 10 MV in small field where measuring has been done through flattening filter and free flattening filter with various detector types as shown in Table 3.3 and detectors then were divided into groups according to the specifications of TG-106 where micro detector: EFD, SFD, and micro diamond has found that SFD detector has a higher value than needed when measured in large field size compared to measuring with alamine where the value is around 1% for field size $10\times10cm^2$. The microDiamond, on the other hand, has a relating value with measuring with alamine not exceeding 0.5%. When the standard detector has been taken into consideration, it is found that when active volume increases, the output factor will therefore decrease. Wolfgang's study has used semiflex, cc13, IC10, and NPI2611 resulting in output factors of 0.811, 0.796, 0.795, and 0.612 accordingly. Nevertheless, when compared with values obtained from measuring with alamine, it is found that there is a relativity of less than 3%, where the results of measuring with diode has an over response value which as well related to numerous researches that use Monte Carlo simulation by Pantelis E [13] and Ralston A [14], where over response value might result from silicon ship with higher density than water. At the same instance, microdiamond detector compared with SFD in field size of 10×10 cm^2 resulted in microdiamond not displaying over response value in large field size but SFD does.

Table 3.3 The details of detectors used in the experiment

Category	Label	Vendor	Type	Active volume (mm^3)	Material	Z_{eff}	Sensitivity (nC/Gy)
Micro							
	microDiamond 60019	PTW	SYD	0.004	Diamond	6	0.7–1.2
	SFD	IBA	UD	0.017	Silicone	14	6
	DiodeP 60008	PTW	SD	0.03	Silicone	14	9
	EFD	IBA	UD	0.188	Silicone	14	25
	PFD	IBA	SH	0.188	Silicone	14	33
	microLion 31018	PTW	LIC	2	Wall: graphite Electrode: graphite Medium: isooctane		9.8
Mini							
	CC01	IBA	AIC	10	Wall: 0.5 mm C-552 Electrode: Ø 0.35 mm steel		0.33
	PinPoint14 310014	PTW	AIC	15	Wall: 0.57 mm PMMA 0.09 mm graphite Electrode: Ø 0.3 mm Al		0.4
	PinPoint16 310016	PTW	AIC	16	Wall: 0.57 mm PMMA 0.09 mm graphite Electrode: Ø 0.3 mm Al		0.4
	CC04	IBA	AIC	40	Wall: 0.4 mm C-552 Electrode: Ø 0.35 mm C-552		1.3
Standard							
	Semiflex 31010	PTW	AIC	125	Wall: 0.5 mm PMMA 0.15 mm graphite Electrode: Ø 1.1 mm Al		3.3
	IC10	Wellhöfer	AIC	140	Wall: 0.4 mm C-552 Electrode: Ø 1 mm C-552		4.4
	CC13	IBA	AIC	150	Wall: 0.4 mm C-552 Electrode: Ø 1 mm C-552		4.4
	NPL2611	NPL	AIC	325	Wall: graphite Electrode: Al		11

Ionization chamber, an air filled detector, gives off an under response value due to lower density, excluding CC01 which gives off over response value as a result of CC01's steel-made electrode that has a high density. Moreover, in 2014 Wolfram U [15] has studied radiation dose measurement with new commercial synthetic single crystal diamond detector (SCDD) or microdiamond detector (PTW Freiberg, Germany) Type 60019 which is a detector modified from Type 60003 to correct the detector response that depends upon dose rate. Wolfram U. has used microdiamond detector to measure radiation dose with photon of 6, 10, and 15 MV as well as electron beam with energy of 6, 9, 12, 15, and 20 MeV, which were then pre-irradiated with dose of around 8 Gy to keep the detector's response constant according to the suggestions given by the manufacturer. This study has examined the detector response to absorbed dose, energy dependence, dark current, PDD, and beam profile compared to Semiflex type 31010, microLion type 31018, P diode type 60016, SRS diode type 60018 for photon beam and Markus chamber type 23343 and E-doide type 60017 for electron beam. The results from Wolfram's study in the section of detector response has found that diamond detector type 60003 has a declining response along with absorbed dose initially where the response will remain constant when the absorbed dose is over 5 Gy as shown in Figure 3.8 and has a dark current value of around 30 pA as shown in Figure 3.9. And used about five minutes of time will be required for the dark current value to return to its constant stage or is at 2.5 pA. With microdiamond detector type 60019, when pre-irradiated at 250 cGy, was able to keep the detector response constant at around 0.1% as shown in Figure 3.10 without a need to further pre-irradiate. The results are deemed to be similar to that of Ciancagloni's study, where signal stability has a value of not over 0.5% when pre-irradiated at 60 cGy. Therefore, Wolfram has observed that by pre-irradiating at 800 cGy according to the manufacturer's suggestion is not neccessary for microdiamond detector type 60019.

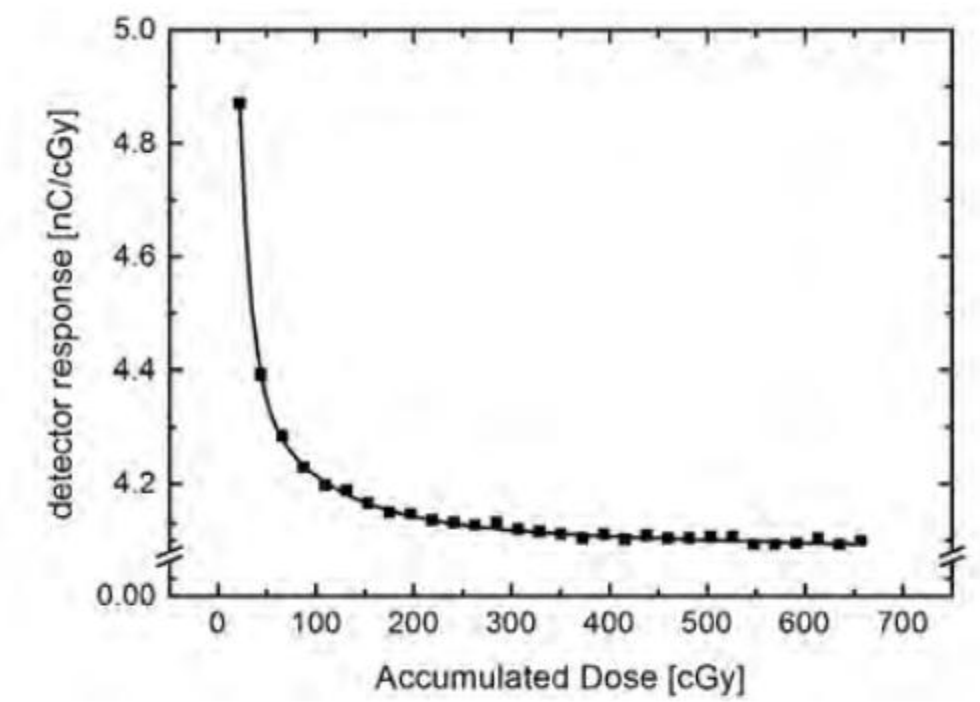


Figure 3.8 Diamond detector type 60003 versus absorbed dose

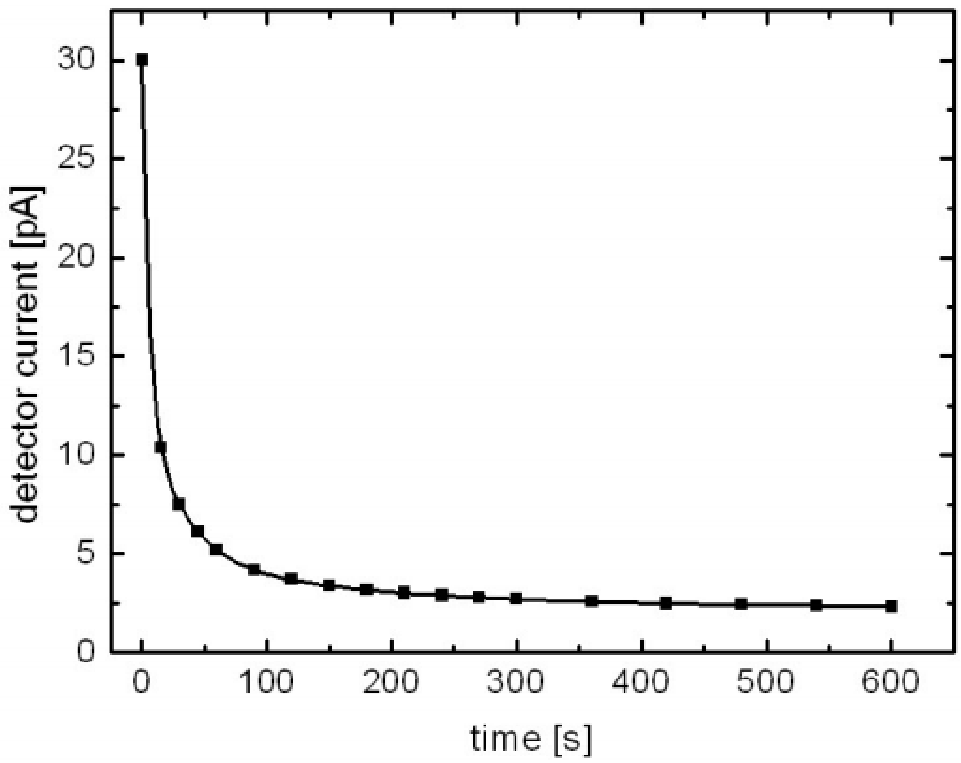


Figure 3.9 Dark current of Diamond detector type 60003

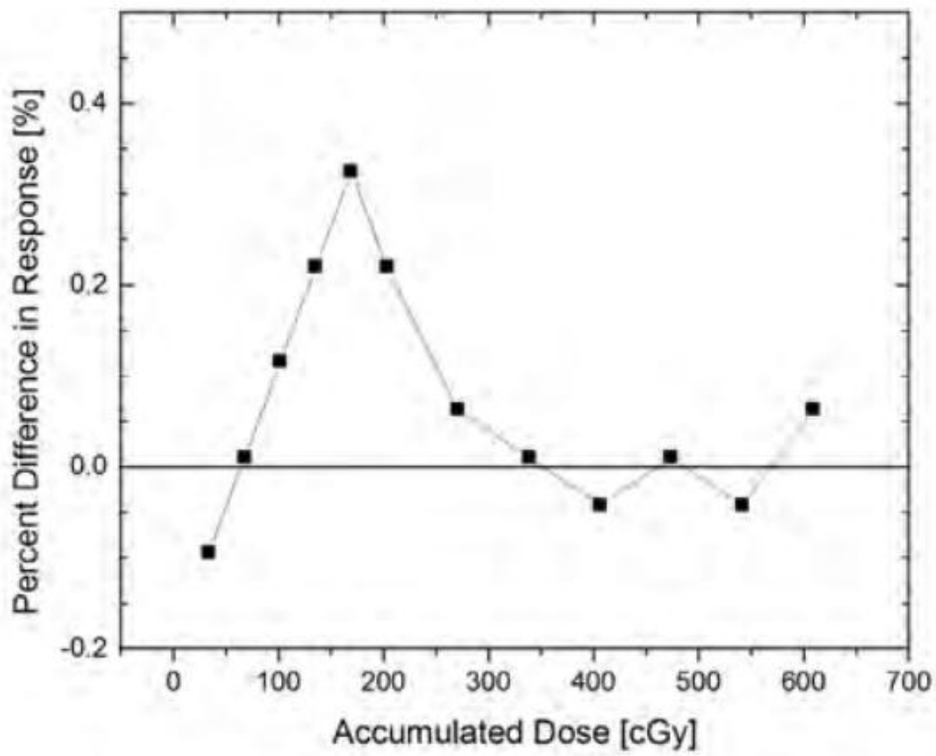
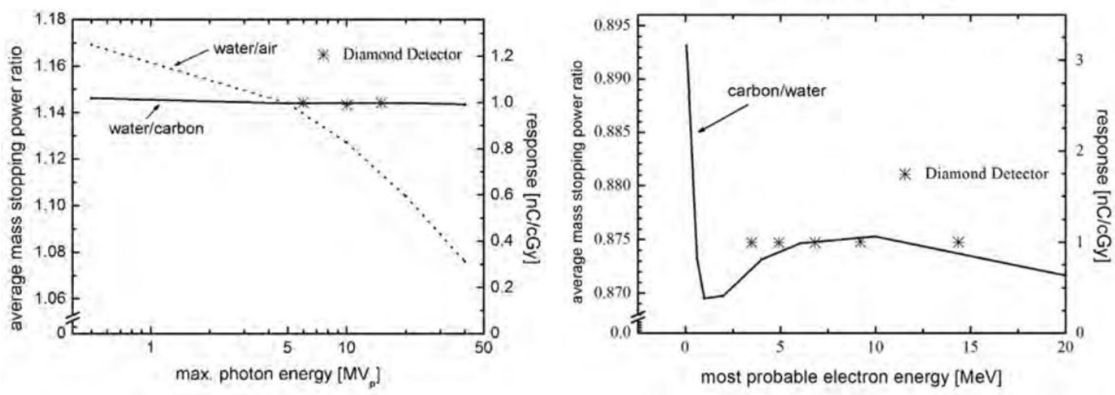


Figure 3.10 Effect of pre-irradiate for microDiamond detector type 60019

Response to photon beam with varying energy ranges (6-15 MV) of the diamond detector has a difference of not over 1% and uncertainty of measuring of 2%, while the electron beam (6-20 MeV) as well resulted in similar outcomes like those of photon beam, where the changes are not over 1%. Therefore, it is considered that microdiamond detector depends upon little energy as shown in Figure 3.11 where the part that gives off response along with energy is the housing part of the detector; when electron beam PDD were compared, it is found that the E-diode gave relating results to the microdiamond detector as shown in Figure 3.12 while Markus resulted in different values from microdiamond especially in the buildup region due to PDI curve and its change of form to a PDD curve. It can be observed that the microDiamond detector has a spatial resolution better than Markus chamber while electron profile obtained from measuring with microDiamond relates to E-diode (Figure 3.13) and Markus chamber within 1%. However, when the bremsstrahlung region is taken into account, E-diode related to Markus chamber more than the microdiamond detector.



(a) (b)

Figure 3.11 Response of microDiamond for photon (a) and electron (b)

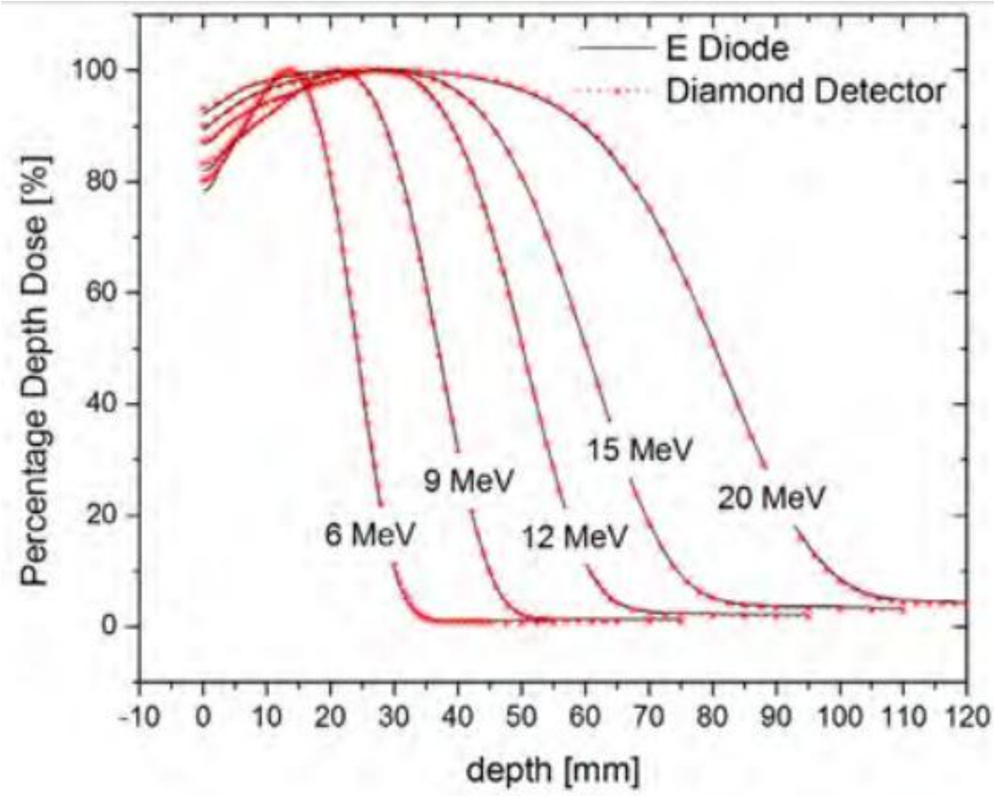
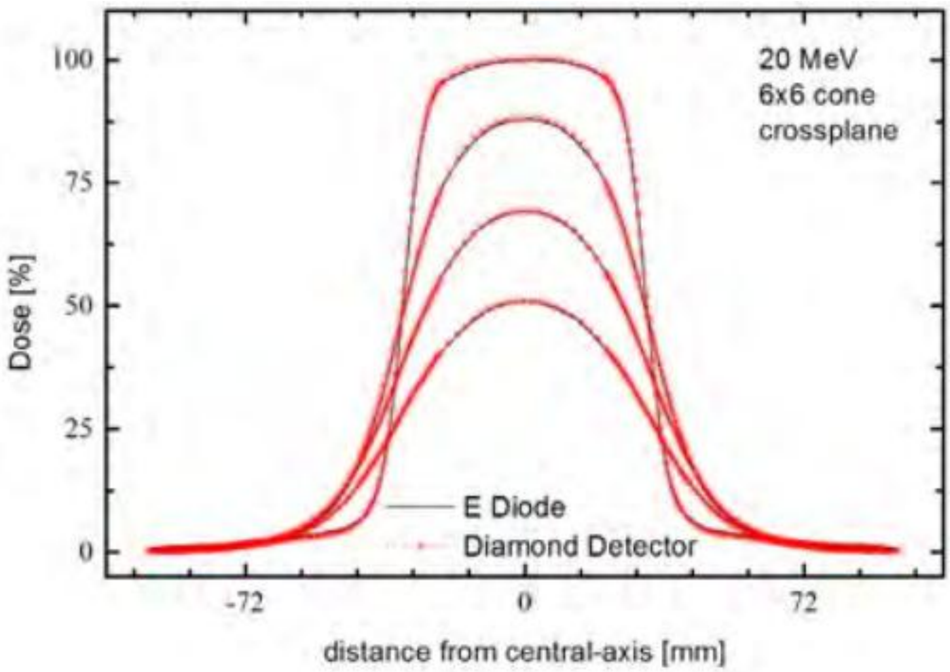
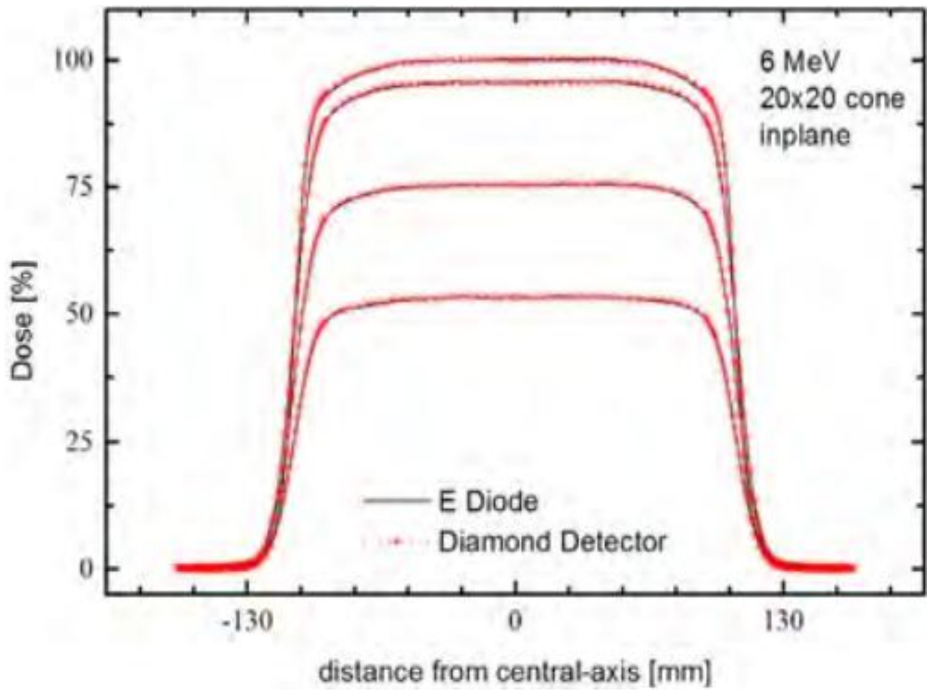


Figure 3.12 Comparison percentage depth dose between E-diode and Diamond detector

Photon beam's PDD has been found to be related to microLion in small filed size of 3×3 cm^2 and gave off relating results to those measured in large field size of 30×30 cm^2 measured with semiflex 0.125 cc not over 0.25% and a difference of 0.5-1.0% once compared to P-diode as shown in Figure 3.12. Therefore, study conducted by Wolfram U proved microdiamond detector type 60019 to work with both photon beam and electron beam, when pre-irradiated at around 300 cGy, the stability value will be of around 0.5-1.0% which is enough to measure radiation dose. Moreover, microDiamond detector as well shows that there is response to very little energy where it can be considered as having no significance and has an adequate amount of spatial resolution to measure in both small field and large field for photon beam and electron beam, making it a detector with varying properties for use although its main limitation is its high price.



(a)



(b)

Figure 3.13 (a) Dose profile for electron beams 20 MeV and 6 MeV and b) between E-diode and diamond detector

From the research of Lechner W and crew [12] experimented with measuring output factor in field size of 0.6×0.6 cm^2 to 10×10 cm^2 with 14 detector types as shown in Table 3.3. This experiment has used alanine pellets as a reference detector and has found that micro detector made from silicon will lead to over-response in large field due to silicon's thickness that exceeds that of water's properties. However, in microDiamond detector does not lead to over-response in large field and therefore can be used to measure output factor in field size of 10×10 cm^2 and up to small field. Air filled ionization chambers display under-response due to its property of being lesser thickened than water except for CC01 which has a dose response ratio close to one due to amends as a result of steel electrode's over-response. Nevertheless, this mini detector has lesser response (low sensitivity) than those of solid state and air filled ionization chambers with larger quantities. To compare micro, mini, and standard detectors, measuring output factor is required where Se An Oh and crew 8 have measured output factor in field sizes of 0.5×0.5, 1×1, 2×2, 3×3, 5×5, and 10×10 cm^2 with Novaris Linear accelerator photon 6 MV measured with CC13, CC01, and EDGE detectors in water phantom, while TLDs and Gafchromic EBT2 were measured in solid As this study has been conducted, the difference value of the output factors of 5 detectors in 2×2 cm^2 sized fields and above has a value of lesser than 2%. However, with 0.5×0.5 cm^2 field size, the output factor of EDGE obtained a closest value to EBT2. Therefore, this study has shown that measuring radiation dose in small field (smaller than 3×3 cm^2) is best with EDGE detector which has a small volume. Nevertheless, each detector differs from one another especially when used to measure in small field sizes. Silicon-made detector or diamond detector are detector to have higher efficiency in measuring radiation dose in small field compared to ionization detectors due to its small volume, and those that are diamond-made detectors are as well tissue-equivalent [16]. The microdiamond is a single crystal diamond detector (SCDD) which is a detector that combines the pros of natural diamond and silicon detectors together, making it a suitable alternative to measure radiation dose in small fields [10]. Chalkley and Heyes [16] have experimented using microdiamond PTW-60019 (V_{active} = 0.004 mm^3) in measuring relative output factors compared to diode E detector (V_{active} = 0.03 mm^3), Diode SRS (V_{active} = 0.3 mm^3), IBA SFD diodes (V_{active} = 0.017 mm^3), CC01 (V_{active} = 10 mm^3), PinPoint (V_{active} = 15 mm^3), and Semiflex

(V_{active} = 125 mm^3) with a depth of 15 mm in water where SDD was distanced as 800 mm then radiated at 100 MU with CyberKnife, field size of 5, 7.5, 10, 12.5, 15, 20, 25, 30, and 60 mm with fix collimator sizes calculating the relative output factor as displayed in Figure 3.14.

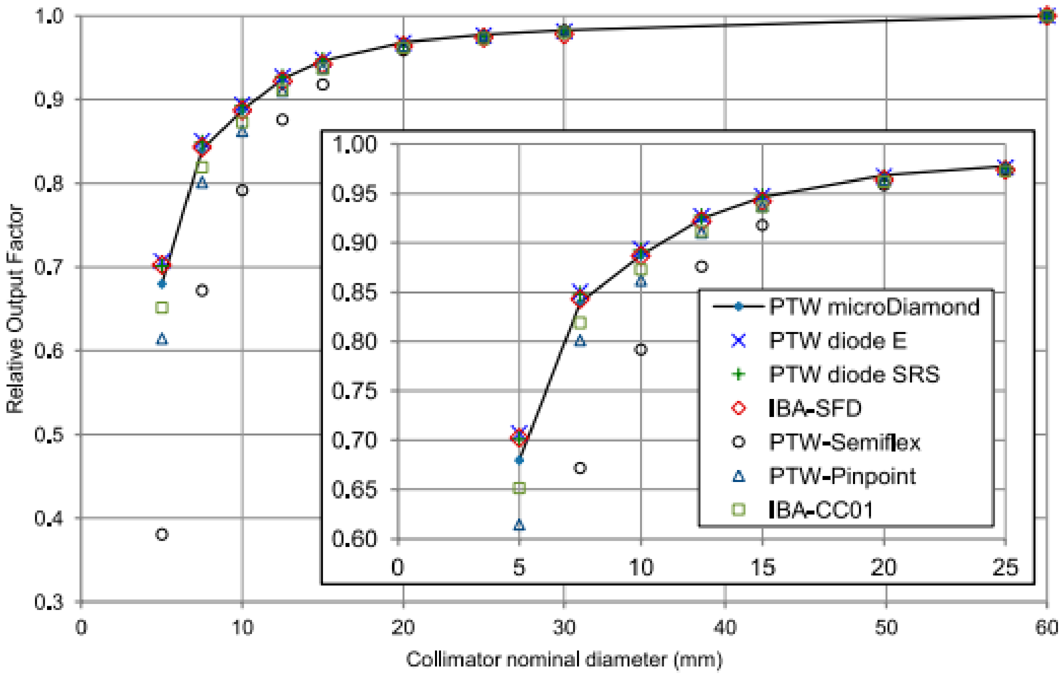


Figure 3.14 Output factors for detectors used within the study.

Then, the corrected output factors of Diode E, Diode SRS, and Pinpoint calculated by using Monte Carlo as a resource published by Franceson and crew [17] were compared with output factors obtained from measuring with microDiamond as shown in Figure 3.15.

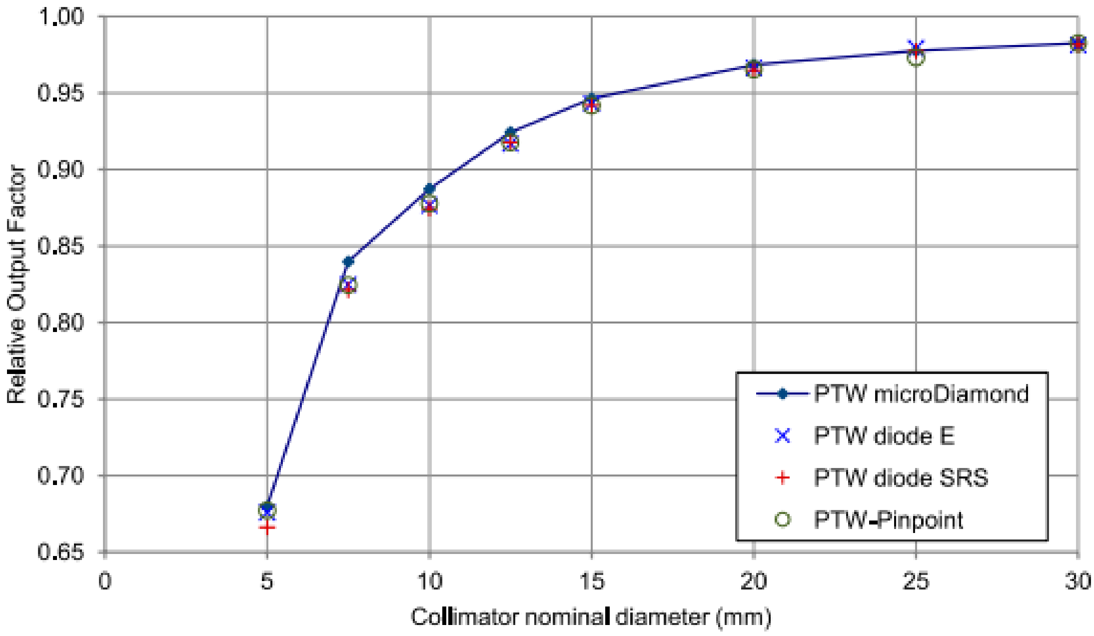


Figure 3.15 Corrected output factors which were corrected via the corrected values obtained from Monte Carlo and published by Francescon et al for Diode E, Diode SRS, and PinPoint compared to measuring with microDiamond.

From a Figure 3.15 has shown that calculating the corrected relative output factor of microDiamond compared with three detectors will result in values close to 1 more than correction factors required for Diode E, Diode SRS, and PinPoint. Therefore, correction factor used with microDiamond will have lesser value compared to other detectors as shown in Table 3.4. It can be seen that microDiamond requires lesser corrected relative output factors than other detectors. Field size of 5 mm will have microDiamond corrected relative output factor of around 1% when compared to Monte Carlo, for diodes 5%, and ionization chamber 10% which will enhance confidence in using microDiamond for measuring small fields to reduce effects of volume averaging and will not be dependent upon energy during mega voltage 2 and information will be used in treatment planning program, making the calculation results accurate.

Table 3.4 Comparison between microDiamond with corrected relative output factors of Diode E, Diode SRS and Pinpoint detectors

Ø (mm)	**microDiamond**	**Monte Carlo correction factor**		
		Diode E	**Diode SRS**	**PinPiont®**
60.0	1.000	1.000	1.000	1.000
30.0	0.999	1.000	1.000	1.000
25.0	0.999	1.003	1.002	0.997
20.0	0.997	0.999	0.998	1.001
15.0	0.996	0.995	0.994	1.005
12.5	0.992	0.989	0.990	1.008
10.0	0.988	0.981	0.980	1.018
7.5	0.981	0.970	0.968	1.029
5.0	0.990	0.956	0.950	1.102

Lechner's study have shown the properties of micro, mini, and standard detectors which proves that micro is best suited to measure radiation dose in small fields which relates to the study conducted by Se An Oh [18] , where he proved EDGE to be the best alternative in measuring output factor in small field. Although EDGE has a small volume, it is made from silicon, and therefore led to a study where Chalkley and Heyes used microDiamond with tissue-equivalent factor to measure output factor in small fields, comparing with corrected values with Monte Carlo for common detectors used like Diode E, Diode SRS, and PinPoint which gave satisfying results.

Although there are modern detectors ranging from various types to sizes (mini to micro) figures (timber, spherical, plan parallel), types (ionization chamber, semiconductor, chemical, film), but there is yet a study to provide certain guidelines for the selection of the best detector to measure radiation dose in small fields. Although there is a study on beam data to use as a golden beam data for similar radiators, similar energy, and gave results relating to the standard data from Monte Carlo simulation [19-21], however, there is yet a standard to use as a certain reference as well as an international standard. Therefore, measuring radiation dose in a small field has numerous conditions that differs from the measurement of a large field. Measuring radiation dose in a small field requires standardized procedures and provides accurate results, making it possible to enable enhanced treatment plans for the patients. Measuring in a small field requires a suitable detector that is small in size; however, there are cases where small detectors may also be unsuitable for measuring radiation dose in small fields, as its small size may also influence the acceleration of radiation measurement, making the process slower. Therefore, it is highly crucial to consider a detector that is high in sensitivity, has small volume, high spatial resolution, is tissue equivalent, is non-angular dependence, and is dose rate/energy independence. Currently there is yet a detector that has all the properties as earlier described, plus, the process of measuring radiation dose in a small field still lacks the process that determines the standard like that of the large field's. Therefore, the study of the procedure and the characteristics of the detector is required to develop accuracy and enhance the methods to specify the procedure to measure radiation dose in small fields. Along with the current radiation techniques that are continuously developing to limit the radiation spectrum upon the tissues surrounding the tumor cells like IMRT, VMAT, SRS, and SRT that has small fields, they have become widely known. The study of the impacts from radiation dose measurement including the advantages and disadvantages of each detector, also leads to continuous development of latest detectors. Often, adjustments or alternations are made to lessen the disadvantages of the initial detector and enhance the properties required to measure radiation dose in small fields. One of the developed detectors is the microDiamond detector, which altered and adjusted the disadvantages of the Diamond detector to enhance accuracy in the measurement of radiation dose. Moreover, with its characteristic as being tissue equivalent, the Microdiamond detector is currently considered the most suitable detector for the measurement of radiation dose in small fields. Nevertheless, Ramathibodi Hospital's radiation therapy measures the radiation dose in a small field with the stereotactic field detector that has a small sensitive volume and is also considered as one of the detectors that is widely accepted for measuring radiation dose in small fields. However, as there are still several disadvantages to its characteristics like being non-water equivalent, angular dependent, and energy dependent, it is an interesting topic to consider. A microDiamond detector is a developed detector aimed towards the measurement of radiation dose in small fields. It is an expensive detector which gives off different results upon the measurement of radiation dose in small fields, as compared to the SFD detector significantly. Measuring the radiation dose as a data for calculation and treatment planning for the patients will require the accurate measurement of important factors like the percentage depth dose (PDD), Beam Profile, and output factor. It is highly advised that the output factor should be the first to determine and influence the measurement of radiation dose.

Ramathibodi hospital's radiotherapy department owns Varian particle accelerator model Clinac iX which can both use intensity-modulated radiation therapy (IMRT) and volumetric modulated arc therapy (VMAT) techniques and is considered an essential tool to treat patients along with both IMRT and VMAT techniques which are for small fields as earlier stated. Still, there are controversies upon the study regarding measuring radiation dose in small fields. Ramathibodi hospital's radiotherapy department has a Stereotactic Field Diode (SFD diode) detector which provides relation in measuring with Monte Carlo simulation obtained from previous study. However, diamond detector still, is considered a detector that relates best with Monte Carlo simulation as it is an intensity-modulated radiation therapy technique alters the dose rate which affects the response of diamond detector. However, diamond detector has been modified to become smaller as its dose rate dependence has as well been decreased, introduced under the name microDiamond detector. Therefore, many are curious upon how much the microDiamond detector has an ability to measure radiation dose in small fields different from that of the Stereotactic Field Diode's (SFD Diode), leading to this study in comparing measurements of output factors with Stereotactic Field Diode (SFD Diode) and microDiamond detector as guidelines for selection and further benefits.

CHAPTER IV
MATERIALS AND METHODS

4.1 The scope of research

4.1.1 Simulation of linear accelerator Varian Clinac iX with Monte Carlo technique

4.1.1.1 Information regarding the mechanical parts of the linear accelerator Varian Clinac iX from Monte Carlo Package data obtained from Varian Medical System company studied.

4.1.1.2 Initial electron beam parameter value and full width at half maximum value (FWHM) found comparing with percentage depth dose and beam profile obtained from simulation and real process.

4.1.2 Comparing relative output factor in small fields in a homogeneous phantom (water phantom) by simulating with Monte Carlo and measuring with stereotactic field diode (SFD IBA Dosimetry) and microDiamond detector.

4.1.2.1 Relative output factor measured with Monte Carlo in small fields by simulating linear accelerator Varian Clinac iX.

4.1.2.2 Relative output factor in small fields with photon energy 6 MV calculated by measuring with stereotactic field diode (SFD) and microDiamond detector.

4.1.2.3 Relative output factor values obtained from simulation with Monte Carlo and measured with both stereotactic field diode (SFD IBA Dosimetry) and microDiamond detector compared.

This study has focused upon radiology and oncology, Ramathibodi hospital's department of radiotherapy, which has materials and equipments as follows:

4.2 Material

4.2.1 Personal Computer

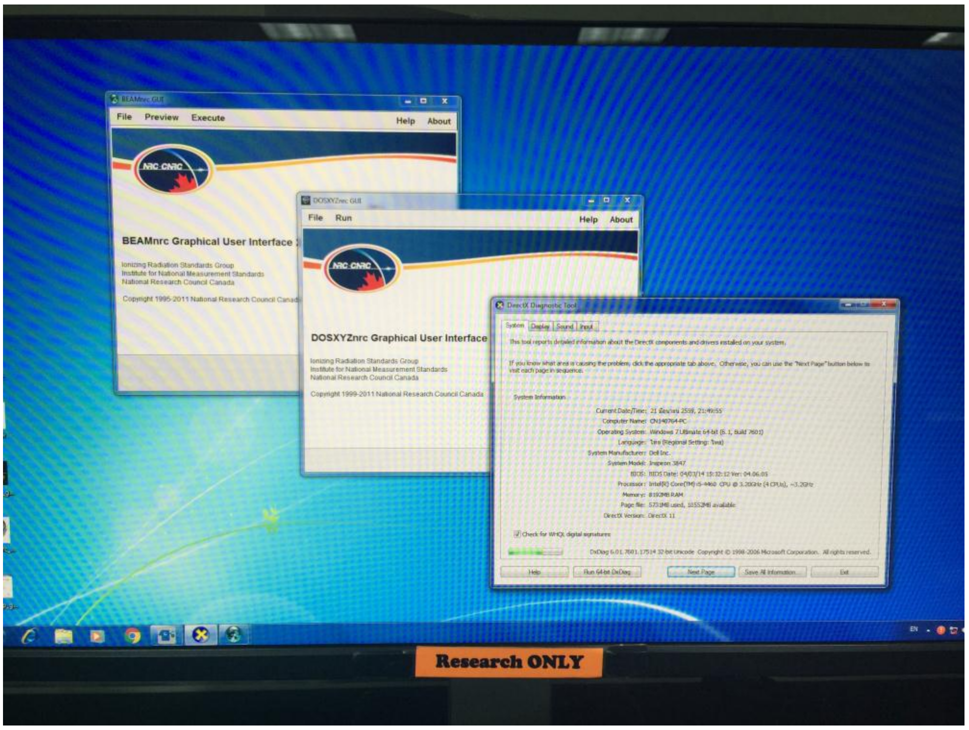


Figure 4.1 Personal Computer (PC) for Monte Carlo simulation.

A computer used in the process of Monte Carlo simulation is a personal computer (PC) branded DELL Inc, model Inspiron 3847 with windows 7 Ultimate 64 bit as its operating system and Intel(R) Core(TM) i5-4460 CPU 3.20 GHz (4CPUs) with DDR2 RAM 8192 MB, a separate display with NVDIA GeFoece GT 705 of 4049 MB, and 1 TB of memory space as shown in Figure 4.1.

4.2.2 Linear Accerelator

Figure 4.2 Varian Clinac iX linear accelerator

This study has used Ramathibodi hospital's particle accelerator Varian Oncology System, Palo Alto, USA model: Clinac iX serial number 5288 (Figure 4.2) which is a particle accelerator that can give off both photon beam of 6 and 10 MV and electron beam that gives off 4, 6, 9, 12, 16, and 20 MeV. Changes can be made to the dose rate from 100 to 600 MU per minute, field size can be opened from 0.5×0.5 cm^2 to 40×40 cm^2, distance source to axis distance equals to 100 cm, and furthermore, has an image guide system to examine the accuracy of radiating location in kV Imager known as On Board Imager, or the OBI which can take photos in a CBCT (Cone Beam Computed Tomography) mode and MV Imager, also known as Electronic Portal Imager Device or EPID where this study has focused upon the relative output factor in fields with 6 MV photon energy..

4.2.3 Radiation Detector

4.2.3.1 microDiamond Detector [22]

The microDiamond detector (PTW Freiburg, Germany) is a solid-state waterproof detector made from synthetic single crystal diamond detector (SCDD) and as a small sensitive volume of 0.004 mm^3. It is in a form of a sphere with radius of 1.1 millimeters, 1 micrometers thick, and is perpendicular to the detector axis without high voltage operation (detector bias 0 V) required. Moreover, the properties in which the manufacturer has added in energy response has no significant value for high energy photons and electrons. It has a tissue-equivalent factor that enables it to measure field size of 1×1 cm^2 up to 40×40 cm^2 both in photons of 100 keV to 25 MV and electrons of photon energy 6 to 25 MeV. With it being a synthetic single crystal diamond detector (SCDD), it owns a unique fabrication process which makes it suitable and appropriate to measure radiation dose in small fields. microDiamond detectors can measure both in photons and electrons, enabling measurements in both small (1×1 cm^2) and large (40×40 cm^2) fields, making it a multi-tasking detector that covers radiation dose measurement required in radiotherapy and its processes.

Figure 4.3 The microDiamond detector PTW-60019 (PTW-Freiburg, Germany)

4.2.3.2 Stereotactic Diode Field Detector [23]

Stereotactic Diode Filed Detector (SFD, IBA Dosimetry, Germany) is a waterproof semiconductor similar to a microDiamond detector. It has an active detector diameter of 0.6 mm and does not require a bias to operate (0 V); it has a volume of around 0.01 mm^3, sensitivity of around 4 nC/Gy compared to ionization chamber CC13 which is 130 mm^3 in volume and has similar sensitivity. It can be observed that the SFD detector is 13,000 times smaller which marks it a suitable alternative to measure radiation dose in small fields. However, as for large fields, it may be affected by scatter low energy radiation which may result in value overestimation, especially in penumbra regions. The SFD detector has a marker at the top end of the detector to specify the location during measurement where it requires aligning between the marker on the detector and linac machine's cross wire. In terms of the lifespan, sensitivity decreases when absorbed dose increases, which can be referenced from E. Grusell and G. Rikner's study [24]

4.2.4 Electrometer [25]

This study has used IBA Dosimetry's electrometer specifically model DOSE 1: High Performance Reference Class Electrometer as shown in Figure 4.4 which is an electrometer that can transport and can be used along with ionization chamber, semiconductor, and diamond probe which are all capable of measuring both the charge and current mode. DOSE 1 electrometer (serial number 17239) has a wide angle (160°) luminescent display which allows complete display of parameter values in one screen; for example: measured values, chamber, and correct factor, where the connector type is standard (triaxial TNC in combination with triaxial BNC) and supports M-Type, BNC/Banana and triaxial TNC, and triaxial BNC convertor connectors. Furthermore, the bias voltage can be adjusted from -500 to +500 V with steps of volts where DOSE 1 is 259 × 259 × 165 mm^3 (L×W×H) and weighs around 3.5 kg.

Figure 4.4 DOSE 1: High Performance Reference Class Electrometer.

4.2.5 Monte Carlo Simulation Software

Monte Carlo simulation refers to a simulation and tracking of particles as earlier explained, and there are various types. This study has used BEAMnrc and DOSXYZnrc code which is a monte carlo code working with EGSnrc system, developed by the National Research Council of Canada (NRC) and it can simulate both photons and electrons in energy of 1 keV up to 100 GeV.

BEAMnrc is used to simulate the linear accelerator's head of gantry with component modules (CMs). With information obtained from the manufacturer, important parameters in simulating with BEAMnrc includes: initial electron energy distribution and FWHM. When particles are tracked in position, energy, direction, and charge of each particle, information will be kept in a form of a phase space where it will be used as an input for DOSXYZnrc code, which is a simulation of radiation fields in various mediums according to interesting locations. The radiation field position and interesting locations will be specified in a form of voxel to obtain deposited dose at a specific voxel location. Monte Carlo software can be downloaded at http://www.iris.inms.nrc.ca/EGSnrc/EGSnrc.html. [26, 27]

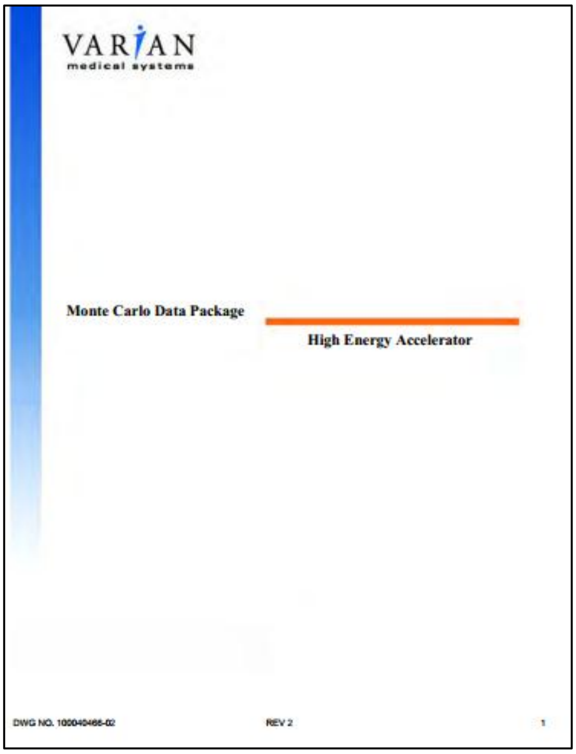
4.2.6 Beam Analysis System: Blue Phantom^2 [28]

It is a beam analyzer that both analyzes and controls the quality of a detector known as the Scanditronix Wellhofer, model Blue Phantom^2 which is made from Perspex water phantom taking up the form of a 48 × 48 × 48 cm^3. It can be used along with cylindrical ionization chamber, parallel chamber, semiconductor detector, and array detector like chamber array (CA24) and Profiler 2, where they require a specific holder of that detector.

Blue phantom relies upon the control of the computer through a control unit (CCU) and OmniPro Accept software in which this study has used CU500E and OmniPro accept version 7.1. Apart from the stated characteristics, it can control the scanning speed, the methods of scanning or information gathering in both continuous and step by step formats, and measure radiation dose: point dose, PDD, and Dose Profile (Inline and crossline) which can analyze the post processing with functions to examine and edit information including normalization, smoothing, mirror, and others.

4.2.7 Monte Carlo Package [29]

Monte Carlo simulation requires a draft of the inner composition within particle accelerators, including the size, shape, distance, and material of each part, which is a special feature of each linac model and brand. The latter information is considered essential and is not to be publicized. However, a request to the manufacturer to use the information for research and study purposes is possible. This study has simulated linac with specific information obtained from Linac High Energy according to the Monte Carlo Data Package: High Energy Accelerator DWG NO. 100040466-02 (Figure 4.5) from Varian medical system which has already specified the information upon the compositions deemed important and affecting to the attributes of the radiation field.



VARIAN
medical systems

Monte Carlo Data Package

High Energy Accelerator

DWG NO. 100040466-02 REV 2 1

Figure 4.5 Monte Carlo Data Package: High Energy Accelerator DWG NO. 100040466-02

4.3 Methods

This study has divided the research into 3 sections: simulation of Clinic iX linear accelerator with Monte Carlo code (BEAMnrc), measuring PDD and Beam profile with stereotactic field diode detector (SFD), and measuring relative output factor with SFD and microdiamond detector, including the comparison of relative output factors obtained from calculating with Monte Carlo and real-time measuring, where the details are as follows:

4.3.1 Simulation of Medical Linear Accelerator with Monte Carlo technique through BEAMnrc Code

Simulating linear accelerator with Monte Carlo code: BEAMnrc is a simulation of the mechanical compositions of the treatment head or the gantry of linac, in which there are specific components according to the linac model. This study has simulated Varian Linac's head of gantry.

According to Varian medical system: Monte Carlo Data Package: High Energy Accelerator DQG NO. 100040466-021, Clinac iX is composed of important mechanical components which affect the attributes of the radiation field, for example: target, primary collimator, flattening filter, monitor collimator, and secondary collimator. The component module in BEAMnrc code, which is an essential component of the treatment head, has been specified as shown in table 4.1, and further details can be found in the appendix of this conducted research. Nevertheless, information obtained from the manufacturer only covers the inner structure of the head of gantry and does not include the initial electron beam parameter which is considered an essential parameter required for the making of particle accelerator simulation. Important information required, in which has not been stated by the manufacturer, includes two parameters which are the energy value of electron radiation field that hits the x-ray target and the distributing value of electron radiation field that hits the x-ray target. The values can be found by comparing the attributes of photon radiation field in the part of percentage depth dose (PDD) and beam profile (at 5 cm and 10 cm deep, field size of 10×10 cm^2) as obtained from real-time measuring.

As this study has modified the field size, simulating with BEAMnrc has divided the phase space into two sections which are phase space I, located under the mirror of the particle accelerator, and phase space II, located under the secondary collimator jaw to decrease the time needed to simulate the components of the fixed component head in which phase II will be used to further calculate radiation dose with DOSXYZnrc.

Table 4.1 Components of linear accelerator simulated with BEAMnrc Code

Component Modules (CM)	Accelerator components
SLABS	X-Ray Target
CONS3R	Primary Collimator
SLABS	Vacuum window
FLATFILT	Flattening Filter
CHAMBER	Ionization chamber
MIRROR	Field light mirror
JAWS	Secondary Collimator (Y jaw and X Jaw)

Parameter values in BEAMnrc were specified with ECUT, PCUT that equals to 0.7 MeV, 0.01 MeV accordingly and has used incident particle of 1×10^{9}-4×10^{14} to ensure uncertainty value from monte carlo simulation is lesser than 1. Moreover, the efficiency of simulation with Variance reduction, Directional bremsstrahlung splitting, and Russian roulette has been enhanced, enabling the study of parameter value referenced from the EGSnrc/BEAMnrc manual [2].

Initial energy electron value can be found through the hypothesis of the unknown initial radiation field value, starting with the specification of five electron radiation field energy that hits the x-ray target which are: 5.8, 6.0, 6.3, 6.5, and 6.7 MeV accordingly [3]. Then, the distributing value of electron radiation field that hits the X-ray target has been specified as 1 mm. The transportation of particles passing through various particle accelerator heads from the target to the secondary collimator was then simulated. Information of the particles obtained from simulation will be kept in Phase Space II, then, Phase Space II value obtained from the specification of each initial radiation field value are then taken into account in order to calculate the radiation dose in phantom with DOSXYZnrc code. The phantom has been specified as $48\times48\times48$ cm^2 and the calculation of radiation dose is done in cross plane with a depth of 5 and 10 cm accordingly. Radiation dose in phantom has been calculated with Field size of 10×10 cm^2 where PDD and dose profile obtained from simulation with Monte Carlo technique has been selected to compare with the values obtained from real-time measuring. Chi square value has been used, where it will only be accepted when values obtained from simulation and values obtained from real-time measuring has the least Chi square value and when the difference value of radiation dose does not exceed ± 1%. The equation of chi square can be displayed as follows, $X^2 = \frac{1}{n}\sum_i (D_i^M - D_i^C)^2$ where the electron radiation field value that hits the target giving off Chi square value and difference value, once compared to the least measurement will be considered as the appropriate value of electron radiation field energy that hits the X-ray target.

Then, the appropriate value of electron radiation field energy that hits the x-ray target has been specified as constant and the value of the distribution of electron radiation field that hits the x-ray target has been modified from 1 mm to hypothesized values of 1, 1.2, 1.3, 1.4, and 1.5 mm accordingly. The percentage depth dose and beam profile are then calculated with phase space obtained from the modification of the value of the distribution of electron radiation field that hits the x-ray target. Then, the percentage depth dose and dose profile obtained from simulation and real-time measuring with Chi square were compared, where the value of the hypothesized distribution of electron radiation field, that gives of the least chi square value and difference value of radiation dose obtained from measurement will be considered as the value of the appropriate distribution of electron and be used as a value in simulation to further compare the relative output factor. The process of monte carlo simulation can be shown in a chronological order as displayed in Figure 4.6.

Once the initial electron beam value and a width of half the distribution (FWHM) have been obtained, simulation has been done with BEAMnrc by using parameter values obtained, where the radiation field scope has been specified as 1×1, 2×2, 3×3, 5×5, and 10×10 cm^2 accordingly then simulate the transportation of particles according to the specified parameter. The information of all particles are kept in phase space II, where phase space in each field size are calculated with DOSXYZnrc. The radiation dose calculation has been specified with depth of 5 and 10 cm at a field size equal to 1×1, 2×2, 3×3, 5×5, 10×10 cm^2 accordingly to find the relative output factor.

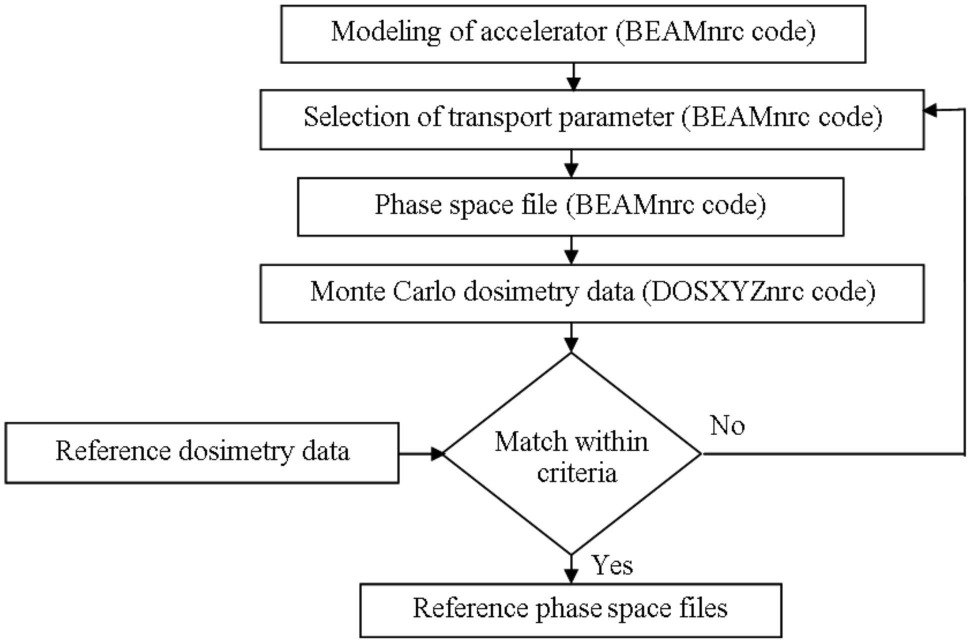


Figure 4.6 The process of Monte Carlo simulation

4.3.2 Measuring PDD and beam profile with stereotactic field diode (SFD IBA Dosimetry)

In Monte Carlo simulation requires information obtained from measuring to find initial electron beam value, and a width of half the distribution (FWHM) as topic 1 has stated. As for Ramathibodi hospital's radiotherapy department, radiation dose is measured with SFD detector in order to keep beam data collection into the treatment planning system. Therefore, this study has simulated with Monte Carlo technique by using an SFD detector as a control to find the initial electron beam value, and a width of half the distribution (FWHM). The percentage depth dose and beam profile were measured with Scanditronix Wellhofer, model blue phantom and Omnipro accept V.7 which were measured in field sizes of 1×1, 2×2, 3×3, 5×5, and 10×10 cm^2.

For percentage depth dose, it was obtained by measuring with scanning direction from downwards to upwards at a depth of 30 cm to 0 cm with scanning speed of 0.3 cm/s and scanning resolution of 0.1 cm. The latter specifications were done by Ramathibodi hospital's radiotherapy department in keeping beam data collection; measuring will be done under photon energy of 6MV, 100 cm SSD with field size of

1×1, 2×2, 3×3, 5×5, and 10×10 cm^2 accordingly. On the other hand, beam profile required similar specification with measuring percentage depth dose where the process involved a cross plan format scan with 5 and 10 cm depth in each field size; percentage depth dose and dose profile obtained from field size 10×10 cm will be used to compare with monte carlo simulation to find initial energy electron and FWHM.

4.3.3 Measuring relative output factor

This study has measured the relative output factor by using blue phantom along with external electrometer, model DOSE 1. Measuring has been done with photon energy of 6 MV, 100 cm SSD, 100 MU, and has specified gantry and collimator's location at 0 degrees. Then, SFD detector's location has been specified, where the sensitive volume of chamber is upon the water's surface and the center of chamber perpendicular to the central axis with parallax set up technique (as shown in Figure 4.7). Next, the chamber level has been modified to have depths of 5 and 10 cm, where the radiation field scopes sizes of 1×1, 2×2, 3×3, 5×5, and 10×10 cm^2 have been opened accordingly, where each location / position has been measured 5 times, temperature pressure modified, and the average value has been calculated as a result of measuring 5 times, similar to measuring with microdiamond detector.

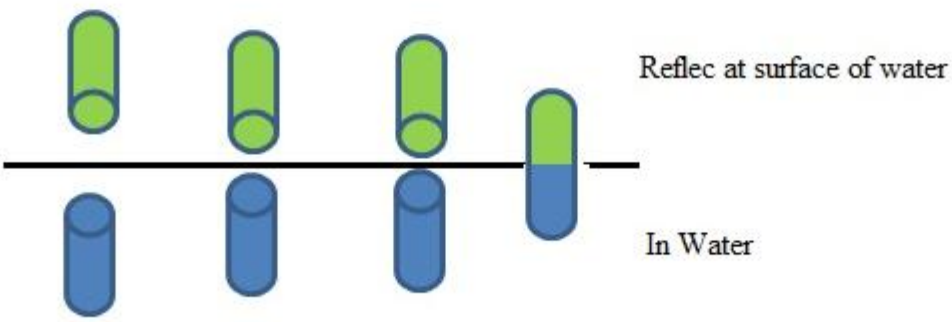


Figure 4.7 Parallax Method

The relative output factor has been found from the ratio of an interesting radiation dose scope to radiation dose of radiation field size of 10 × 10 cm^2 with similar depth according to the ROF equation. Once relative output factor has been obtained from measuring with both SFD and microDiamond, the values obtained from measuring and calculating with Monte Carlo has been compared. The percentage difference according to the equation has been used where it has been specified to have a difference percentage of lesser than ±2 percent, and will be considered acceptable.

CHAPTER V
RESULTS AND DISCUSSION

5.1 Simulating Medical Linear Particle Accelerator Simulation with Monte Carlo through BEAMnrc Code

Simulating Linac, model Clinac iX with energy of 6 MV has been done through hypothesized parameter values to find initial energy electron and FWHM by setting up hypothesized the value of initial electron radiation field energy that hits the target and the distribution value of electron radiation field energy that hits the target, which are displayed in Table 5.1 and 5.2.

Table 5.1 Percentage difference obtained from the comparison between real-time measured values with SFD detector and values obtained from Monte Carlo simulation, where FWHM values of initial electron equals to 1 were hypothesized. Further, it has modified the value of energy electron between 5.8 MeV to 6.7 MeV in field size 10×10 cm^2

Initial electron beam (MeV)	% Difference ± SD
5.8	1.435 ± 1.53
6.0	1.108 ± 1.37
6.3	1.016 ± 1.11
6.5	1.054 ± 1.19
6.7	1.072 ± 1.24

Table 5.2 Percentage difference obtained from the comparison between SFD detector real-time measured values and values obtained from Monte Carlo simulation that has hypothesized energy electron value equals to 6.3 and modified FWHM value from 0.4 mm to 1.7 mm in field size 10×10 cm^2.

FWHM (mm)	**% Difference ± SD**
0.4	2.57 ± 1.38
0.6	1.94 ± 1.05
0.8	3.75 ± 1.98
1	3.87 ± 2.04
1.3	3.42 ± 1.92
1.7	4.40 ± 1.86

From Table 5.1 and 5.2 which has found that the energy electron value and FWHM value equals to 6.3 and 0.6 mm, which are the values that give off the least percentage difference compared to real-time measuring with SFD detector. Therefore, the latter stated parameter values have been used to simulate in BEAMnrc in order to find relative output factor for the comparison with values obtained from measuring with SFD and microDiamond

5.2 PDD and Beam profile with Stereotactic Field Diode (SFD)

Scanning has been done to measure percentage depth dose and beam profile with SFD through Blue phantom along with Omnipro Accept software by measuring radiation photon energy of 6 MV and SSD 100 cm with field size and parameter setting according to Ramathibodi hospital's radiotherapy department method of storing beam data collection. Percentage depth dose and beam profile at a depth of 1.5, 5, 10, 20 and 30 cm can be shown in figure 5.1 and 5.2 accordingly

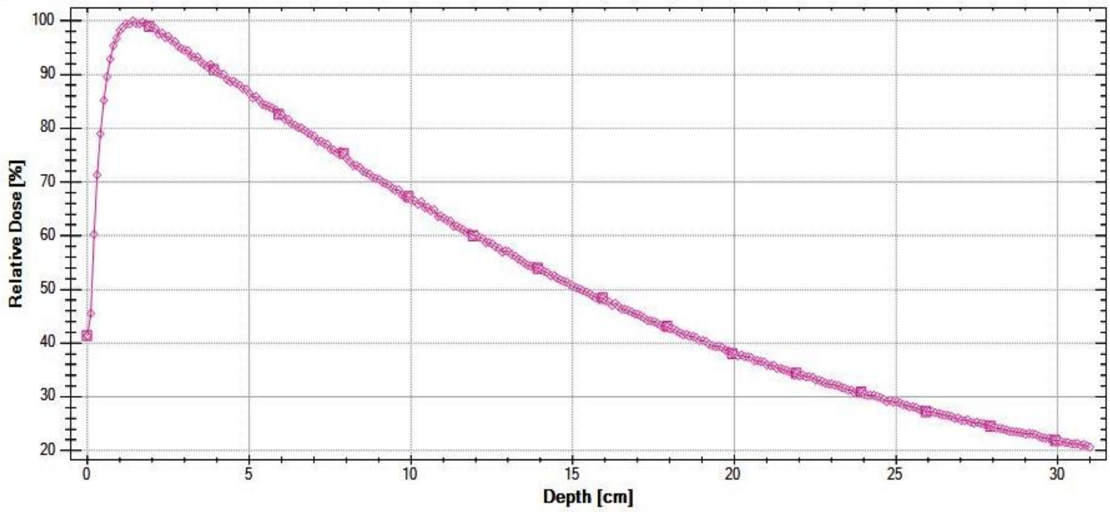


Figure 5.1 Percentage depth dose of 6 MV photon energy, 10x10 cm^2 field size, and SSD of 100 cm measured with SFD detector.

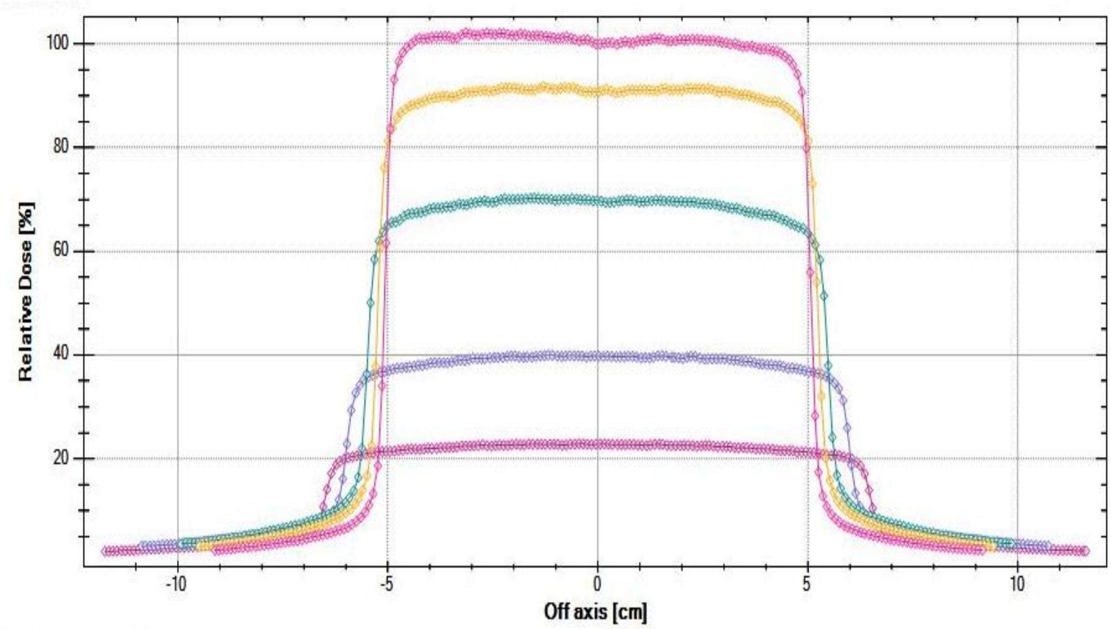


Figure 5.2 Beam profile of 6 MV photon energy, 10x10 cm^2 field size, and SSD of 100 cm at depths of 1.5, 5, 10, 20 and 30 cm measured with SFD detector

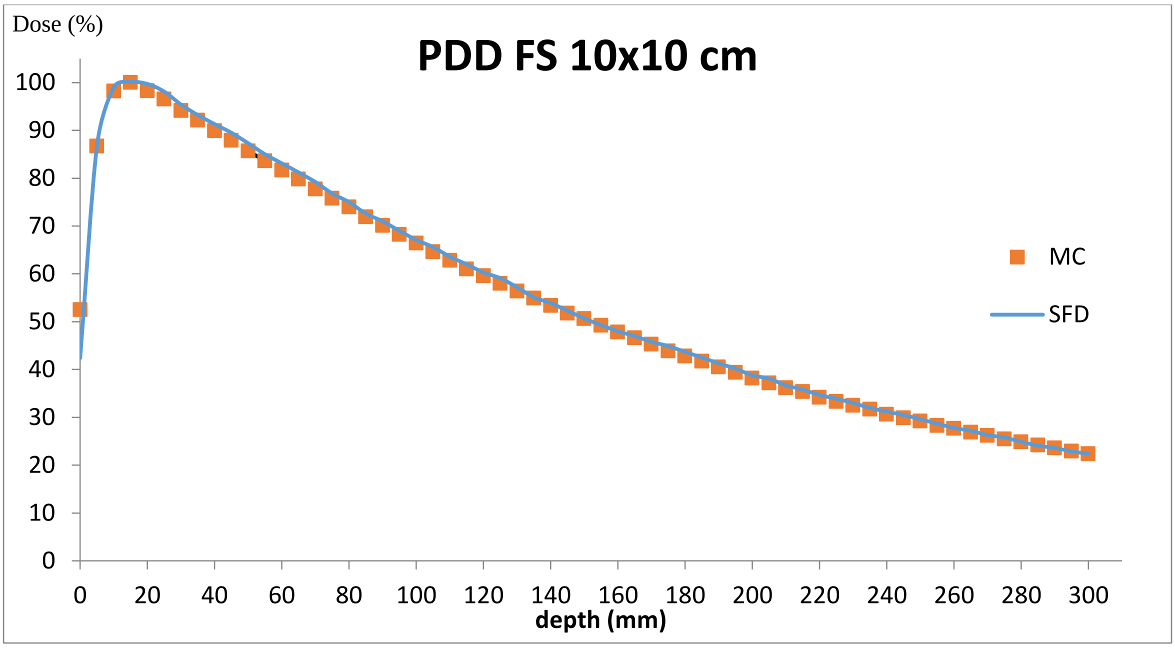


Figure 5.3 Percentage Depth Dose, comparison between MC and SFD for 6 MV photon beam, 10×10 cm^2 field size

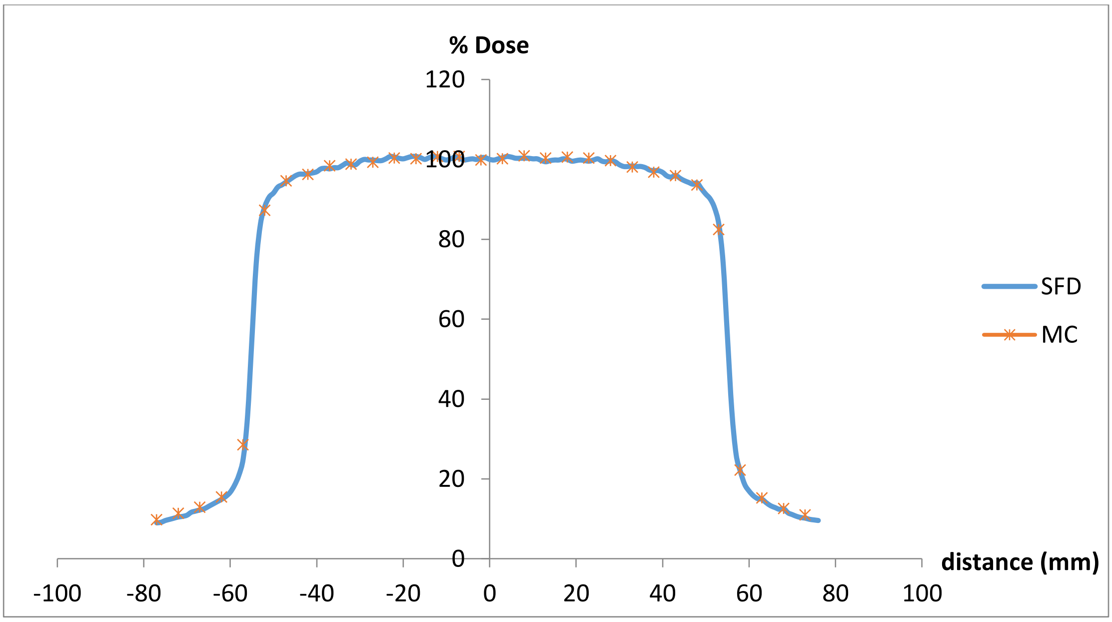


Figure 5.4 Beam profile, comparison between MC and SFD for 6 MV photon beam, 10×10 cm^2 field size, 10 cm depths

5.3 Relative Output Factor

It is found that measuring output factor value with microDiamond detector resulted in fluctuations in reading measured value, where the value increased and decreased within 5 attempts of measuring. Therefore, the monitor unit value has been increased from 100 MU to 200 and 300 MU, giving off relative output factor as shown in Table 5.3 and Figure 5.5

Table 5.3 Relative output factor values obtained from measuring with microDiamond detector with photon energy of 6 MV through the monitor unit that equals to 100, 200 and 300 MU.

FS	100MU	200MU	300MU
1	0.698	0.705	0.706
2	0.868	0.776	0.790
3	0.947	0.816	0.831
5	0.900	0.882	0.894
10	1.000	1.000	1.0000

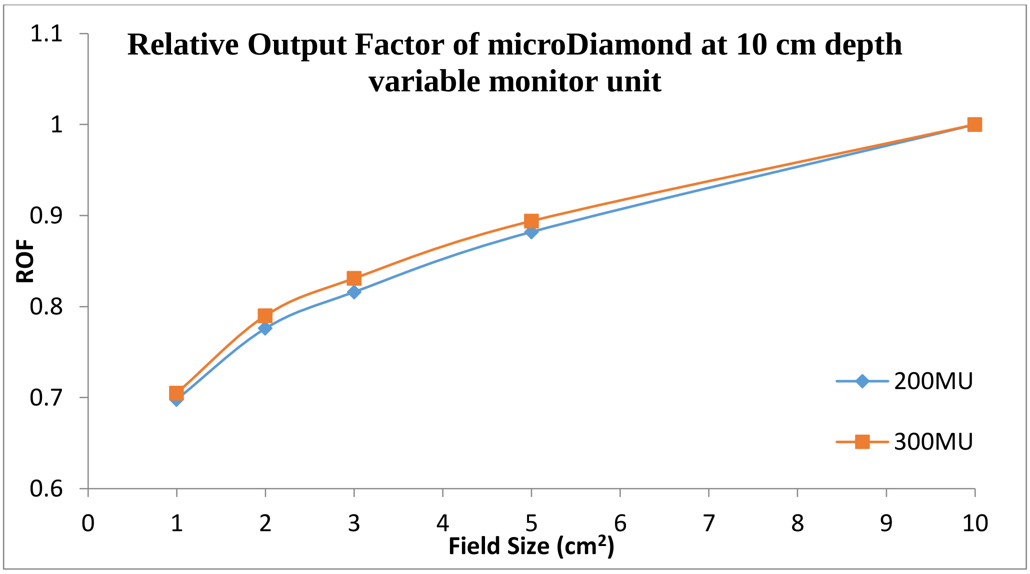


Figure 5.5 Relative output factor of microDiamond at 10 cm depth variable monitor unit

The comparison between relative output factor value obtained from calculation with Monte Carlo simulation and real-time measuring (SFD and microDiamond detector) is displayed in Table 5.4 and Figure 5.6 and 5.7

Table 5.4 Relative output factor values at 5 cm depth compared between measuring and calculating with Monte Carlo simulation.

Type	**Relative Output Factor (SD,%diff)**			
	1×1 cm^2	**2×2 cm^2**	**3×3 cm^2**	**5×5 cm^2**
SFD	0.7574 (0.68,1.57)	0.8786 (0.54,1.30)	0.9140 (0.72,1.35)	0.9373 (0.42,1.15)
microDiamond	0.7586 (0.57,1.73)	0.8799 (0.51,1.45)	0.9115 (0.64,1.08)	0.9381 (0.38,1.24)
Calculation (MC)	0.7457 (1.15, -)	0.8673 (0.72, -)	0.9018 (0.52, -)	0.9266 (0.47, -)

Table 5.5 Relative output factor values at 10 cm depth compared between measuring and calculating with Monte Carlo simulation.

Type	**Relative Output Factor (SD,%diff)**			
	1×1 cm^2	**2×2 cm^2**	**3×3 cm^2**	**5×5 cm^2**
SFD	0.6995 (0.85,1.48)	0.8243 (0.70,1.78)	0.8669 (0.75,1.59)	0.9052 (0.58,1.27)
microDiamond	0.7028 (0.77,1.96)	0.8113 (0.66,1.73)	0.8645 (0.63,1.31)	0.8920 (0.42,1.29)
Calculation (MC)	0.6893 (1.26, -)	0.8099 (0.84, -)	0.8533 (0.77, -)	0.9037 (0.51, -)

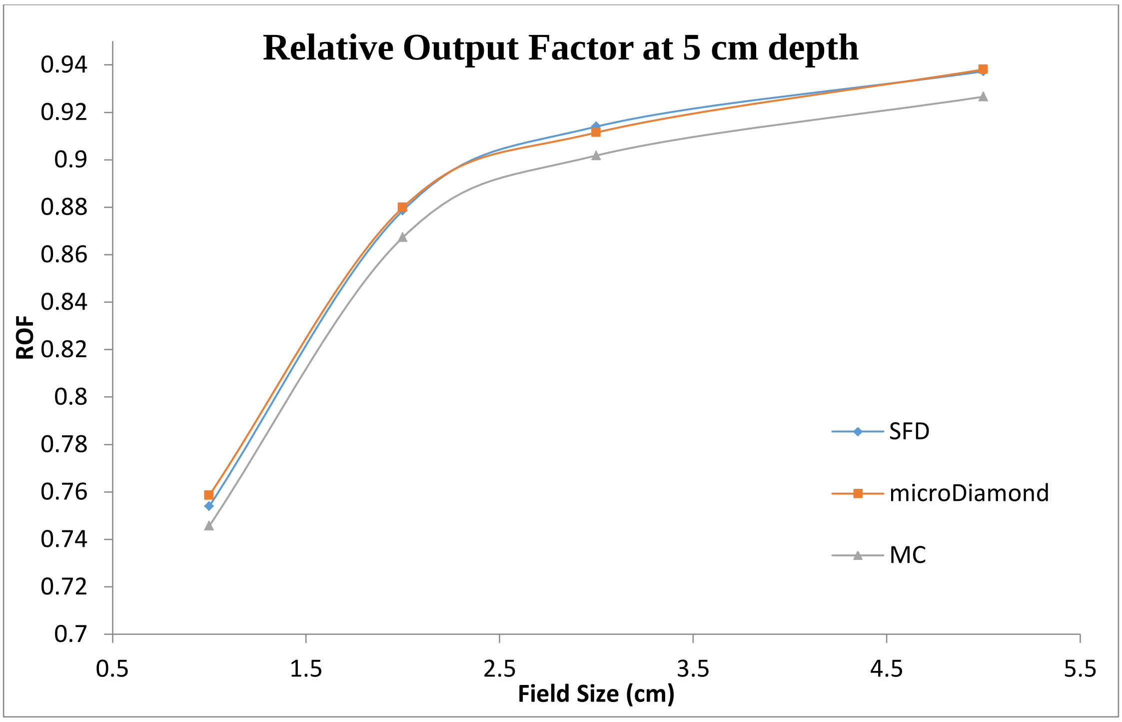


Figure 5.6 Relative output factor at 5 cm depth compared SFD, microDiamond and Monte Carlo (MC)

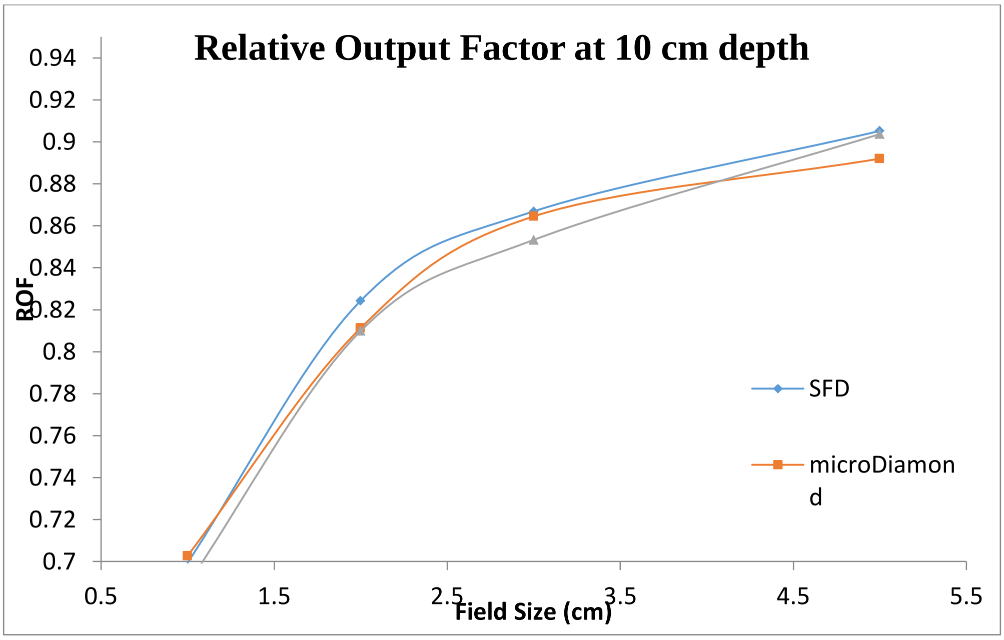


Figure 5.7 Relative output factor at 10 cm depth compared SFD, microDiamond and Monte Carlo (MC)

Furthermore, the capabilities of the detector in measuring radiation dose in penumbra, which requires a high spatial resolution detector as it is an area with high radiation dose difference, have been compared. The penumbra width has been measure at depths of 5 and 10 cm, which resulted in values displayed in Table 5.6.

Table 5.6 Comparison of penumbra width measurement from SFD, microDiamond detector, and calculation from Monte Carlo simulation (MC) (only for field size of 1×1 cm^2 and 5×5 cm^2)

Detector	**Penumbra Width (mm)**							
	Depth 5 cm				**Depth 10 cm**			
	1×1 cm^2	**2×2 cm^2**	**3×3 cm^2**	**5×5 cm^2**	**1×1 cm^2**	**2×2 cm^2**	**3×3 cm^2**	**5×5 cm^2**
SFD	2.74	2.95	3.19	3.36	3.46	3.97	4.21	4.63
microDiamond	3.52	3.88	4.29	4.68	3.81	4.09	5.19	5.44
MC	2.62	-	-	3.28	2.97	-	-	3.87

Measurements performed to find the most suitable value for simulation through Monte Carlo simulation technique showed that the best energy of initial electron beam and radial intensity distribution width is 6.3 MeV, and a width of half the distribution (FWHM) which is 0.6 mm, as it does not exceed 1% of the difference between measurement and simulation. Once values obtained from the measurement of depth dose and dose profile (depths of 5 cm and 10 cm) in fields between 1×1 cm^2 and 5×5 cm^2 with SFD detector and microDiamond detector have been compared with the values obtained from Monte Carlo simulation technique, the results from the measurements of both detectors match where its value does not exceed over 2%.

Once the value obtained from relative output factor at an energy of 6MV and F.S. between 1×1 cm^2 and 5×5 cm^2 measured with the SFD detector and microDiamond detector has been compared with Monte Carlo simulation's measurement, the difference of the values did not exceed 2% as illustrated in table 5.4. However, with the penumbra in consideration as shown in Table 5.5, it is found that the microDiamond detector has a slightly wider penumbra (≈1mm) than that of the

SFD detector and Monte Carlo simulation for all field sizes consistent with A Chalkley (2013)[16]

The values of the penumbra width obtained from measuring with microDiamond are higher than those measured with SFD although the sensitive volume of microDiamond is lesser than that of SFD detector's. Which then could result from cross section of microDiamond (2.2 mm in diameter), larger than SFD housing which resulted in an increase size of microDiamond. Therefore, although there is a small sensitive volume, with its physical attributes larger than that of SFD detector's, interference of photon fluence[10] increases accordingly.

To measure radiation dose with ion chamber with an inner diameter of 4-6 mm, it is recommended to use field size of not smaller than 4×4 cm^2. However, for fields smaller than 4×4 cm^2, it is recommended to use microchamber[5] detector, which is often a diode detector. Nevertheless, diodes are higher density than water, and therefore will result in over-response in large radiation fields whereas microDiamond is no result in over-response[6] as it is a detector with density similar to water.

Apart from the selection of a detector, it is important to ensure that the detector is located at the center of the radiation field and that adequate amount of time is needed during radiation to lessen the effects of noise within a detector small in size. Cross-calibration is required to measure radiation dose in small fields with mini or microchamber detectors, where ionization chamber like the semiflex 0.125 cm^3 is used to measure 4×4 or 5×5 cm^2 sized radiation fields and perform an equivalence test upon the wanted/desired detectors in small fields[2].

Along with Monte Carlo simulation, this study has no intentions to focus upon the comparison of the impacts resulting from materials that make up the detectors, including the results of SCDD's water equivalence, and therefore has not simulated the components of both detectors. Information used to modify the parameter values in BEAMnrc in order to simulate Clinac iX Linac were obtained from real-time measuring with SFD detector, which may be one of the reasons that led to both the output factor and penumbra width values measured with SFD detector closely similar to those calculated with monte carlo and slightly more than the microDiamond detector. With this reason, this study still has doubts and will continue the research to simulate the components of the detector and may further simulate with values obtained from real-time measuring with SFD and microDiamond detector in modifying the parameter values of BEAMnrc, which might provide guidelines upon accurate measuring of radiation dose in small field dosimetry as well as to specify clear guidelines upon the process of measuring and using detectors in small fields.

On other hand, in this study the validation of Monte Carlo simulation was compared by measurement data from SFD at 10×10 cm^2 only. The SFD might be over-estimated for large field size as states by Wolfram U. Laub (2003). Therefore to decrease uncertainty and increase accuracy in small field dosimetry, smaller field size than 10×10 cm^2 should be validated for Monte Carlo simulation as well.

CHAPTER VI
CONCLUSION

This study has showed that microDiamond detector (PTW Freiburg, Germany, PTW-60019), a synthetic single crystal diamond detector with a small sensitive volume of 0.004 mm3 (smaller than that of SFD detector's which is: 0.017 mm3) and is tissue equivalent, owns appropriate characteristics in the measurement of the output factor in small fields although is high in price, which is one of the factors that questions its usage. The results of this study have revealed that the measurement of output factor in small fields result in same values obtained from SFD detector, with a difference not exceeding 2%. On the other hand, microDiamond detector has a lesser sensitive volume than that of SFD detector's, but a slightly wider penumbra width than that of SFD's. At any rate, this study aims to compare the values of the output factor obtained from the measurement of both detectors with the calculation. Moreover, this study does not involve the study of specific characteristics of the microdiamond detector, therefore, the information obtained from this study may not be enough to point the advantages and disadvantages, the necessity upon purchases, replacement, or encourage promotional agencies the detectors in use.

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