Monolithic Silicon Transmission Pixelate Detector for Small Field Dosimetry

Kananan Utitsarn

University of Wollongong

UNIVERSITY OF WOLLONGONG

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Recommended Citation

Monolithic Silicon Transmission Pixelate Detector for Small Field Dosimetry

Kananan Utitsarn

This thesis is presented as part of the requirements for the conferral of the degree:

Doctor of Philosophy

Supervisor: Professor Michael Lerch
Co-supervisor: Distinguished Professor Anatoly Rosenfeld

The University of Wollongong
Centre for Medical Radiation Physics, Faculty of Engineering and Information Sciences

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This research has been conducted with the support of an Australian Government Research Training Program Scholarship.
DECLARATION OF ORIGINALITY

I, Kananan Utitsarn, declare that this thesis, submitted in fulfilment of the requirements for the award of Doctor of Philosophy, in the Centre for Medical Radiation Physics, Faculty of Engineering and Information Sciences, University of Wollongong, is wholly my own work unless otherwise referenced or acknowledged. This document has not been submitted for qualifications at any other academic institution.

Kananan Utitsarn
December 20, 2017
ABSTRACT

In recent years new radiotherapy systems have emerged that are utilized for small tumour treatments with improvements to enable improved dose coverage of the target. The treatment is accurate but still would benefit from a real-time treatment monitoring with high spatial and temporal resolution sampling for regular quality assurance (QA). A monolithic silicon diode array, the Magic Plate 512 (MP512) was developed as a potential candidate for such QA. The detector was design for use as an in-phantom 2D dosimeter and 2D transmission mode detector for real-time dose measurements.

The first part of this thesis evaluates the radiation response of the Magic Plate and the impact of an air gap immediately above the MP512. This air gap is then optimize for using the MP512 for small field dosimetry in both photon and electron fields. The output factor (OF), percentage depth dose (PDD) and enhanced dynamic wedge (EDW) beam profiles were measured as a part of these studies. The optimized air gap is then taken into account in the later chapters that focus on in-phantom dosimetry using the MP512. MP512 response reduces with increasing air gap above the detector. The OF measured with MP512 with air gaps of 0.5 mm and 1.2 mm show a good agreement with OF measured with the EBT3 film (within ±2%) and MOSkin for 6 MV and 10 MV, respectively. Similar results were observed for the PDD measurement. The EDW dose profile matched well with the EBT3 for the air gap of 0.5 mm within ±2% (1 standard deviation) for all wedge angles. The PDD measured by electron beams demonstrated no significant effect of the air gap size above MP512 for all energies.
The second part of this thesis demonstrates the use of MP512T as a transmission detector. The influence of operating the MP512T in transmission mode (TM) on the surface dose of a phantom was evaluated as a function of different field sizes and distances from the solid water phantom to transmission detector (D_{sd}). For all D_{sd} and all field sizes, the MP512T led to the surface dose increasing by between 5% and 25% when in the beam, depending on the configuration. The transmission factor of the MP512T ranged from 1.020 to 0.9950 for all measured D_{sd} and field sizes.

The last part of this thesis showed the correlation of transmission mode response (TM) and dose mode response (DM) of the Magic Plate512 (MP512T) for different detector to surface distances (D_{sd}) and treatment field sizes. The measured correlation between TM and DM was then employed to predict the dose at d_{max} for regular fields, and intensity modulated fields. The calculated dose for regular fields of 1 x 1cm$^2$ and 4 x 4cm$^2$ fell in the range of [-2.18% and +1.95%] compared to the measured dose. For the calculated IMRT planar dose at d_{max} and gamma criteria of 3%/3mm and 2%/2mm pass rates of 98.14%/90.5% and 97.22%/93.8% were found when compared to the dose predicted by the TPS for D_{sd} 4 and 24cm, respectively. Good agreement was also observed for these gamma criteria when comparing TM measurements taken at D_{sd} 4 and 24cm with EBT3 yielding pass rates of 96.89%/92% and 97.53%/93.8%, respectively.

The thesis therefore ultimately demonstrates that the dose in the phantom can be calculated based on TM measurements and these data represent the first step in the development of real-time high spatial resolution 3D dose reconstruction technique based on TM measurements from the MP512.
I would like to acknowledge all those who deserve a special mention for their assistance and support in the development of this work over the last four years.

First, Professor Michael Lerch who had been my main supervisor. Professor Michael always monitored me and provided wonderful advice throughout my time as his student.

Many thanks to Distinguished Professor Anatoly Rosenfeld who has provided me with a patient guidance and encouragement. Distinguished Professor Anatoly Rosenfeld always responded to my questions and queries so promptly.

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In my work, I often have to make special phantom of the holder for the detector. I was glad the UOW workshop staff always there for technical help and suggestions. Thank Martin, Ron, Stuart and Andrew for making the detector movable stand and the detector cover with a big and small size that I needed.

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LIST OF PUBLICATION

Peer review journal papers and conference proceedings


LIST OF CONFERENCE PRESENTATIONS

1. **K. Utitsarn, Z.A. Alrowaili, M. Lerch, M. Petasecca, M. Carolan, A.B. Rosenfeld**


9. T. Al-Sudani, M. Metcalfe, D. Cutajar, **K. Utitsarn**, A. Ceylan, J. Davis, A.B. Rosenfeld, “Dose build up characteristics with eXaSkin bolus during 6MV radiotherapy: MoSkin dosimetry results” presented at Micro-Mini & Nano-
Dosimetry and innovative technologies in radiation (MMND-ITRO conference),
6-11 February 2018, Queensland, Australia.
<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
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<tbody>
<tr>
<td>2D</td>
<td>Two dimensional</td>
</tr>
<tr>
<td>3D</td>
<td>Three dimensional</td>
</tr>
<tr>
<td>AAPM</td>
<td>American association of medical physics</td>
</tr>
<tr>
<td>ADC</td>
<td>Analogue-to-digital</td>
</tr>
<tr>
<td>AIMS</td>
<td>Area integrating energy fluence monitoring sensor</td>
</tr>
<tr>
<td>ASTRO</td>
<td>American society for radiation oncology</td>
</tr>
<tr>
<td>CAX</td>
<td>Central axis</td>
</tr>
<tr>
<td>CF</td>
<td>Calibration factor</td>
</tr>
<tr>
<td>cGy</td>
<td>Centigray</td>
</tr>
<tr>
<td>CMRP</td>
<td>Centre for Medical Radiation Physics</td>
</tr>
<tr>
<td>CPE</td>
<td>Charge particle equilibrium</td>
</tr>
<tr>
<td>CT</td>
<td>Computed tomography</td>
</tr>
<tr>
<td>DAQ</td>
<td>Data acquisition system</td>
</tr>
<tr>
<td>CAX</td>
<td>Central axis</td>
</tr>
<tr>
<td>DM</td>
<td>Dose mode measurement</td>
</tr>
<tr>
<td>Dsd</td>
<td>Detector to surface distances</td>
</tr>
<tr>
<td>d_{max}</td>
<td>Depth of maximum</td>
</tr>
<tr>
<td>EBRT</td>
<td>External beam radiation therapy</td>
</tr>
<tr>
<td>EDR</td>
<td>Extended dose rate</td>
</tr>
<tr>
<td>EDW</td>
<td>Enhanced dynamic wedge</td>
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<tr>
<td>EPI D</td>
<td>Electronic portal imaging device</td>
</tr>
<tr>
<td>FPGA</td>
<td>Programmable gate array</td>
</tr>
<tr>
<td>FWHM</td>
<td>Full width at half maximum</td>
</tr>
<tr>
<td>Abbreviation</td>
<td>Full Form</td>
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<tr>
<td>--------------</td>
<td>-----------</td>
</tr>
<tr>
<td>IAEA</td>
<td>International atomic energy agency</td>
</tr>
<tr>
<td>IC</td>
<td>Ionisation chamber</td>
</tr>
<tr>
<td>ICCC</td>
<td>Illawarra cancer care centre</td>
</tr>
<tr>
<td>ICRP</td>
<td>International commission on radiation unit and measurement</td>
</tr>
<tr>
<td>ICRU</td>
<td>International Commission on Radiological Protection</td>
</tr>
<tr>
<td>IQM</td>
<td>Integral quality monitory system</td>
</tr>
<tr>
<td>IMRT</td>
<td>Intensity modulated radiation therapy</td>
</tr>
<tr>
<td>keV</td>
<td>Kilo electron volt</td>
</tr>
<tr>
<td>LCPE</td>
<td>loss of lateral charge particle equilibrium</td>
</tr>
<tr>
<td>Linac</td>
<td>linear accelerator</td>
</tr>
<tr>
<td>MeV</td>
<td>Mega electron volt</td>
</tr>
<tr>
<td>MLC</td>
<td>Multi leaf collimators</td>
</tr>
<tr>
<td>MOSFET</td>
<td>Metal-Oxide Semiconductor Field Effect Transistor</td>
</tr>
<tr>
<td>MP121</td>
<td>Magic Plate 121</td>
</tr>
<tr>
<td>MP512</td>
<td>Magic Plate 512</td>
</tr>
<tr>
<td>MRI</td>
<td>Magnetic resonance imaging</td>
</tr>
<tr>
<td>MU</td>
<td>Monitor unit</td>
</tr>
<tr>
<td>MV</td>
<td>Megavoltage</td>
</tr>
<tr>
<td>OD</td>
<td>Optical density</td>
</tr>
<tr>
<td>OF</td>
<td>Output factor</td>
</tr>
<tr>
<td>OSLD</td>
<td>Optical stimulated luminescent dosimeter</td>
</tr>
<tr>
<td>pC</td>
<td>Pico Colum</td>
</tr>
<tr>
<td>PCB</td>
<td>Printed circuit board</td>
</tr>
<tr>
<td>PDD</td>
<td>Percentage depth dose</td>
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<tr>
<td>PMMA</td>
<td>Poly methyl methacrylate</td>
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<tr>
<td>Acronym</td>
<td>Description</td>
</tr>
<tr>
<td>---------</td>
<td>-------------</td>
</tr>
<tr>
<td>PTV</td>
<td>Planning target volume</td>
</tr>
<tr>
<td>QA</td>
<td>Quality assurance</td>
</tr>
<tr>
<td>RGB</td>
<td>Red green blue</td>
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<tr>
<td>SBRT</td>
<td>Stereotactic body radiation therapy</td>
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<tr>
<td>SD</td>
<td>Standard deviation</td>
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<td>SRS</td>
<td>Stereotactic radiosurgery</td>
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<tr>
<td>SSD</td>
<td>Source to surface distance</td>
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<tr>
<td>TF</td>
<td>Transmission factor</td>
</tr>
<tr>
<td>TLD</td>
<td>Thermoluminescent dosimeter</td>
</tr>
<tr>
<td>TM</td>
<td>Transmission mode measurement</td>
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<tr>
<td>TPS</td>
<td>Treatment planning system</td>
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<td>UV</td>
<td>Ultraviolet</td>
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<tr>
<td>V</td>
<td>Voltage</td>
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<tr>
<td>VMAT</td>
<td>Volumetric modulated radiation therapy</td>
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<td>WHO</td>
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CHAPTER 1

INTRODUCTION

Any radiotherapy treatment aims to give a high radiation dose to the tumour but as low as possible to the surrounding healthy tissue. This aims to provide the maximum chance of curing or shrinking a cancer while minimizing the risk of side effects. Many studies demonstrate that such a high dose improves the outcome of tumour control, and a corresponding low dose reduces normal tissue toxicity [1]–[4]. Advanced radiation treatment techniques such as IMRT, VMAT, and SRS/SBRT have been used for delivering a high a conformal radiation dose to the cancer. These advanced radiotherapy techniques use a computer-controlled linear accelerator and typically consist of many intensity-modulated treatment fields which are incident from numerous different beam directions. Due to its complexity, high precision radiotherapy planning checks before any patient can start the treatment is required.

The use of real-time dosimetry verification for a complicated treatment plans is recommended [5]. This verification aims to immediately detect errors that may occur during treatment delivery. High spatial resolution, real-time dosimetry devices are therefore needed due to the complex radiation fields that make up the treatment plan.

The Centre for Medical Radiation Physics (CMRP) at the University of Wollongong has developed a two-dimensional monolithic silicon diode array called Magic Plate 512 (MP512) with high spatial resolution. The detector is designed for use in both in-phantom dosimetry and transmission measurements. Both modes of operation of the MP512 will be characterized and optimized as part of the work presented in this thesis.
Chapter 1: Introduction

1.1 Thesis outline

This thesis describes the use of the monolithic silicon diode array Magic Plate 512 for dose mode measurement and transmission mode measurements. Chapter 2 discusses the relevant literature related to the work presented in this thesis. Chapter 3 describes the design and fabrication of these two systems. Chapter 4 evaluates the impact of an air gap on the MP512 response and optimization of this gap for in-phantom dosimetry. The device is then characterized for its performance in small and standard treatment radiation fields. The optimized air gap size from this chapter is then used for in-phantom dose measurements in chapter 6. Chapter 5 investigates the effect on surface dose, as a function of different field sizes and distances from the solid water phantom to transmission detector (D_{sd}), of using the monolithic silicon detector MP512T in transmission mode. In addition, the transmission factor for the MP512T and the printed circuit board (PCB) were also evaluated. Chapter 6 investigates the correlation of transmission mode response (TM) and dose mode response (DM) of the MP512T for different detector to surface distances (D_{sd}) and treatment field sizes. The measured correlation between TM and DM was then employed to calculate the dose at d_{max} for regular fields. A clinical application using intensity modulated radiation fields was used to evaluate this correlation. Chapter 7 presents the overall conclusions arising from the results and data found in studies presented in this thesis. The advantages, limitations and the future work of this device are also discussed.
CHAPTER 2
LITERATURE REVIEW

This chapter aims to provide an introduction to some of the advanced radiation treatment techniques, including IMRT, VMAT, SRS and SBRT. An extensive background on the dosimeters for both pre-treatment verification, real-time treatment verification and small field dosimetry is presented.

2.1 Radiation Therapy

According to a study released by the World Health Organization (WHO), cancer is the leading cause of death worldwide with the incidence increasing at a rate of 2% each year from 2012-2017 [6], [7]. The three most used methods for this lethal disease treatment, include surgery, chemotherapy and radiotherapy. It is estimated that more than 50% of cancer patients would benefit from radiotherapy at some stage of their treatment course [8].

Radiotherapy uses a high energy of ionising radiation to treat tumours. Ionizing radiation works by damaging the DNA of cancerous tissue leading to cellular death [9]. The beam can be delivered with various type of ionising particles such as electrons, photons (X-ray), gamma (Gamma Knife) and protons [10]. Several methods are used to deliver radiation to the patient. External beam radiation therapy (EBRT) or teletherapy is the most common method of radiotherapy. EBRT uses a linear accelerator to generate high radiation dose delivery to a target from outside the patient body [11]. Other methods are internal
radiation therapy or brachytherapy where the radioactive source is introduced directly into a tumour [12], but this method is outside the scope of this thesis.

EBRT aims to maximise the radiation dose to the cancer cells while sparing normal healthy tissue surrounded [13]. Some modern radiation treatment techniques have been developed to achieve the treatment goal, such as intensity modulated radiation therapy (IMRT), volumetric modulated radiation therapy (VMAT), stereotactic radiosurgery (SRS) and stereotactic body radiation therapy (SBRT).

### 2.1.1 Intensity Modulated Radiotherapy (IMRT)

IMRT delivers non-uniform radiation fluence to the target through various directions of the treatment beams from several gantry angles [14], [15]. With multiple beams, high doses can be delivered to the target volume, especially in the curvature shape and low dose to the critical organs [16]. To modulate the intensity of radiation, the change in the Multi-Leaf Collimator (MLC) position in the field is optimized by the inverse treatment planning system (TPS) based on advanced computing calculation [17], [18]. Inverse TPS for IMRT has been described in detail and can be found in the literature [19]–[22].

IMRT has two methods of treatment delivery including step and shoot technique and sliding window technique. With the step and shoot technique, the beam is on when the MLC are stable and allow multiple segments per field to be given. The beams stay still when the MLCs are changed from one segment to another. The sliding window technique keeps the beam on while the MLCs move through the irradiated field [23], [24].

### 2.1.2 Volumetric Modulated Arc Therapy (VMAT)

VMAT technique is developed from the IMRT technique. While IMRT delivers the radiation with static gantry, in VMAT the gantry is rotated around the patient for one or
more arcs continuously [25]. A number of parameters are varied during the treatment such as MLC shaping, dose rate, gantry speed and MLC orientation [26]. VMAT, in principle, is able to provide higher conformal dose compared to IMRT dose delivery because this technique uses all angles that are available in the inverse TPS to optimise the dose distribution [27]–[29]. The variation of gantry speed and dose rate enhances the significant advantage of VMAT and provides the shortest treatment time [30]. This reduces the effect of patient movement and intra-fraction motion in between radiation delivery. However, the physical constraints of the linear accelerator should be considered, such as maximum gantry speed, maximum leaf speed and the MLC orientation constraints [31]. Additionally, the angular dependence of dosimeter can be present, which must be corrected for verification technique [32]. Studies suggest the use of VMAT rather than IMRT for the complex treatment areas such as the head and neck region, prostate region and nasopharynges region [33]–[35].

2.1.3 Stereotactic radiosurgery (SRS) and stereotactic body radiation therapy (SBRT)

SRS and SBRT have been used for more than ten years to eliminate benign and malignant lesions. The SRS had been developed to treat small brain tumours as well as functional abnormalities of the brain that could not be surgically removed [36]–[39]. These treatment techniques were initially developed to treat a small tumour which is usually less than 4 cm [40]. SRS delivers in a single treatment of fraction but in SBRT the treatment dose delivered in a few focused radiation treatments, typically one to five fractions. These treatments lead to a shorter overall course of treatment time compared to other treatment techniques resulting in a reduction in radiation-related biological effects [41].
Chapter 2: Literature Review

SRS and SBRT are complex due to the use of small field delivery techniques using IMRT or VMAT for SBRT employing either high definition MLCs and/or small circular cones in the case of SRS to fit a tumour [42]. This technique also optimises the gantry angle and weights with multiple isocentre or dynamically shaping the field during gantry rotation by mini or micro MLC [43], [44].

The challenge for SRS and SBRT is to deliver accurately and precisely, high radiation dosage to a small area target and minimise the dose to normal tissue. Thus, a specialised planning and treatment delivery technique is needed. The treatment planning verification is required to ensure that the patient will receive accurate and precise radiation treatment [45], [46].

2.2 Quality Assurance (QA) and current QA tools

2.2.1 Introduction

The complexity of IMRT, VMAT and SRS/SBRT requires the precision of radiation delivery [47]. The treatment would not be effective if the tumour received a radiation dose less of than the prescribed dose. Additionally, the patient would develop radiation sickness if the normal healthy tissue and the vital organ received more than the radiation dose [48]. To ensure that the calculated radiation dose from the TPS is matched well with the dose delivered to the patient, the dose distribution needs to be accurately verified before, during or in between treatment fraction delivery [49]. Thus the treatment verification becomes an important part of radiotherapy to provide a safe radiation delivery and consistency in patient outcomes [50].

The treatment plan can be verified by transferring the patient plan to a phantom, measured using a dosimeter and compared with the calculated or predicted dose by the TPS at the
same points [51], [52]. The verification that mostly operates a day before the first
treatment fraction in the absence of the patient is called pre-treatment verification. This
procedure can detect some errors such as incorrect positioning of the MLC leaves; an
incorrect plan exported from the TPS to the Linac; or any accidental changes occurring
in the plan [53]. Thus, those errors can be corrected before the implementation of the
clinical patient treatment plan.

Various type of detectors have been used for pre-treatment verification; for example, there
is point dose measurement systems such as ionisation chamber (IC), a semiconductor
detector and metal-oxide semiconductor field-effect transistor (MOSFET) [54]–[57].
However, in a complex dose distribution, such as that used in advanced treatment
technique, point dose measurements is unsuitable as it requires multiple measurement
points for a treatment plan verification. A two-dimensional (2D) dosimetry techniques
such as Film dosimetry, 2D array detectors have been produced for measuring energy
fluence or absorbed dose in two dimensions [58], [59]. Additionally, to detect and
measure the dose over the entire treatment volume, 3D detectors have been used to verify
higher dimensionality measurements.

Factors leading to errors which can occur during the whole treatment procedure in
radiotherapy include patient miss positioning and a change in the patient’s anatomy due
to weight loss or organ movement [60]. Additionally, the treatment parameters such as
Linac setting and beam modifiers can be changed in between the pre-treatment
verification, which has been reported by Huang et al. [61]. There is a considerable demand
for real-time dose verification which enables real-time detection of major errors and can
assess the dosimetric impact quantitatively during radiation delivery [62].

Real-time verification can be carried out by using the transmission-type detector
positioned in the photon beam between the Linac head and the patient or by means of
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electronic portal imaging device (EPID) during the treatment. The commercial transmission detectors such as COMPASS system (IBA, Dosimetry), Dolphin system (IBA, Dosimetry) and David system (PTW-Freiburg, Germany) have been introduced for real-time dose verification.

As mentioned above, the complexity of advanced treatment techniques with the use of small field dose delivery requires the careful selection of the suitable dose measurement. The ideal detector for radiation dose measurement should have the following characteristics:

i) The detector should be tissue equivalent and not perturb the radiation beam.

ii) The detector should have the small sensitive volume to avoid the volume averaging effect.

iii) The detector should have a high dynamic range to manage with a large dose gradient that may be present per fraction of the treatment delivery.

iv) The detector should have energy, dose rate, directional independence and dose response linearity [63]–[66].

In this section, a brief introduction of both pre-treatment QA tools and real time treatment verification tools is presented.

2.2.2 Pre-treatment verification QA tools

2.2.2.1 Point Dosimetry

i) Ionisation chamber (IC)

The ionisation chamber is widely used as an absolute point dosimeter [67]–[69]. The detector can have a wide range of physical shapes depending on the specific requirements, such as parallel plate chamber, concentrate cylindrical or a wire within cylindrical,
thimble chamber, condenser chamber, extrapolation chamber and waterproof chamber [13].

The principle of this dosimeter is to measure an ion-pair generated by ionising radiation passing through the sensitive volume (gas cavity) of the detector. The ion-pair attach to either in the positive plate (anode) and a negative electrode (cathode) which connects to the battery to collect the signals. The diagram of an ionisation detector is shown in Figure 2.1.

![Diagram of ionisation detector](image)

**Figure 2.1.** The diagram of the ionisation chamber [70].

There are various operating regions of ionisation chambers depending on the voltage applied [71], [72]. For the normal ionisation region, the chamber operates when the amount of voltage is appropriated to collect all the ions produced in the active volume (100V-400V). However, it is insufficient to cause an increase in ion pairs due to gas amplification as the voltage increases [72]. The current read by the electrometer is
converted to the absorbed as the dose is proportional to the total ionising signal (charge) measured.

A change in air density in the chamber may occur as a result of changes in temperature and pressure and may affect the reading as it will increase as the temperature increases or as the pressure increases [73]. Therefore, to convert the measured ionisation signal to the absorbed dose, a correction for temperature and pressure is needed [73]. The standard protocols such as TG-51 or TRS-398 outline the procedures that deal with this process [74], [75].

The ionisation chamber provides accurate and precise measurements and is recommended for beam calibration with essential corrections. More advantages of the ionisation chamber are its long-term stability, relative ease of use and instant direct read out [76]. However, the ion chamber indicates some volume averaging due to the detectors finite size of sensitive air volume [77], [78]. This effect limits the use of ion chambers where high dose gradients exist such as in complex treatment plans in IMRT, VMAT and SRS/SBRT. The chamber can overestimate the dose in certain circumstances due to its large volume [79]. Moreover, the under-response of the detector is presented in small fields due to the volume averaging and needs to be corrected [69].

ii) Semiconductor detector
A silicon diode is the most common semiconductor detector and also sometimes referred to as a solid-state ionisation chamber. It is widely used in radiation dosimetry for radiation protection, radiation imaging and radiotherapy dosimetry [76], [80]. The detector is produced by taking pure silicon and doping it with phosphorus to produce n-type or with boron to produce p-type material [81]. The n-type material semiconductor has a significant number of free electrons compared to intrinsic silicon. The n-type is electrically neutral due to the free electron positive donor ions. The p-type semiconductor
has a significant number of holes compared to intrinsic silicon. The p-type is electrically neutral due to the holes negative acceptor ions [82].

Unlike the ionisation chamber, diodes can be operated without a bias applied [83]. A diode with an impurity of the opposite type is implanted into the surface region to create a p-n junction. The p-n junction alters the local electron and hole densities as well as creates the electrostatic potentials (built-in potential) in a diode close to the junction called the depletion region [84]. In this region, the free electrons will recombine with the hole, leading to diffusion of surplus charge carriers to the other material until thermal equilibrium is reached. At this point, the fermi level is equalised. The remaining ions create a space charge and an electric field stops further diffusion.

When ionising radiation passes through the diode, electron hole-pairs will be produced. The free electron on p-type and the hole on n-type will diffuse toward the p-n junction [81]. By applying the external voltage (V), the charges produced in the diode are swept or drift across the depletion region under the action of the electric field and can be read by the electrometer [80]. The diagram of the p-n junction is shown in Figure 2.2.

![Diagram of p-n junction](image)

Figure 2.2. The diagram of p-n junction [85].
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The silicon detector shows high sensitivity compared to the ion chamber with the same active volume. This allows the diode to be designed with an extremely small sensitive volume giving rise to high spatial resolution measurements being able to be made [86]. Therefore, diodes are well suited to measuring in a high dose gradient region such as beam penumbra region. It also provides for excellent charge carrier collection because of high mobility and a long mean free path of the majority and minority charge carriers [87]. However, on the negative side, the occasional recalibration for diodes is necessary as they suffer radiation response degradation due to radiation damage. The sensitivity of the diode decreases with accumulated dose with both p-type and n-type diodes [88]. The radiation response of the diode is also dependent upon temperature which should be taken into account. Grusel et al. found that the diode signal is increased by about 1-3% per 10 °C change in temperature [89].

Moreover, diodes yield over response to low energy photons ranging from 10 keV to 200 keV due to the high atomic number (Z=14) of the silicon relative to tissue (water). This leads to an increased photoelectric effect in this energy range compared to water where the Z is about 7 [68], [80].

iii) MOSFET Dosimeter

The principle of the radiation sensitivity of a Metal-Oxide-Semiconductor Field Effect Transistor (MOSFET) is based on the creation of ‘electron-hole (e-h)’ pairs similar to the semiconductor detector. However, the relevant e-h pairs, when a MOSFET is used as a dosimeter, are generated in the gate oxide. MOSFETs were first utilised for radiation dosimetry by Andrew Holmes Siedle et al. in 1978 [90], and they have been used in radiotherapy for decades [90]–[92].

When ionizing radiation passes through the silicon oxide layer (SiO₂), the positive charges drift under the electric field (if applied) and accumulate via traps at the Si-SiO₂
interface. The accumulation of these charges influences the current flow between the source and the drain of the MOSFET for a given applied bias [93]. In a constant source-drain current configuration the difference between the threshold voltage measured under the constant current, before and after radiation exposure, is a function of the absorbed dose in SiO$_2$ [94].

The MOSFET is small, and the gate oxide is very thin, providing the possibility of the shallow dose measurement. Many papers discuss the use of MOSFET for surface dose measurement as an *in-vivo* radiation dosimeter [95]–[99]. The MOSFET is not dose-rate independent but does offer real-time readout. As the radiation sensitive property of the MOSFET is of an integrating nature, the detector total dose history can be stored [93]. Similar to other semiconductor detectors, the MOSFET has temperature dependence and a limited lifespan that should be taking into account with regular recalibration necessary [76], [96].

### 2.2.2.2 Two-Dimensional (2D) dosimetry

i) Film dosimeters

The traditional radiographic film consists of a base of thin plastic with a radiation sensitive emulsion, silver bromide (AgBr) crystal, coated on one or both sides of the film [100]. The interaction of radiation to the AgBr forms the latent image in the film. When the film is developed, the small grains of metallic silver is reduced making the film opaque [101]. This opacity is defined in terms of the optical density (OD) which is a function of radiation dose exposed to the film [102]. The radiographic film has an excellent spatial resolution (<1mm) and can be cut into various shapes of the different area [102]. The development of an extended dose range (EDR) film allows radiographic film to be used over a wider dose range than traditional films [103]. However, the
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requirement of chemical processing to develop or fix the image and the requirement of darkroom facilities is leading to the decreasing usage of the radiographic film [104], [105].

The self-developed film has been introduced by David Lewis and is referred to as radiochromic film [106]. Unlike radiographic film, radiochromic film is self-processing, eliminating the wet chemical processing step [107]. This film consists of a single or double layer of radiation-sensitive organic microcrystal monomers, on a thin polyester base with a transparent coating. This radiation sensitive monomer is polymerised by a topochemical (solid state) process to form the film opacity without any latent image [108].

The radiochromic film material is close to tissue equivalent when compared to radiographic films, and it can be measured in various ranges of high radiation dose (1-10 Gy) [107]. Thus, radiochromic film has been shown to be used as a QA tools in clinical dosimetry [109]–[112].

The film is a high spatial resolution, large sensitive area QA tool. It provides a 2D intensity map from a single exposure which can be converted to the 2D dose mapping by using the calibration curve [110]. However, the film dosimetry is not generally used for real-time measurement because the film developing takes around 24-48 hours to stabilise the film response [106]. Additionally, the film response depends on many factors including; the film plane orientation, densitometer/digitiser artefacts, temperature, humidity and storage conditions [106], [113]–[116].

ii) 2D Ionisation chamber array

The PTW 2D ARRAY developed by PTW-Freiburg Germany is an example of 2D ionisation array. The ARRAYS are available in two versions that provide different numbers and sizes of ionisation chambers. Version 1 consist of 256 ion chamber arranged
in a 16 x 16 matrix while Version 2 consists of 729 ion chamber in a 27 x 27 cm$^2$ matrix. Both versions cover the area of 27 x 27 cm$^2$. The basic characteristic were studied by Poppe et al. The detector demonstrates both short-term and long-term reproducibility within 0.2% and 1%, respectively [117]. Poppe et al. examined the use of PTW 2D ARRAY for IMRT verification. The plan was verified by placing the detector perpendicular to the Linac gantry which was set to a 0-degree delivery [117], [118].

Another example of 2D ionisation chamber array is MatriXX (IBA dosimetry, Scanditronix Wellhofer GmbH, Germany). This detector consists of 1024 ion chamber in an active area of 24 x 24 cm$^2$. The detector diameter is 4.5 mm with 7.62 mm to each adjacent detector distance. Each detector has a sensitive volume of 0.08 cc. Yan et al. reported that the detector has good dose linearity and provides stable long-term reproducibility with respect to low dose rate dependence [119]. Wolfsberger et al. characterised the angular dependency of MatriXX and found the discrepancies in detector response (up to 11%) as a function of gantry angle (Anterior-Posterior VS Posterior-Anterior fields). This effect is due to air-high-Z material interfaces [120]. Han et al. examined this detector for IMRT QA and found that the detector was successfully used for IMRT QA, but the issues of the detector volume averaging effect were reported [121].

iii) 2D silicon diode array
MapCHECK (Sun Nuclear, Melbourne, USA) consisted of 445 N-type silicon diodes and covered the active area of 22 x 22 cm$^2$. The active volume of the detector is 0.8 x 0.8 mm. MapCHECK separates into two areas; the outer band array has 2.0 cm horizontal and vertical spacing. The inner band array with the area of 10 x 10 cm$^2$ has 1.0 cm horizontal and vertical spacing [122]. The detector response presents excellent dose linearity. However, the N-type diodes have a temperature coefficient of 0.54%/°C. Thus, the
detector storage at or close to the treatment room temperature is recommended [59]. As a silicon diode detector, MapCHECK degrades as they accumulate dose. So, it is necessary to check and update the array calibration annually regularly or as needed [123]. Several studies examined MapCHECK for IMRT and VMAT pre-treatment verification [33], [124]–[126]. Liu et al. found that the detector presented inconsistency with a gamma comparison (3%/3mm) and that this deviation increased when the IMRT plan is more complex [119]. This might be due to the few sampling points within the field, especially for small fields and the detector resolution influenced by the non-uniform detector distribution spacing of between 7 mm and 14 mm.

2.2.2.2 Three-Dimensional (3D) dosimetry

Advanced radiotherapy typically delivers a dynamic radiation beam with the dose rate, MLC geometry, and gantry angle continuously varying. The 2D dosimeter mentioned above is often used to validate a treatment in a single planar dose distribution. To provide the information in full 3D dose distribution throughout the entire treatment volume, 3D dosimetry is required.

i) ArcCHECK

ArcCHECK (Sun Nuclear, Melbourne, USA) is an example of a 3D dosimeter. The detector consists of 1386 N-type diodes in a cylindrical phantom. This system was developed for the rotational therapy QA. The diode air embedded with 10 mm spacing in a spiral pattern to increase the spatial sampling rate. The cylindrical phantom is 21 cm in diameter and length. The active volume of the detector is 0.8 x 0.8 mm. The detector spacing is 1cm x 1cm, and smaller when projected at the different source to surface distance (SSD). The basic characteristics of ArcCHECK were studied by Li et al [127].
The detector showed the good response for short-term and long-term reproducibility, dose rate dependence, dose linearity, dose per pulse dependence. Nevertheless, ArcCHECK presented directional dependence. At the gantry 105°, the directional dependence was varied about 9.1%[127]. ArcCHECK has been reported to be used for IMRT and VMAT QA [128], [129].

ii) Gel dosimetry
Gel dosimetry was first introduced in 1950 by Stein et al [130]. At that time, they studied the colour changes in the gel containing dyes produced by radiation in aqueous solution. The chemical change related to the absorption of radiation dose.

More recently, the dosimeter polymer gel is made from various agents which are sensitive to radiation such as gelatine, agarose, Sephadex and polyvinyl alcohol [131], [132]. Gels are nearly tissue equivalent. Thus, no energy correction is required for both photon and electron beams. It can be modelled to any desired shape and have high spatial resolution. The dose distribution was recorded in three-dimensions and can be read out by several methods such as magnetic resonance imaging (MRI), computed tomography (CT) or ultrasound technique [133]–[135].

Gel dosimetry became one choice of radiation dosimetry. It was previously used for IMRT, VMAT, SRS and SBRT dose measurement [136]–[138]. The disadvantage of gel dosimetry is that it’s processing is complicated and expensive. Reproducing the gels with similar radiation sensitivity is difficult. The use of this dosimetry for routine radiation measurement is limited [139].
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2.2.3 Real-time treatment verification QA tools

2.2.3.1 Amorphous Silicon Electronic Portal Imaging Device (EPID)

The EPID system consists of a flat panel array mounted on a retractable arm opposite the Linac beam. It was primarily designed for daily imaging to verify the geometric accuracy of radiation filed on the patient [140].

The new generation of EPID system is the amorphous silicon-based system. The EPID system has been extended for the purpose of dosimetric verification related to image information. There are several ways to use EPID as a dosimetry QA tool. One method is to use the device to verify the MLC position by capturing a series of snapshot images during a dynamic MLC prescription. This method can be operated both online and off-line [141], [142]. Another method is to reconstruct the radiation dose to the patient by using the exit images acquired by the system during the treatment [143]. This method requires some corrections from the scattering of the build-up material [144]. The recently used method is to convert EPID image to an incident fluence distribution and use as the input to compute the dose to the patient by the back projection method [145].

The characteristics of EPIDs have been reported and mention that the response of an EPID is linear with integrated dose and not dependent on dose rate [146], [147]. It is non-tissue equivalent material and over sensitive to low energy photon beam. Image contrast with high photon energy (MV) is lower than with kV beam used in diagnostic radiography [148]. Thus, for the purpose of patient setup, the patient image is performed with the kV source incorporated with the Linac gantry, e.g. Varian OBI system [149].

EPID has been used for IMRT plan verification with advantages over the traditional use of films such as ease of use, real-time imaging display and instant comparison with TPS [150], [151]. However, the procedure to convert the image to the dose is complex and it is still a challenge for VMAT verification because of the dose rate changes continuously.
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[152]–[154]. This leads to a phase shift between MLC leaf openings and the portal arc [155].

2.2.3.2 DAVID System

DAVID system (PTW-Freiburg, Germany) is a transmission-type detector which has been developed for only a standard Siemens Linac (Siemens Healthcare Pty Ltd, Germany). The detector consists of a flat, vented translucent multi-wire ionisation chamber [156]. The number of IC is equal to the number of MLC leaf pair of the Linac machine. The device can be placed at the wedges or block tray slot or can be permanently installed at the Linac head, which is only used for IMRT. The distance from one detection wire to the next one is 4.31 mm [156]. Each wire of DAVID system corresponds to the MLC leaf-pair to verify the movement of the leaf during IMRT treatment. The measured signal of all wire is a dose area product of the transmitted photon beam and the total radiation dose to the patient [156]. The influence of the detector, when used in transmission measurements, has been reported by Poppe et al [157]. They reported that there is radiation absorption by the chamber due to the tray factor and there is surface dose perturbation especially at small field sizes. Karagoz et al [158] investigated the David system for IMRT QA. They found that the deviations in leaf position of static IMRT plan were within 3% from the first week of treatment with a strong correlation with EPID. The deviation of leaf position is dependent on the intensity level.

2.2.3.3 Delta$^4$ AT Discover

Delta$^4$ AT (Scandidos AB, Uppsala, Sweden) consists of 4,040 p-type diode detectors. The detectors have an active area of 1 mm in diameter [159]. The active detector covers an area of $25 \times 20 \text{ cm}^2$ when the beam is projected to a distance of 100 cm source to axis...
distance. The device has various detector spacing, depending on the axis direction, e.g. along the MLC leaf in X direction the detector spacing is 1.6 mm apart while in Y direction space is 3.2 mm intervals. The overall detector thickness is 4.0 cm when placed at 63.6 cm source to device distance [160].

This device is an additional component to a Delta$^4$PT (Scandidos AB, Uppsala, Sweden) [161] which provides the pre-treatment QA data. Both systems share software to be used during patient treatment. The patient pre-treatment QA data is used to calculate the virtual dose based on the Delta$^4$ AT measured data by the QA software [160]. The interval time for transferring the Delta$^4$ AT measured data to the software is 25 ms. The beam perturbation when the device was in the beam path was evaluated. The surface dose increase varies from 1% - 9% depending on photon energy and radiation field size [160]. The largest change in the percentage depth dose (PDD) measurement was observed at a depth of 10 cm with 0.5% decrease in dose [160].

2.2.3.4 COMPASS

COMPASS (IBA Dosimetry, Germany) is a pixel segment 2D ionisation chamber array. The array consists of 1600 air vented plane-parallel ion chambers. The detector’s active volume is 0.02 cm$^3$ in the active area of 40 x 40 cm$^2$. The detector spacing is 6.5 mm. COMPASS can be attached to the Linac head (Varian Linac 2100iX). The source to detector distance is 65 cm. Sankar et al. studied the influence of the COMPASS system when used in the transmission mode for 6 MV photon beams. They found an increase in the surface dose for shorter SSD and the large irradiated field. Beyond the depth of maximum dose, the perturbation properties were in good agreement with the open field [162].
2.2.3.5 Dolphin dosimetry system

Dolphin dosimetry system (IBA Dosimetry, Germany) consists of 1513 air-vented plane-parallel ionisation chambers. The active area is 24 x 24 cm$^2$. Each chamber has a diameter of 3.2 mm and 2 mm height. The pixel pitch is 5 mm in the inner detector area and approximately 8 mm in the outer area with the active volume of 0.016 cm$^3$ [163]. The device corresponds to the COMPASS dosimetry software (IBA Dosimetry, Germany) [163]. The surface dose increasing when the Dolphin detector was in the beam was found the maximum about 11% at the SSD of 80 cm [164]. The influence on PDD measured by dolphin detector was 1% beyond the depth of maximum [164]. Thoelking et al. evaluated the clinical performance of this system and found that a good agreement for dose reconstruction based on dolphin detector read-out compared to TPS was observed for IMRT plans with a 3% error of MLC position [165].

2.2.3.6 The Integral Quality Monitoring system (IQM)

IQM system consists of an area integrating energy fluence monitoring sensor (AIMS) and a calculation module (IQM_CALC). This detector is designed to be mounted with the final beam shaping device, the MLC and the patient. The detector is made from Aluminum and has a physical size of 22 x 22 cm$^2$. The detector sensitive volume is 530 cm$^3$ and can monitor the maximum radiation field size of 34 x 34 cm$^2$ at the isocentre. The principle of this system is that the dose measured by AIMS will be compared with the predicted dose calculated by IQM_CALC. The signal from the ion chamber (AIMS) provides spatially dependent dose-area-product for each beam segment. The calculation dose (from IQM_CALC) is based on the integration method associated with the information from the TPS. The signals from AIMS and IQM_CALC are compared in real-time. Islan et al. studied the IQM system and found that the chamber attenuates the
beam intensity by 7% and 5% for 6 and 18 MV beams, respectively, without changing depth dose, surface dose and dose profile characteristic for field size 10 x 10cm² [166]. Although some transmission QA dosimetry devices are available, the spatial resolution and beam perturbation for some of them make their use in clinical practice for the advanced treatment technique such as SRS and SBRT for real time treatment verification, questionable.

2.3 Center for Medical Radiation Physics (CMRP) semiconductor dosimetry

Many devices have been developed for the purpose of advanced treatment technique QA such as SRS and SBRT as mentioned previously. Due to the complexity of the SRS and SBRT delivery with a very small field, there is a considerable demand for real-time dose delivery verification with high-resolution detectors during patient treatment. Center for Medical Radiation Physics (CMRP), University of Wollongong developed the 2D diode array Magic Plate 121 (MP121) as an online radiation detector. The detector is based on small single epi-diodes embedded in a KAPTON carrier with pitch 1 cm and overall thickness of 0.45 mm only. MP121 has been used as a transmission detector for real-time dose monitoring in which the detector was mounted on the Linac head. The detector presented minimal beam perturbation leading to an increase in the surface dose of less than 0.5% [167]. However, the spatial resolution of the MP121 detector is precluding its effectiveness for small field real-time QA.

A new detector, the Magic Plate 512 (MP512), has been developed with a better spatial resolution. MP512 is a silicon monolithic with a low resistivity p-type substrate. The detector consists of 512 pixels with a 2 mm detector pitch. The silicon detector arrays with thickness 0.45 mm are wire bonded to a thin 0.5 mm tissue equivalent printed circuit
board (PCB) and covered with a layer of resin to protect it from any accidental damage. The use of MP512 in in-phantom dosimetry has been previously studied [168], [169]. Additional studies for in-phantom dosimetry of MP512 will be presented in Chapter 4. The dose mode measurements such as output factor, percentage depth dose and beam profile with different air gap size above the detector will be evaluated with various field sizes (include small field size) for both photon beams and electron beams. The following chapters (chapter 5, chapter 6 and chapter 7) present the use of a thin transmission monolithic MP512 detector as a transmission QA tool.

2.4 Small field dosimetry

Advanced radiation treatment techniques such as SRS and SBRT traditionally use small radiation fields in sub centimetre range to treat tumours and spare normal tissue. IAEA/AAPM define a small field as a field where the dimension is smaller than the lateral range of the charged particles. Generally, the small field is defined as a filed size of less than 4 x 4 cm$^2$ [170]. There are three existence conditions represented in small filed dosimetry and are discussed below.

2.4.1 A loss of lateral charge particle equilibrium (LCPE).

LCPE is a part of charge particle equilibrium (CPE) associated with a range of secondary electrons [171]. CPE effect can be explained by the Bragg Gray cavity theory. In this theory ionisation chamber (IC) is used as a reference detector and assumes that it does not disturb the particle fluence when inserted into a medium [172]. The CPE exists for volume $v$ when the energy absorbed per unit mass equals the energy imparted per unit mass [171]. This means the ionisation produced within the gas-filled cavity inside the
medium is proportional to the energy absorbed. Therefore, absorbed dose (D) equals total kerma of primary radiation photon (ratio =1).

\[
\frac{D}{K_{coll}} = 1
\]

In a photon beam the kerma is defined as the initial kinetic energy of all charged particles, mainly electrons and positrons, liberated by photon interactions per unit mass in a medium. Since part of this kinetic energy may be converted back to energetic photons mainly through bremsstrahlung and annihilation in flight processes it is useful to analyses that part to the kerma which remains as kinetic energy of charged particles, namely the collision kerma, \(k_{coll}\) [173], [174].

When the field size decreases, the variation of electron fluence depends on the radiation field size [175]. The maximum of secondary electrons range is larger than the closest field edge. This causes an uncertainty of the dose in the small field compared to the calibration filed where Bragg Gray cavity is broken down, and lateral charge particle disequilibrium occurs [176].

Another concept of CPE was explained by Das et al. They stated that the beam is broad and parallel. Thus, there exists an artificial source everywhere to compensate for any photon loss during interaction [177]. LCPE is used for nonstandard beam and is used for an infinite flat, broad beam and homogeneous phantom where the photon fluence is laterally uniform at all depths and in all directions for all energies [178]. The loss of scattered photons for each primary photon is replaced by scattered photons which is generated by other primary photons in the beam direction. If the medium is changed, the number of photons and scattered photons generated laterally depend on the properties of the medium such as medium density. So LCPE can no longer be existed [179]. In a small field, a lack of LCPE is considered when radiation direction passes through an
inhomogeneous medium including the detector. Hence the correction for beam perturbation in an inhomogeneous medium is needed for the detector [177], [179].

### 2.4.2 Partial source occlusion

The photon fluence from the Linac machine consists of primary photons and secondary photons. The former generates directly from the target focal spot while the latter is produced from the structure of Linac head as the scattered photons [180]. Generally, the target is spread over the area, and the source profile size is determined by full width at half maximum (FWHM) [13]. The beam size is collimated by the collimators. By decreasing the collimator setting, the field size decreases. The primary photon and scattering photon is blocked by the collimator leading to a reduction in the absorbed dose. Moreover, the FWHM of the source profile is reduced when decreasing the collimator setting [181]. Thus, the small field size output is lower than the field size at which the whole source can be seen from the detector point of view as shown in Figure 2.3.

[177], [178], [182].
Figure 2.3. (a) Broad beam and (b) small beam from the detector point of view.

### 2.4.3 The detector volume averaging effect

The radiation dose measured by a dosimeter is associated with the averaging charge over the entire sensitive volume of the detector, where the dose is proportional to a number of charged particles [183]. To measure the accurate dose, the detector should be irradiated uniformly. In a small field, a steep dose gradient can be affected by the dosimeter which has a large sensitive volume compared to the radiation field size. A flat field profile includes a portion of penumbra might be measured over the sensitive volume [178]. The measured dose would not be accurate and may result in reduced signal [78].

The dose calculation for treatment planning requires beam data to drive the calculation model such as output factors [184]. When measuring the output factor for SRS/SBRT field (small field) with a large dosimeter, an underestimation of the output factor will be present in the measurement. The incorrect beam parameter for dose calculation would cause the miscalculation which significantly affects the treatment [185], [186]. So the
Chapter 2: Literature Review

high spatial resolution dosimeter with a small active volume and good reliability which minimise the volume averaging effect is important for small field dosimetry [178].

2.5 Skin and surface dose

The International Commission on Radiation Units and Measurements (ICRU) and the International Commission on Radiological Protection (ICRP) recommend the skin depth for practical dose assessment should be at a depth of 0.07 mm below the surface. This depth is the deepest layer of the epidermis and lies above the basement of the membrane and is also known as the basal layer [187]–[191].

Many studies have shown that in radiotherapy, the skin dose is affected by changing in parameters such as patient geometry, beam energy, SSD, field size, the use of wedges, blocks, block trays, thermoplastic mask and bolus [192], [193].

Yadav et al., estimated the skin dose for various beam modifiers and SSD for 6 MV photon beams. They found that skin doses were increased as the SSD decreased and were dominant for larger field sizes. The measured skin dose due to a motorised 60° wedge for the 10 x 10 cm² field was 9.9%, 9.5%, and 9.5% at 80 cm, 100 cm and 120 cm SSDs. The measured skin dose due to acrylic block tray, of thickness 1.0 cm for a 10 x 10 cm² field was 27.0%, 17.2% and 16.1% at 80, 100 and 120 cm SSD, respectively [194].

Doracy et al. have measured 79% of the maximum dose when treating through the material versus 22% of the maximum dose when no beam modifier or immobilisation devices are used [193].

Kim et al. confirmed that the skin dose increased as the field size increased. They showed that for all field sizes the skin dose increased with the use of block tray; 7% to 59% for 8 MV and 5% to 62% for 18 MV beam [195].
Several radiation detectors have been used to investigate the surface dose and build up regions such as a parallel-plate ion chamber, TLD, film and MOSFET. Each detector has its own advantages and disadvantages that may make it more favourable than the others in the various applications.

TLD can be used for in vivo dosimetry because of its low cost and tissue equivalent. The extrapolation method is used for TLD skin dose measurement. However, it cannot read in real-time due to the complexities of TLD processing [196]. Film dosimetry has been used for surface dose measurement due to the effective film depth being near the basal layer and its high spatial resolution. So it is not necessary for any correction unlike the dose measured by the ion chamber [197]. However, film dosimetry is time-consuming and the signal can be affected by many parameters which have been discussed previously in 2.2.2.2 [198]. Rosenfeld et al. promoted MOSFET for surface dose measurement. The MOSFET data showed excellent agreement with the reference chamber (Attix ion chamber) in the build-up region. It is small in size and has a simple reading circuit which can be read out online [92], [93].

2.6 Gamma evaluation

The gamma evaluation method as presented by Low et al. [51] is designed to compare the measured dose distribution and the calculated dose distribution. Figure 2.4 shows a diagram of the gamma evaluation method. This figure is presented for a single measurement point. Generally, all measurement points are repeated for the comparison in the clinical practice. The measured dose ($r_m$) is used as reference information, and the calculated dose ($r_c$) is queried for comparison. X and Y axes are the spatial locations of $r_c$. 

28
and the Γ axis represents the difference between the measured dose \([D_m(r_m)]\) and calculated dose \([D_c(r_c)]\).

![Diagram of the gamma evaluation method](image)

**Figure 2.4.** A diagram of the gamma evaluation method [51].

The acceptance criteria are defined by \(\Delta D_M\) for the dose difference and \(\Delta d_M\) for the distance to agreement. The acceptance criteria is an ellipsoid defined by

\[
1 = \frac{\Delta r^2}{\Delta d_M^2} + \frac{\Delta D^2}{\Delta D_M^2}
\]

where \(\Delta r\) is the distance between the reference and compared point

\[
\Delta r = |r_m - r_c|
\]

and \(\Delta D\) is the dose difference between dose distribution at \(r_m(D_m)\) and \(r_c(D_c)\), Thus

\[
\Delta D = D_c(r_c) - D_m(r_m)
\]
A quantitative measure of the accuracy of the correspondence is determined by the point with the smallest deviation from the reference point lying within the ellipsoid of acceptance, i.e. one point for which:

\[ \Gamma_m(r_c, D_c) = \frac{\Delta r^2}{\Delta d_M^2} + \frac{\Delta D^2}{\Delta D_M^2} \leq 1 \]

The pass and fail criterion therefore become

\[ \Gamma(r_m) \leq 1, \text{ correspondence is within the specified acceptance criteria.} \]

\[ \Gamma(r_m) > 1, \text{ correspondence is not within specified acceptance criteria.} \]

An implicit assumption is made that once the passing criteria are selected, the dose difference and DTA analyses have equivalent significance when determining calculation quality.
In the previous chapter, the QA tools for advanced treatment techniques for both pre-treatment and real-time treatment were reviewed. The development of a new silicon diode detector array ‘MP512’ was introduced as a QA system. This chapter describes the Magic Plate 512 detector system that was used in this thesis for in-phantom dosimetry and transmission dosimetry. The electronic readout system will be described as well as other detectors that have often been used to compare with the MP512 dose response.

3.1 Linear accelerator and field arrangement

For the work described in this thesis, the photon beams and electron beams were generated by a linear accelerator Varian model 2100IX (Varian Medical Systems, Palo Alto, CA). All experiments were performed at the Illawarra Cancer Care Centre, Wollongong Hospital, Wollongong.

The Linac provides dose rates of between 100 MU/min and 600 MU/min with 100 MU/min increments. The maximum MU that can be delivered is 9999 MU for any one treatment. The field size in this study is defined at the 100 cm SSD. The field sizes are collimated by the Linac jaw and MLCs and varied from 1 x 1 cm² to 20 x 20 cm².
Chapter 3: Methodology

3.2 Magic Plate 512

The Magic Plate 512 (MP512) was designed and developed at the CMRP. It is a 2D array of isolated p-i-n silicon diodes embedded together in an ion-implanted silicon monolithic diode detector, manufactured on a bulk $p$-type substrate. The silicon substrate is 0.45 mm thick. The MP512 array consists of 512 pixels with a detector array-element size of 0.5 x 0.5 mm$^2$ and pitch 2 mm with an overall dimension of 52 x 52 mm$^2$ as shown in Figure 3.1. The MP512 monolithic detector is mounted and wire bonded to a printed circuit board (PCB) 0.5 cm thick and covered by a thin layer of resin to preserve the silicon detector from moisture and chemical contamination and to protect the wire bonds [168].

![Image of Magic Plate 512 bounded with the PCB](image)

Figure 3.1. Magic Plate 512 bounded with the PCB

The PCB provides the fan-out for connecting the sensor to the readout electronic system. The MP512 detectors operate in passive mode and have no bias voltage applied to the diodes. In this thesis, a thin monolithic silicon detector MP512 was designed to operate
Chapter 3: Methodology

in both dose mode measurement (in-phantom) and transmission mode measurement arrangements.

3.2.1 Magic Plate 512 for in-phantom measurement

To use the detector for in-phantom dosimetry, the MP512 was placed between two polymethyl methacrylate (PMMA) slabs to protect the detector from mechanical damage and shielding the sensor from ambient light [199]. Figure 3.2 presents the MP512 and the schematic diagram of MP512 embedded for in-phantom measurement. Some characterisation of MP512 and the use of detector as an in-phantom QA tool has been previously reported [168]. In this thesis work, the characterisation is extended significantly as it focused on operating the device in transmission mode. In chapter 4, more details of the detector characteristics will be studied such as the effect of the air gap on detector response and the optimisation of the suitable air gap size upstream of the silicon detector.
Figure 3.2. (a) MP512 detector wire bonded to the PCB sandwiched with the two PMMA slabs, (b) MP512 when used in dose mode in phantom dosimetry and (c) a schematic diagram of the MP512 packaged between two PMMA slabs.
3.2.2 Magic Plate 512 for transmission mode measurement

To use MP512 as a transmission detector, the detector is embedded with different packaging. The detector is sandwiched between 3 mm thick PMMA sheets with an opening area of 9.5 x 9.5 cm² at the centre of the board in both front and back of the detector to generate a 0.45 mm thick transmission detector.

When operating the MP512T in transmission measurement mode, the detector is covered with a black plastic sheet (80 µm), to reduce light leakage to the detector. The MP512T is placed on a movable stand holder which has the capability of moving in a vertical direction. Moving the detector along the beam axis between the patient surface and the Linac head enables the detector to change the effective spatial resolution that the radiation field is sampled. The concept of a movable transmission high-resolution detector and more details will be explained in chapter 5. Figure 3.3 shows the MP512T when used in transmission dosimetry and the detector packaging schematic.
Figure 3.3. (a) MP512T detector wire bonded to the PCB and sandwiched with the two PMMA slabs with the opening in place of the detector, (b) MP512T placed on the movable stand holder when used in transmission mode measurement (c) A simplified schematic of MP512T packaging
3.3 The data acquisition system (DAQ)

The MP512 data acquisition system was custom designed at the CMRP. The system is based on a multichannel electrometer chip named AFE0064 from Texas Instruments. The AFE0064 chip is a current integrator which consists of 64-channels. For each channel, the analogue differential output which is proportional to the charge accumulated in a capacitor during a particular configuration is provided. The chip is set electronically through a serial protocol interface on the lowest gain available to span the full scale up to 9.6 pC, with a resolution of 16 bit and a non-linearity of less than 0.1% [200].

The DAQ system uses eight AFE chips to readout all the 512 channels, and the MP512 signal is synchronized with the Linac pulse. It is read out by four analogue-to-digital (ADC) converters. When the beam is on, all acquisitions were synchronised to the Linac trigger signal by a field programmable gate array (FPGA). The FPGA DAQ and the Linac machine is connected to each other by means of a coaxial cable. The FPGA itself connects via a USB 2.0 link to the host computer. The signal from each pixel is acquired in synchronisation with each Linac pulse (pulse-by-pulse) or by using an internal trigger at a frequency of up to 5 MHz. More details about DAQ system can be found elsewhere [201].

The CMRP furthermore designed the graphical user software for external beam radiotherapy which provides real-time visualisation and flat field correction, however not part of this thesis work. While the beam is on, the interface is able to present in both instantaneous detector response and integral detector response Figure 3.4 shows the software interface for reference and all the commands for the device controller.
Figure 3.4: The AFE-MP512 software interface version 1.37.

Due to the different in operational characteristics of photon beam and electron beam production, the relevant parameters used in the software interface setting was set as shown in Table 3.1.

Table 3.1. Relevant parameters for controlling the AFE-MP512 software interface for photon beams and electron beams.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Photon</th>
<th>Electron</th>
</tr>
</thead>
<tbody>
<tr>
<td>Range</td>
<td>7</td>
<td>7</td>
</tr>
<tr>
<td>Buffer size (kB)</td>
<td>4096</td>
<td>4096</td>
</tr>
<tr>
<td>Acquisition Length (s)</td>
<td>15</td>
<td>20</td>
</tr>
<tr>
<td>Integration time (µs)</td>
<td>78</td>
<td>20</td>
</tr>
<tr>
<td>Frequency (kHz)</td>
<td>0.36</td>
<td>0.18</td>
</tr>
<tr>
<td>Integration time of offset (µs)</td>
<td>65</td>
<td>65</td>
</tr>
</tbody>
</table>
3.4 Ionisation chamber

In the thesis work described here, two types of ion chambers were used including the Farmer IC and Markus IC and both are often used as a comparative dosimetry in clinical settings. The following details are briefly discussed regarding these two detectors.

3.4.1 Farmer ionisation chamber

A Farmer chamber (Model 2517A) was used in this thesis for transmission factor (TF) measurements. A Farmer ionisation chamber is a thimble-type (or cylindrical) chamber which is a fundamental tool for medical dosimetry. The chamber has a cylindrical cavity in which an electric field is applied between a conductor coated on the inner surface wall and collector electrode that lies along the centre of the cavity. The leakage current from the high voltage electrode is prevented by the guard electrode of the thimble chamber. The ion-collecting volume is also defined by this guard [202]. Figure 3.5 schematically presents the thimble ionisation chamber. The thimble wall is often made of pure graphite and the central electrode of pure aluminium with the typical air volume between 0.05-1.00 cm³. The wall material is designed to be thick enough to establish CPE or TCPE or thin enough not to perturb the fluence of charged particles. The wall thickness is about 0.1 g/cm². The chamber is at the ground and the guard is kept at the same potential as the collector. The chamber radius is typically 2-7 mm and length 4-25 mm [203].
3.4.2 Markus ionisation chamber

Markus ionisation chamber (Model N23343) was used in this thesis for surface dose measurements and percentage depth dose measurements. The Markus is a plane parallel ionisation chamber. It has been recommended for the surface dose measurement for photon beams and electron beam [173], [205]. This chamber has a flat cavity which can minimise in-scattering perturbation effect. The detector consists of a guard ring surrounded by the collecting electrode. The purpose of guard ring is to prevent undue curvature of the electric field over the collector [173]. An electrode spacing of the detector is about 1-2 mm, and the sensitive volume is 0.35 cm$^3$. The collecting surface on an insulator is coated with graphite. Figure 3.6 shows a schematic of a parallel plate ionisation chamber.

Figure 3.5. The schematic representation of thimble ionisation chamber [204]
Markus ionisation chamber is a good alternative as the extrapolation chambers. However, this chamber is known for their over-response due to the secondary electron generated from their small guard ring and their internal dimensions [206]. Generally, this effect occurs only at build-up region [207].

All data measured by Markus IC in this thesis is corrected for detector over-response by using Velkley correction as modified by Rawlinson [208], [209]. The chamber dimension used for correction calculation was obtained from Chen et al. as shown in the equation (3.1), (3.2) and (3.3) [210].

\[
P(d, E) = P'(d, E, G) - \xi (d, E, G) \tag{3.1}
\]

\[
\xi (d, E, G) = \xi (0, E, G) x e^{-4.0d/d_{max}} \tag{3.2}
\]

\[
\xi (0, E, G) = c(E) x \left(\frac{s}{w}\right) x \rho^{0.8} \tag{3.3}
\]
Chapter 3: Methodology

Where \( P \) is the true PDD and \( P' \) is the measured PDD. The \( \xi \) is an over response correction factor, \( E \) is the photon energy, \( d \) is the depth in the phantom, \( \rho \) is mass density of the chamber wall. The \( s/w \) is the ratio of electrode separation to the diameter of wall. For 6 MV and 10 MV photon beam \( d_{max} = 1.5 \) cm and 2.1 cm, respectively. As mentioned in Rawlinson study, the \( c(E) \) for 6 MV is 27% and for 10 MV is 18.41%. All parameters for the original Markus IC is shown in Table 3.2.

Table 3.2. The original Markus Ionisation chamber specification [211].

<table>
<thead>
<tr>
<th>Material</th>
<th>Wall (w)</th>
<th>Collecting electrode</th>
<th>Window (CH(_2))</th>
</tr>
</thead>
<tbody>
<tr>
<td>Density (g/cm(^3))</td>
<td>1.189</td>
<td>1.189</td>
<td>0.93</td>
</tr>
<tr>
<td>Diameter (mm)</td>
<td>6</td>
<td>5.3</td>
<td>-</td>
</tr>
<tr>
<td>Separation (mm)</td>
<td>-</td>
<td>2</td>
<td>-</td>
</tr>
<tr>
<td>Thickness (mg/cm(^2))</td>
<td>-</td>
<td>-</td>
<td>2.5</td>
</tr>
</tbody>
</table>

3.5 Radiochromic Film dosimetry

Gafchromic EBT3 film (ASHLAND, Wayne, NJ) was often used for comparison with the dose measured by MP512 such as output factor, wedge beam profile, PDD and IMRT plan dose measurement. The characteristic and the usefulness of the EBT3 was briefly described in Chapter 2. This section will explain the film workflow that performed in this thesis.

3.5.1 Calibration phase

To characterise the radiation dose sensitivity curve of a batch of EBT3 film, dose calibration measurements were performed.

The film sheet was cut into multiple pieces with the size of 3 x 3 cm\(^2\). In this thesis, 12 pieces were used for clinical practice. Each film was Pre-scanned before exposure by a
Microtex ScanMaker i800 scanner. In order to warm up the scanner for better film analysis consistency, each film was scanned six times, and the last three scans were kept for analysis [113]. The film was positioned at the centre of the scanner and was scanned in 48-bit RGB colour mode with 70 dpi scanning resolution. All films were placed in the same orientation to minimise any uncertainties [113].

For film dose calibration measurement, the film was positioned at $d_{\text{max}}$ (1.5 cm for 6MV) in a solid water phantom and aligned at the beam centre. The full backscatter condition was set with 10 cm thick of solid water. A known dose ranging from 0 – 40 Gy was delivered to each film. All calibration setup was repeated for 10 MV photon beams.

All films were kept for at least 48 hrs for full development at the unexposed UV area to avoid any possible darkening of the film [212]. The post-scanning were performed to produce the image with the similar scanner setting as the pre-scan.

To obtain a calibration data from the scanned images, a set of multiple dose optical densities (OD) was investigated by two software tools including; the Image J version 1.48v (National Institute of Health) and MATLAB (The Math Works Inc., Natick, MA). The scanned image consists of 3 components; red, green and blue channels, only the red channel is used for the dose conversion. Find the average intensity of the interested area (at the centre of the film each film) and convert the intensity to the optical density (OD) by equation (3.4) [109].

$$ OD = \log\left(\frac{I}{I_0}\right) $$ (3.4)

Where $I$ is the intensity (post-scanning value), and $I_0$ is background intensity (pre-scanning value)
Chapter 3: Methodology

Then plot the OD vs the known dose to generate the calibration curve [113], [213]. Devic et al. and Battum et al. recommended to fit a non-linear calibration curve with a second or higher order polynomial to generate the calibration equation[214], [215]. Figure 3.7 shows the example of the calibration curve used in this thesis.

![Calibration Curve](image)

Figure 3.7. The calibration curve for 6MV photon beam

### 3.5.2 Measurement phase

Similar to calibration phase, film sheets were cut into the desired size for clinical measurement. For instance; the film was cut into a size of 7 x 7 cm² for output factor measurement and 10 x 10 cm² for IMRT plan delivery. Pre-scanning was performed to obtain a background intensity.

A blank EBT film was then irradiated to the radiation field of interest. A waiting time is similar to the calibration phase, and then post-scan the film. The images were converted into the dose distribution by converting OD value into the dose using the calibration
equation. The average uncertainty calculated across all measurements by film dosimetry in this thesis is approximately 1.98%.

3.6 MOSkin dosimetry

MOSkin was often used as a comparison dosimeter in the part of surface dose measurement. The MOSkin is a Metal-Oxide-semiconductor Field Effect Transistor (MOSFET) designed and built at CMRP.

It was designed using a p-MOSFET sensor with a thick gate oxide and sealed within a Kapton pigtail strip using drop-in technology [216]. A film layer protects the detector from any moist and dust and build layer up providing a water equivalent depth of 0.07mm [216]. More advantages of MOSkin are its small physical size and provide real-time reading [217]. Figure 3.8 shows a MOSkin dosimetry system and its schematic.

The characteristic of MOSkin has been previously reported. It showed excellent reproducibility and linearly for dose range of 50 cGy to 300 cGy. MOSkin presented stability response to various factors such as SSDs, field sizes, surface, radiation incident angles, and wedges [216]. It is found to be suitable for in vivo skin dosimetry in radiotherapy [217]–[219].

Similar to MOSFET principle, the difference between two threshold voltage values, $\Delta V_{th}$, was calculated to find the measured absolute dose using the following equation (3.5). The calibration factor was initially measured for 10 x 10 cm$^2$ field at a depth of $d_{max}$. The time gap between two signals was set at 30 seconds [220].

\[
Sensitivity \ (cGy) = \frac{\Delta V_{th} \ (mV)}{Calibration \ Factor \ (mV/cGy)} \tag{3.5}
\]
Figure 3.8. (a) A MOSkin dosimeter system and (b) MOSkin schematic.
CHAPTER 4

OPTIMIZING THE UPSTREAM AIR GAP
OF THE MAGIC PLATE 512 WHEN
OPERATING IN DOSE MODE

This chapter evaluates the impact of an air gap on the MP512 response when operating in dose mode for both photon beams and electron beams, i.e. output factor, PDD and beam profiles.

4.1 Introduction

Two-Dimensional (2D) silicon diode arrays implemented in radiation therapy quality assurance (QA) applications have a lot of advantages such as real time operation, the small size of the sensitive volume of a single diode and a large dynamic range. However, currently most diode arrays have a detector pitch that is not suitable for routine use in small treatment field applications [59], [122], [221], [222].

The CMRP introduced a monolithic high spatial resolution silicon detector called Magic Plate (MP512). A silicon monolithic detector, the MP512 has a high spatial resolution, yet a large, overall size and requires packaging that is associated with non-water equivalent materials and air gaps that can affect small field dosimetry measurements.

The air gap has a significant impact on small field dosimetry since a loss in charge particle equilibrium can occur depending on the size of the low density cavity [223], [224]. Several studies have shown that the reduction in dose is affected by increasing the air gap size.
Chapter 4: Optimization of the air gap upstream the MP512 when operate in dose mode

Charles et al [225] reported the effect of very small air gaps, less than 1 mm, on small field dosimetry used for stereotactic treatments. They simulated by Monte Carlo, the response of an optically stimulated luminescent dosimeter (OSLD) in a 6 mm x 6 mm 6 MV photon field. A dose reduction of about 5% for an air gap of 0.5 mm upstream of OSLD relative to the simulation with no air gap was observed. A 0.2 mm air gap caused a dose reduction of more than 2%. The authors also noted that the thin air gap can cause a significant reduction in the measured dose.

In addition, the air gap can be useful for correcting the response of non-water equivalent detectors in small field dosimetry [226]. Charles et al [226] demonstrated that silicon diode overresponse relative to water in small fields can be neutralised by a small upstream air gap which depends on the diode design and its packaging. That approach led to the “air diode” concept for stereotactic dosimetry [227].

In this chapter, the effect of the upstream air gap on the response of MP512 is investigated. The air gap size that changes the response of the MP512 to water in small field dose measurements was optimized for both photon beams and electron beams. The parameters tested were:

(a) Output factor (OF)

(b) Percentage Depth Dose (PDD)

(c) Wedge beam profile

The responses of all tests above with different sized air gaps upstream of the MP512 detector was measured in comparison with EBT3 film, the MOSkin, and an Ionisation Chamber (IC).
4.2 Materials and Methods

4.2.1 PMMA slabs for MP512 cover

The MP512 detector was placed between two polymethyl methacrylate (PMMA) slabs. The purpose of the PMMA envelope is to protect the detector from mechanical damage and to shield the sensor from ambient light [199]. Figure 4.1 shows the MP512 enveloped with the PMMA slabs and an example of the PMMA slabs with an aperture at the centre area. This aperture created an air gap between the detector itself and the PMMA slab as presented in the schematic diagram shown in Figure 4.2.
Chapter 4: Optimization of the air gap upstream the MP512 when operate in dose mode

By altering the size of this air gap it is possible to affect dosimetric measurements in different radiation fields. To investigate the best air gap size for the MP512 detector, the air gap thickness between the PMMA slab and the PCB used in this study was adjusted between 0.5 mm, 1.0 mm, 1.2 mm, 2.0 mm and 2.6 mm from the PCB surface. Taking into account the thickness of the silicon substrate is 0.45 mm, the actual air gap size above
the MP512 for the studies described here are therefore 0.05 mm, 0.55 mm, 0.75 mm, 1.85 mm and 2.15 mm.

4.2.2 MP512 equalisation

Before performing and dosimetric measurements, equalization of the detector is required because the MP512 consists of 512 silicon pixels and each pixel is connected to an individual readout input of the multichannel electronics. Based on this information, the radiation response is going to be related to the sensitivity of each individual pixel, the gain of its corresponding preamplifier channel, etc. Thus, it is possible to generate a flat field correction protocol to allow for the small variations of the overall detector system response on a pixel by pixel basis.

This equalization process adjusts the detector values and the detector configuration to generate a flat response output from the detector assuming a uniform beam is incident on the detector. Figure 4.3 shows the schematic of the equalisation setup.
Chapter 4: Optimization of the air gap upstream the MP512 when operate in dose mode

Figure 4.3. The schematic of the equalisation setup

The MP512 was placed at a depth of 10 cm in the solid water phantom with 10 cm back scattering. 200 MUs were delivered to a 20 x 20 cm² field size at SSD of 100 cm. The flat field correction can be calculated following equation (4.1), (4.2) and (4.3) [228], [229].

The response of the MP512, vector $X_i$, for all pixels is considered the same. The average response from all channels was calculated $(X)$. The equalization factor vector, $F_i$, was then

$$ F_i = \frac{X_i}{(X)} $$  (4.1)
And the equalised detector response for each pixel, \( X_{eq-i} \), was then

\[
X_{eq-i} = \frac{X_i}{F_i}
\]  

(4.2)

The detector uniformity can be calculated by the equation (4.3) where \( X_{cen} \) is the vector at a central pixel which is located at row 11, column 12.

\[
X_\% = \frac{X_{eq-i} - X_{cen}}{X_{cen}}
\]  

(4.3)

The variation in the response of all 512 detectors relative to the average response can then be calculated before and after flat field correction for any beam energy desired.

**4.2.3 Output Factor measurement**

The output factor (OF) presents the dose rate and the amount of radiation exposure produced by a treatment machine. It can be defined as the ratio of the dose per monitor unit (MU) for a given field size to the reference field size [230]. The reference field size in this study is 10x10 cm\(^2\) at the source to surface distance (SSD) of 90 cm [13]. Figure 4.4 shows the MP512 output factor measurement setup.
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The MP512 was placed on solid water phantom at the depth of 10 cm with an additional 10 cm of solid water to act as back scatter and was aligned at the centre of the beam. The OF was measured for square fields ranging from 0.5 x 0.5 cm$^2$ to 10 x 10 cm$^2$ and was deduced based on the response of the central pixel which is located at row 11 and column 12. The measurements were performed in 6 and 10 MV photon beams with a 600 MU/min dose rate. 100 MU was delivered with open field MLCs. The size of the air gap above MP512 detector was set at 0.5, 1.0, 1.2, 2.0 and 2.6 mm. The measurements were taken three times at least and directly compared with EBT3 films and MOSkin response under the same conditions.

Figure 4.4. The output factor measurement setup of Magic Plate 512 with various detector air gaps.
4.2.4 Percentage Depth Dose (PDD) for 6 MV and 10 MV photon beam

The PDD profiles were acquired using different air gap sizes upstream of MP512 detector. The air gaps used for the 6 MV photon beam was 0.5 mm and 2.6 mm, and for the 10 MV photon beam they were 1.2 mm and 2.6 mm. The MP512 was placed perpendicular to the direction to the central axis (CAX) of the beam at an SSD of 100 cm for field size of 2 x 2 cm$^2$, 5 x 5cm$^2$ and 10 x 10cm$^2$. The PDDs were obtained by scanning the MP512 from a depth of 0.5 cm to 10 cm. The PDDs were normalized to the MP response at the depth of $d_{\text{max}}$ for all photon energies investigated. The $d_{\text{max}}$ for 6 MV and 10 MV photon beam is 1.5 cm and 2.1 cm, respectively. For all irradiation geometries, 100 MU was delivered with a 600MU/min dose rate. The PDD measured by the MP512 with different air gaps were directly compared with the PDD response measured by an original Markus IC (PTW, Freiburg, Germany, model: N23343) for field sizes of 10 x 10 cm$^2$ and 5 x 5 cm$^2$. For the small field size of 2 x 2 cm$^2$ the results were compared to the EBT3 films to gain an understanding of the impact of any volume averaging effects as well as any beam perturbation effects from the window thickness of the ionisation chamber [231].

4.2.5 Wedge beam profile measurements for 6 MV photon beam

Wedge beam profile measurements were done using a 5 x 5 cm$^2$ radiation field size which was the smallest field the Linac can generate for enhanced dynamic wedge (EDW) field. The EDW profiles were produced by varying the jaw position and/or the output rate during the treatment dose delivery. The EDW of 15°, 45° and 60° were used for this study and generated by Varian Linac (model 2100IX). The MP512 was placed at a depth of 10 cm in solid water phantom and aligned on the central axis of the beam. A 100 MU was
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delivered at 100 cm SSD for each wedge angle for a 6 MV photon beam. The air gap sizes of 0.5 mm and 2.6 mm were used as part of this particular study.

To convert the MP512 response to dose (Gy) the MP512 responses were multiplied by the calibration factor (CF). To obtain the CF, the MP512 was placed at a depth of $D_{\text{max}}$, the dose-maximum depth, and exposed to a 10x10 cm$^2$ field size at the 100 cm SSD. The calibration factor can be calculated using equation (4.4) below where “MU” is the known MU delivered to the MP512 and “MP” is the average response of the central pixel of the MP512.

$$CF = \frac{MU}{MP}$$ (4.4)

The dose measured by the MP512 was directly compared with the independently calibrated EBT3 film response measurements made under the same conditions.

4.2.6 Percentage Depth Dose for electron beams

The PDD measurements in electron beam fields were performed in the same solid water phantom. The SSD was set to 100 cm. A 10 x 10 cm$^2$ applicator and a standard cerrobend cutout of 10 x 10 cm$^2$ were used to define the electron field dimensions. The MP512 was placed in the solid water phantom and aligned at the centre of the beam. The measurements were performed at a depth of 0.5 cm to 10 cm in a solid water phantom. The results were investigated for 6 MeV, 12 MeV and 20 MeV electron beams with air gap sizes of 0.5 mm and 2.6 mm. All measurements were performed at least three times. The results were compared with the PDD measured by Markus IC.
Chapter 4: Optimization of the air gap upstream the MP512 when operate in dose mode

4.2.6.1 Use of plastic phantom for depth dose measurement correction

The IAEA has recommended that to measure a central-axis depth dose for electron beam in a plastic phantom, the dosimeter reading at each depth must be scaled because for the electron beam, the water-to-air stopping power ratio, $S_{w,\text{air}}$, changes rapidly with depth [232]. In this part of the study, the electron beam depth dose measured by an ion chamber was followed using the code of practice for radiotherapy dosimetry, TRS-398 [75].

In this study, the solid water phantom; RMI-457 was used for the measurements. The equivalent depth in water, $Z_w$, can be calculated by equation (4.5) below.

$$Z_w = \frac{Z_{pl}}{C_{pl}}$$  \hspace{1cm} (4.5)

Where $Z_{pl}$ is the depth in the solid water phantom (cm) multiplied by $P_{pl}$.

$P_{pl}$ is the density of the plastic, $P_{pl}$ for the solid water (RMI-457) is 1.030 gcm$^{-3}$. $C_{pl}$ is the depth scaling factor, $C_{pl}$ for the solid water (RMI-457), which is 0.949 gcm$^{-3}$.

The reading from the Markus IC was multiplied by the fluence-scaling factor, $h_{pl}$, for certain plastics to find the equivalent reading at $Z_{rel}$ in water ($M_Q$) as presented in equation (4.6).

$$M_Q = M_{Q,pl}h_{pl}$$  \hspace{1cm} (4.6)

Where $M_{Q,pl}$ is the IC reading at the scaling depth in water ($Z_w$), and the $h_{pl}$ is the fluence scaling factor. The fluence scaling of the solid water (RMI-457) is 1.008.
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All IC current, $M_Q$, was then multiplied by the appropriate stopping-power-ratio, $S_{w,\text{air}}$, at each depth. Table 4.1 summarizes the stopping-power-ratio as a function of beam quality $R_{50}$ (g cm$^{-2}$) and relative depth $Z/R_{50}$ in water of each depth use for electron beam PDD measurement is this study [75].

Table 4.1. The water-to-air to stopping-power-ratio for 6 MeV, 12 MeV and 20 MeV as a function of beam quality $R_{50}$ and relative depth $Z/R_{50}$.

<table>
<thead>
<tr>
<th>Depth (cm)</th>
<th>$S_{w,\text{air}}$</th>
<th>Depth (cm)</th>
<th>$S_{w,\text{air}}$</th>
<th>Depth (cm)</th>
<th>$S_{w,\text{air}}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.50</td>
<td>1.0500</td>
<td>0.50</td>
<td>1.0000</td>
<td>0.50</td>
<td>0.97</td>
</tr>
<tr>
<td>0.70</td>
<td>1.0600</td>
<td>1.00</td>
<td>1.0100</td>
<td>1.00</td>
<td>0.98</td>
</tr>
<tr>
<td>1.00</td>
<td>1.0700</td>
<td>2.00</td>
<td>1.0400</td>
<td>2.00</td>
<td>0.99</td>
</tr>
<tr>
<td>1.20</td>
<td>1.0800</td>
<td>2.50</td>
<td>1.0500</td>
<td>2.60</td>
<td>1.00</td>
</tr>
<tr>
<td>1.30</td>
<td>1.0800</td>
<td>2.90</td>
<td>1.0600</td>
<td>4.00</td>
<td>1.02</td>
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<td>1.0900</td>
<td>3.90</td>
<td>1.0900</td>
<td>6.00</td>
<td>1.06</td>
</tr>
<tr>
<td>2.00</td>
<td>1.1100</td>
<td>4.50</td>
<td>1.1100</td>
<td>6.10</td>
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<tr>
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<td>5.00</td>
<td>1.1200</td>
<td>7.00</td>
<td>1.08</td>
</tr>
<tr>
<td>2.50</td>
<td>1.1400</td>
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<td>1.13</td>
</tr>
<tr>
<td>3.50</td>
<td>1.1800</td>
<td>10.00</td>
<td>1.3180</td>
<td>10.00</td>
<td>1.16</td>
</tr>
<tr>
<td>5.00</td>
<td>1.2700</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

*R_{50} = R_{50} \text{ in water} \times C_{\text{pl}}

**R_{50} \text{ of 6 MeV, 12 MeV and 20 MeV} = 2.18 \text{ gcm}^{-2}, 4.65 \text{ gcm}^{-2} \text{ and 7.97 gcm}^{-2}

4.3 Results and discussion

4.3.1 Output factor for 6 MV and 10 MV photon beam

The average of the four central pixels of the MP512 was evaluated for the OF measurements. The MP512 uncertainty was found to be 0.2% (1 s.d.) for all measurements. Figure 4.5 and Figure 4.6 show the OF measured by the MP512 at a depth of 10 cm in solid water phantom for different air gaps above detector compared to the EBT3 film and MOSkin (with no air gap above them). The response is normalized to the
response measured in a 10 x 10 cm\(^2\) field size, for 6 MV and 10 MV, respectively. Figure 4.5 illustrates that at small field sizes the OF measured with the MP512 reduces with increasing detector air gap. A significant effect of the air gap size has been observed for a 0.5 and 1 cm\(^2\) field size. The air gap has negligible effect for field sizes larger than 4 x 4 cm\(^2\) for the air gap of 0.5 mm, 1 mm and 1.2 mm above the detector. The MP512 with the air gap of 0.5 mm shows good agreement to the output factors measured with the EBT3 film and MOSkin within ±2\% (1 standard deviation) for very small field sizes up to 3 cm\(^2\) and not more than 3 \% for 3-5 cm\(^2\) and no difference with field size increasing (zero for 10x10 cm\(^2\)).

As expected, for small radiation fields of 0.5 x 0.5 cm\(^2\) and 1 x 1 cm\(^2\) the output factor reduces with the air gap increasing for the 10 MV photon beam as presented in Figure 4.6. The MP512 response with air gap size of 1.2 mm best matched the output factors measured with the EBT3 and MOSkin within ±2\% for field sizes smaller than 4x4 cm\(^2\) (1 standard deviation) for 10 MV photon beam fields.

With increasing of radiation field size, the effect of an electronic disequilibrium produced by air gap is diminishing. It is explained by the fact that laterally scattered radiation dominate response of the MP512 in comparison with lack of secondary electrons generated in an air gap. It is confirmed by experimental results. At field size of 0.5 x 0.5 cm\(^2\) and 1 x 1 cm\(^2\) the percentage different between MP512 and EBT was more than 12\% when the air gap size increased from 0.5 mm to 2.6 mm. Similar behavior was observed when compared with the MOSkin detector. While at field sizes 4x4 cm\(^2\) the output factor measured with MP512 difference of the output factors measured by EBT3 and MOSkin is about only 5-7\% when the air gap size increased from 0.5 mm to 2.6 mm and is not
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changing at all for 10x10 cm$^2$ filed size. Based on these studies the optimal air gap was selected to match OF measured with MP512 to the film.

(a)
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Chapter 4: Optimization of the air gap upstream the MP512 when operate in dose mode

Figure 4.5. Field size response of the MP512, EBT3 film and MOSkin for a 6 MV photon beam, normalized to the response measured in a 10 x 10 cm² field size at a depth of 10 cm in a solid water phantom for different air gaps of (a) 0.5 mm, (b) 1.0 mm, (c) 1.2 mm, (d) 2.0 mm and (e) 2.6 mm.
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(a) Output Factor vs. Field Size (cm²)

(b) % Difference vs. Field Size (cm²)

- MP512
- EBT3
- Moskin
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![Graph showing output factor vs. field size for MP512, EBT3, and Moskin, with % difference compared to MP512.](image)

- **Graph (c):** Output Factor vs. Field Size (cm²) for MP512, EBT3, and Moskin.
- **Graph (d):** % Difference vs. Field Size (cm²) for MP512 vs. EBT3 and MP512 vs. Moskin.
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Figure 4.6. Field size response of the MP512, EBT3 film and MOSkin for a 10 MV photon beam normalized to the response at 10 x 10 cm$^2$ field size at a 10 cm depth in a solid water phantom and air gap of (a) 0.5 mm, (b) 1.0 mm, (c) 1.2 mm, (d) 2.0 mm and (e) 2.6 mm.

4.3.2 Percentage Depth Dose (PDD) for 6 MV and 10 MV photon beams

Figure 4.7 and Figure 4.8 present the PDD measured with the MP512 in a solid water phantom for 0.5 mm and 2.6 mm air gap upstream of the detector for different field sizes in comparison with a Markus IC for 6 MV and 10 MV photon beams, respectively. All reading from the Markus IC has been corrected for over-response by using the corrected factor is given by Chen et al [210]. Similar results were observed for the photon beam PDD measurements. As the size of the air gap above the detector increased, the PDD demonstrated a detectable decrease field size of 5 x 5 cm$^2$ and 10 x 10 cm$^2$. The PDD for 2 x 2 cm$^2$ field was within ±3% (1 SD) of the EBT3 for both photon energies. For field
size 5x5 cm² and 10x10 cm², the PDD measured with the MP512 is within ±1.6% (1 SD) and ±1.5% (1 SD) of that measured using a Markus ionisation chamber for 6 and 10 MV fields respectively.
Figure 4.7. PDD measured with 0.5 mm and 2.6 mm air gap above the MP512 of 6 MV photons in comparison with an ionisation chamber and EBT3 film for field sizes of (a) 10x10 cm$^2$, (b) 5x5 cm$^2$ and (c) 2x2 cm$^2$. 
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(a) PDD (%)

(b) Difference (%)

Depth (mm)

1.2 mm
2.6 mm
IC
Chapter 4: Optimization of the air gap upstream the MP512 when operate in dose mode

Figure 4.8. PDD measured with 1.2 mm and 2.6 mm air gap above the MP512 of 10 MV photons in comparison with an ionisation chamber and EBT3 film for field sizes of (a) 10x10 cm$^2$, (b) 5x5 cm$^2$ and (c) 2x2 cm$^2$.

4.3.3 Wedge beam profile for photon beams

Figure 4.9 shows the beam profile measured in the wedge direction at a depth of 10 cm for the MP512 with a 0.5 mm and 2.6 mm air gap above the detector in comparison with EBT3 film. The EDW dose profile matches well with the EBT3 for the air gap of 0.5 mm, within ±1% (1 SD), except at the toe and heel region where the difference was within ±3% (1 SD) for all wedge angles. The difference increases with an increase in the air gap size. For the 2.6 mm air gap, the difference on the heel side was observed to be about ±10% (1 SD). The Wedge profiles show that if only the flattened area of the field is considered, the maximum difference between profile and the EBT3 film is within ±1%
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for small air gap size 0.5mm for all wedges angles. There is a significant difference in the shape of the wedge profile when a 2.6 mm air gap size is used.
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Figure 4.9. Wedge beam profiles measured with the MP512 with different air gaps in comparison with those measured using EBT3 film at a depth of 10 cm for 6 MV photon beam with a field size of 5x5 cm² (a) 15° Wedges, (b) 45° Wedges and (c) 60° Wedges.
4.3.4 Percentage depth dose for electron beams

The PDD measured by the MP512 in electron beams demonstrated no significant effect with increasing air gap above the MP512 for all energies. The correction for use of the plastic phantom for electron depth dose distributions follows the TRS398 instruction [75]. The results for both 0.5 mm and 2.6 mm air gap are within ±3% (1 SD) of similar measurements made using the Markus IC and are shown in Figure 4.10.
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Figure 4.10. PDD measured with MP512 and 0.5 mm and 2.6 mm air gap upstream of the detector on electron beams for a field size 10 x 10 cm² in comparison with a Markus ionisation chamber in a solid water phantom for electron beam energies of (a) 6 MeV, (b) 12 MeV and (c) 20 MeV.
4.4 Conclusion

The MP512 response with different air gaps upstream of the detector in a solid water phantom have been investigated in both photon and electron fields. The results obtained in this study show that the air gaps cause a measurable dose reduction for small radiation field sizes due to the loss in electron equilibrium. Based on these findings, we have tried to optimize the air gap size for a monolithic diode array detector, MP512 for both photon and electron fields. The studies confirmed that the MP512 monolithic diode array is suitable for QA of small fields in a phantom. The small air gap of 0.5 mm and 1.2 mm is the best air gap for small field dosimetry in 6 MV and 10 MV photon beams, respectively. However, the effect of air gap on electron beams is not significant due to electronic equilibrium conditions being fully established and maintained.
CHAPTER 5

DOSIMETRIC IMPACT OF MAGIC PLATE512 OPERATING IN TRANSMISSION MODE ON CLINICAL PHOTON BEAMS

This chapter describes, from a dosimetry perspective, the impact of the monolithic silicon detector, MP512 when operating in transmission mode, so named MP512 T. In particular I studied the effect of the MP512T on the surface dose as a function of different field sizes and the specific MP512T detector mounting position between the solid water phantom and the Linac head. As part of my work the unique transmission factor for the MP512T and the PCB were evaluated.

5.1 Introduction

The Magic Plate 512 has been previously reported in dose mode measurements for SRS/SBRT verification [168]. Further investigation, including the effect of the air gap size above the detector, has been discussed in the previous chapter of this thesis. The MP512 with the difference in the detector packaging called MP512T (described in Chapter 3) has been designed to be used as a transmission detector for real time dose reconstruction measurements.

Any radiation beam perturbation, in particular any surface dose increase, is of significant interest due to its impact on clinical outcomes associated with transmission type detectors [156], [162], [164], [233]. As a transmission-type detector, a detailed study for the MP512T is therefore required to assess its future clinical suitability.
Chapter 5: Dosimetric impact of MP512 operating in transmission mode on clinical photon beams

The MP512T, in contrast to the previously characterized transmission magic plate detector called MP121 [234] is essentially different in design and has a much improved spatial resolution. The details of the MP121 were described in Chapter 2. The MP512T is a monolithic silicon detector that can lead to different perturbation of the skin dose in megavoltage photon fields. Additionally, the MP121 was characterized only when the detector was attached to the Linac head. Unlike the MP121, the MP512T system is designed to be placed at various distances between the Linac head and the solid water phantom surface. Therefore, it is important to investigate, understand and quantify any effect on the skin dose of the different MP512T detector operating positions. This chapter will investigate the Magic Plate 512T when operating in transmission mode, on beam perturbation, in particular on the surface dose and beam transmission with different field sizes and positions between the solid water phantom and Linac head.

5.2 Materials and Methods

5.2.1 A concept of transmission movable high resolution

Figure 5.1. The concept of using a transmission monolithic silicon detector providing a flexible resolution by using a variable detector to surface distance, Dsd, by moving the detector between the Linac head and the patient. Figure 5.1 shows a concept of the transmission monolithic silicon detector providing flexible spatial resolution by changing detector distances from the solid water phantom (Dsd). Moving the detector along the beam axis between the patient surface and the Linac
head enables the effective spatial resolution of the detector monitoring the radiation field to change due to the beam divergence. As the tumour size decreases, moving the detector close to the patient provides a higher effective spatial resolution while allowing the monitoring of the entire treatment field.

![Diagram](image)

**Figure 5.1.** The concept of using a transmission monolithic silicon detector providing a flexible resolution by using a variable detector to surface distance, $D_{sd}$, by moving the detector between the Linac head and the patient.

The proposed movable transmission, high effective spatial resolution silicon monolithic detector has another advantage in comparison with the currently used transmission...
detectors mounted on the Linac head. By moving the detector below the Linac head, the contribution of electrons scattered from the head of the Linac on the response of the detector is minimized, and the detector response is mostly driven by the photon energy fluence, which should simplify the 3D dose reconstruction algorithm.

5.2.2 A movable stand detector holder

The MP512T detector was mounted on a movable stand in order for the detector to be positioned between the Linac head and the solid water phantom at any distance from the phantom surface. The movable stand is made from a PMMA plastic. The holder arms are 40 cm in length and can be adjusted to fit the detector assembly up to a width of 45 cm. The holder has the capability of moving in the vertical direction. When placing MP512T on the movable stand holder, only the detector itself was irradiated. Figure 5.2 shows the movable stand and its geometry.
5.2.3 The influence of MP512T on the surface dose

To obtain the surface dose, a Markus ionisation chamber (PTW, Freiburg, Germany, model N23343) was positioned at the surface of the solid water phantom at central axis (CAX) corresponding to the isocentre, with 100 cm SSD. The back scattering solid water phantom was 10 cm thick. The IC was read out by PTW UNIDOS model T10002-20713 electrometer. All readings from the Markus IC have been corrected for over response by using the corrected factor given by Chen et al. [210] described in Chapter 3.

The perturbation of the surface dose was reported as a percentage difference of the surface dose measured with MP512T in a beam to open field. Both MP512T and the Markus IC were aligned at the centre of the beam. All measurements were performed using a 6 MV photon beams from a Varian linear accelerator (Model 21XI). For each measurement, a
200 monitor unit (MU) was delivered. The MP512T distance from the solid water phantom surface was varied from 0.3 cm to 24 cm. The measurements were carried out for irradiation field sizes (IFS) of 5 x 5 cm², 8 x 8 cm² and 10 x 10 cm² with the MLC matched with the Linac jaws. The measurement setup is shown in Figure 5.3.
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(b)

Figure 5.3. The measurement setup (a) with and (b) without MP512T in a beam.

The radiation field size is defined at the SSD of 100 cm. Thus, the effective irradiation field size at MP512T position depends on the distance away from the solid water phantom and ranges about from 3 x 3 cm$^2$ to 7 x 7 cm$^2$ at the Linac head placement. Table 5.1 presents the actual field size on MP512T detector at various distances from the solid water phantom surface.
Table 5.1. The field size of the MP512T detector at various distances from the solid water phantom surface.

<table>
<thead>
<tr>
<th>$D_{sd}$ (cm)</th>
<th>Field size ($\text{cm}^2$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.00</td>
<td>5.00</td>
</tr>
<tr>
<td>0.30</td>
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</tr>
<tr>
<td>24.00</td>
<td>3.80</td>
</tr>
</tbody>
</table>

To examine the reproducibility of the Markus IC the readings were acquired three times at least under the same conditions. The detector measurement uncertainty was found to be $\pm 0.2\%$ (1 s.d.).

### 5.2.4 The influence of the MP512T on the surface dose when placed face up and face down

The influence of the MP512T on the surface dose when placed at different $D_{sd}$ face up and face down were evaluated. The set of measurements as section 5.2.3 was repeated. Figure 5.4 shows the schematic of the surface dose measurement setup with MP512T in a beam (a) face up; (b) face down. For each position and field size, the readings are obtained at least three times, and the average was calculated.
5.2.5 Effect of Printed Circuit Board on MP512T surface dose measurement

To evaluate the effect of the 0.5 mm thick printed circuit board (PCB) only on the surface dose, the PCB without the mounted silicon detector was placed on the movable stand. The surface dose measurements were performed using the Markus IC in a solid water phantom for open fields and with the PCB in the beam similar to that described in section 5.2.3.

5.2.6 The transmission factor measurement

The transmission factor (TF) of the MP512T and the PCB were investigated by measuring a ratio of the doses at $d_{\text{max}}$ corresponding to the field 10 cmx10 cm, i.e. at depth in a phantom 15 mm depth with and without MP512T in a beam for radiation field size 5x5 cm$^2$, 8x8 cm$^2$ and 10x10 cm$^2$ and SSD of 100 cm for a 6 MV photon beam. The MP512T was placed in the beam at different $D_{\text{sd}}$ ranging from 0.3 cm to 24 cm. A Farmer IC (Model 2571A) was used for dose measurements. The same set up was repeated at a depth of 10 cm and a source axial distance (SAD) of 100 cm for a 6 MV photon beam.
5.3 Results

5.3.1 The influence of MP512T on the surface dose

Figure 5.5 shows the percentage difference of surface dose with and without MP512T in the beam path as a function of field size and distance from the solid water phantom surface. The percentage difference was calculated follow equation (5.1).

\[
\text{%Diff} = \left( \frac{\text{with MP512T} - \text{without MP512}}{\text{without MP512}} \right) \times 100
\]

The maximum difference of surface dose was nearly 30% (1 standard deviation), and this was found at the distance of 0.3 cm especially in the large 10 x 10 cm$^2$ field. The difference in surface dose decreased with increasing distance of MP512T from the solid water phantom surface. At D$_{sd}$>18 cm the difference was less than 5% (1 standard deviation) for all IFSs. At a small field size of 5 x 5 cm$^2$, the percentage difference was within ±1 % (1 standard deviation).
Chapter 5: Dosimetric impact of MP512 operating in transmission mode on clinical photon beams

Figure 5.5. The percentage difference of surface dose with and without MP512T in a beam as a function of distance of MP512T from the phantom surface and field size for a 6 MV photon beam.

The effect of MP on surface dose increasing is increasing with reducing distance between Magic Plate and surface of the phantom. This explain by the fact that surface dose has three component one is due to electron contamination originated from scattered electrons in a linac head, second is from electron generated in air between linac head and the phantom surface and third due to electrons scattered from MP. On the other hand MP is attenuated and scattered partially electrons originated above MP. Based on obtained results attenuation of electrons by MP is not essential and combined effect is increasing of the skin dose with reducing Dsd. It is explained by the fact that size of the photon field incident on a MP is increasing with Dsd decreasing that is leading to more secondary electrons originated from MP scattered to the phantom surface and surface dose increasing respectively. This effect is more pronounced with filed size increasing that is...
reflected in a Fig 5.5. For Dsd larger than 18 cm the effect of electron scattered from the MP is not essential in comparison with other two electron components as a photon filed size seeing by MP is small was observed.

5.3.2 The influence of the MP512T on the surface dose when placed face up and face down

Figure 5.6 shows the percentage difference of surface dose when MP512T is placed face-up and face-down at various distances from the phantom surface and different IFSs. The percentage difference was calculated follow equation (5.2).

\[
\text{\%Diff} = \left( \frac{\text{MP512 face down} - \text{MP512 face up}}{\text{MP512 face up}} \right) \times 100
\]

The difference was within 2.5 % (1 standard deviation) for all distances and field sizes.

![Graph showing percentage difference of surface dose](image)

Figure 5.6. The percentage difference of surface dose when MP512T is face-up and face-down in the beam as a function of distance from the phantom surface and field size for a 6MV photon beam.
Difference in skin dose changing for face down and face up configuration has tendency of slight skin dose increasing for face down configuration for smaller fields when more electron scattered down from the silicon MP not attenuated by PCB than in case of face up configuration. For 10cmx10 cm field size in MP plane is always larger than size of the silicon MP for all $D_{sd}$ and total electron scattering towards the phantom surface do not make such difference as scattering from PCB outside from the silicon MP is the same for both configurations.

5.3.3 The effect of printed circuit board on the surface dose

Figure 5.7 shows the percentage difference of the surface dose measured with and without the PCB in the beam. The percentage difference was calculated follow equation (5.3).

$$\%\text{Diff} = \left( \frac{\text{with PCB} - \text{without PCB}}{\text{without PCB}} \right) \times 100$$  \hspace{1cm} (5.3)

Similarly, to Figure 5.5, the surface dose difference increased when the PCB was closer to the phantom surface. At PCB distances of more than 18 cm, the percentage difference is close to zero for all IFSs. At a PCB distance of 0.3 cm, the surface dose increased by about 15% (1 standard deviation) for all IFSs.
Figure 5.7. The percentage difference of the surface dose with and without the PCB in a beam as a function of distance of the MP512T from the phantom surface and for various field sizes in a 6 MV photon beam.

The effect of PCB only of surface dose increasing with Dds decreasing for all field sizes is explained as in 5.3.1

### 5.3.4 The transmission factor measurement

At the 6 MV d\(_{\text{max}}\) depth, the relative dose difference increases slightly with decreasing of the distance (from 18 cm to 0.3 cm) between the phantom surface and the MP512T (or blank PCB). The TF was calculated follow equation 5.4.

\[
TF = \left( \frac{\text{with MP512}}{\text{without MP512}} \right)
\]  

(5.4)

For D\(_{\text{ad}}\) < 18 cm the TF changes by 1.5-2.0 % (1 standard deviation) and 0.5% (1 standard deviation) for MP512T and PCB respectively for all IFSs and all distances above 18 cm.
is close to 1. These results are presented in Table 5.2. Similar behaviour of the transmission factor is observed at a depth of 10 cm as shown in Table 5.3.

Table 5.2. Measured TF at $d_{\text{max}}$ for 6 MV photon beam, SSD =100 cm. The TF is presented separately for various distances and IFSs for MP512T and the PCB.

<table>
<thead>
<tr>
<th>$D_{sd}$ (cm)</th>
<th>MP512T</th>
<th>PCB</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>5x5 cm$^2$</td>
<td>8x8 cm$^2$</td>
</tr>
<tr>
<td>0.30</td>
<td>1.0130</td>
<td>1.0151</td>
</tr>
<tr>
<td>4.50</td>
<td>1.0120</td>
<td>1.0141</td>
</tr>
<tr>
<td>9.00</td>
<td>1.0096</td>
<td>1.0104</td>
</tr>
<tr>
<td>13.50</td>
<td>1.0040</td>
<td>1.0047</td>
</tr>
<tr>
<td>18.00</td>
<td>0.9985</td>
<td>0.9993</td>
</tr>
<tr>
<td>20.00</td>
<td>0.9981</td>
<td>0.9990</td>
</tr>
<tr>
<td>22.00</td>
<td>0.9975</td>
<td>0.9977</td>
</tr>
<tr>
<td>24.00</td>
<td>0.9971</td>
<td>0.9980</td>
</tr>
</tbody>
</table>
Table 5.3 Measured TF at a depth of 10 cm for 6 MV photon beam, SAD = 100 cm. The TF is presented separately for various distances and IFSs for MP512 and the PCB

<table>
<thead>
<tr>
<th>D_{sd} (cm)</th>
<th>MP512T</th>
<th>PCB</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>5x5 cm²</td>
<td>8x8 cm²</td>
</tr>
<tr>
<td>0.30</td>
<td>1.0190</td>
<td>1.0200</td>
</tr>
<tr>
<td>4.50</td>
<td>1.0182</td>
<td>1.0190</td>
</tr>
<tr>
<td>9.00</td>
<td>1.0132</td>
<td>1.0147</td>
</tr>
<tr>
<td>13.50</td>
<td>1.0069</td>
<td>1.0076</td>
</tr>
<tr>
<td>18.00</td>
<td>1.0022</td>
<td>1.0032</td>
</tr>
<tr>
<td>20.00</td>
<td>1.0020</td>
<td>1.0022</td>
</tr>
<tr>
<td>22.00</td>
<td>1.0010</td>
<td>1.0012</td>
</tr>
<tr>
<td>24.00</td>
<td>0.9993</td>
<td>0.9997</td>
</tr>
</tbody>
</table>

5.4 Discussion

Quality Assurance (QA) in advanced treatment radiotherapy techniques such as SRS and SBRT is complicated due to the small field delivery using IMRT or VMAT for SBRT and high definition of the MLCs and small cones for SRS. Thus, the treatment verification requires high spatial resolution QA tools, which accurately provide the relevant dose information in real-time during the treatment delivery for each gantry angle. Thus, a new
QA device, the monolithic silicon pixelated detectors MP512T was introduced. The MP512T provides a variable, yet high, effective spatial resolution due to its ability to be placed at different positions in the beam between the Linac head and the patient. In this way, the beam projection at any depth is within the area of silicon detector or PCB and avoids the effect of attenuation by a PMMA frame (Figure 5.1). These detectors allow one to obtain a variable effective spatial resolution from 2 mm to 4 mm depending on the position of the detector on the beam axis relative to the Linac head. Another advantage of this approach is the reduction of the contribution of scattered electrons from the Linac head to the response of the transmission detectors. The thin silicon substrate 0.45 mm and 0.5 mm PCB is a prerequisite to minimize any beam perturbation.

It was demonstrated that the MP512T and the PCB alone increases the surface dose due to Compton electrons originating from the silicon and the PCB. The partial contribution of the PCB alone led to the rise in the surface dose of about 60% compared to the increase in the surface dose from MP512T (Figure 5.5 and Figure 5.7). Taking into account that the Compton electrons, in this case, are mostly of a MeV energy range, it suggests that an opening or recess in the PCB under the silicon monolithic detector active area is recommended to further reduce the skin dose excess by a factor of two for all the IFSs considered. We also demonstrated that in the face-up or face-down orientation the MP512T makes only a 2% (1 standard deviation) difference in the excess surface dose (Error! Reference source not found. 5.6) and will be close to zero if an opening or recess is introduced in the PCB substrate. A thin light protective coating should be introduced above the silicon detector to avoid stray light influencing the detector.
response. It can easily be achieved by adding a black filler to the thin layer of resin protecting the silicon detector.

The transmission coefficient of the MP512T measured at $d_{\text{max}}$ 15 mm is close to 1 with a deviation of about 1.010-1.020 with decreasing distance between the MP512T and the phantom surface below 18 cm. The effect of the MP on dose modification at depth 15 mm and 100 mm was less than 1% for $D_{\text{sd}}$ large than 13.5 cm that is within dose modification tolerance and less than 2.2% for all other $D_{\text{sd}}$ (see Tables 5.2 and 5.3). It is recommended for in vivo application not to use $D_{\text{sd}}$ less than 13.5 cm that is safe in terms of avoiding collision with the patient body and for in a phantom QA to introduce correction to measured dose in case of measured dose based on transmission MP for $D_{\text{sd}}$ less than 13.5 cm. Providing an opening in the PCB under the silicon monolithic detector will make the transmission coefficient closer to 1 for any placement of the proposed transmission detectors between the patient and Linac head.

5.6 Conclusions

The MP512T detector used as a transmission detector with a variable spatial resolution of up to 1 mm at variable positions between the Linac head has been introduced. The MP512T is characterized by its minimal beam perturbation. Spatial resolution in dosimetry of the small photon beams can be improved by moving the MP512T along the beam axis with the best spatial resolution reported when the detector is closest to the surface of the patient. Further reduction of the skin dose excess can be achieved by reducing the silicon substrate thickness to 0.3 mm and utilizing a drop-in packaging style
for the detector [228] on the PCB with the recess accommodating the monolithic silicon detector.

The mechanical realization of the proposed transmission detector in a clinical scenario is still to be done but will be straightforward and will be realized by a telescopic jig attached to the Linac head slot, and wireless reader developed at CMRP similar to other transmission [156], [160], [164], [235].
This chapter extends the work of the previous chapter. The MP512T placed on the movable stand position between the Linac head and the solid water phantom will completely use with the data acquisition system. This chapter discusses the correlation of transmission mode response (TM) and dose mode response (DM) of the MP512T for dose prediction at $d_{\text{max}}$.

6.1 Introduction

The quality assurance (QA) of SRS and SBRT treatments are complex due to the use of small field delivery techniques using IMRT or VMAT for SBRT employing either high definition MLCs and/or small cones in the case of SRS, [37], [41], [177], [236], [237]. Many devices have been developed for pre-treatment verification such as two-dimensional (2D) detector arrays based on ionisation chamber (IC) and semiconductor detectors [59], [117], [122], [238]–[240].

There is a considerable demand for real-time dose delivery verification. Such QA technology enables a real-time detection of major errors in the delivered dose [62].
particular Electronic Portal Imaging Devices (EPIDs) have been used for real-time verification. The method can be used for fixed gantry IMRT treatment, and it is not yet available for dynamic arc treatment [153], [154], [241].

A real-time dose measurement can also be carried out using transmission-type detectors where the detector is positioned in the photon beam between the Linear Accelerator (Linac) head and the patient. Commercially available transmission-type detectors such as the David system (PTW-Freiburg, Germany), Dolphin system (IBA Dosimetry, Germany), the Compass system (IBA, Dosimetry), the Delta4 (ScandiDose, Sweden), and the integral quality monitoring system (IQM) are based on pixelated ionisation chambers and semiconductor diode arrays. The use of these systems result in a change of beam characteristics and lead to an increase in the surface dose [156], [157], [160], [162], [164]. Although, some transmission QA dosimetry devices are available and perform well for larger fields, their spatial resolution and resulting perturbation of the radiation field due to their design, limits their utilization as a part of a routine clinical practice for small field real-time treatment verification.

The aim of this chapter is to investigate the possible correlation of the MP512T when using it in transmission mode (TM) and dose mode (DM) for different $D_{\text{sd}}$ and treatment field sizes. The dose at $d_{\text{max}}$ for regular field sizes and intensity modulated treatment fields has therefore been calculated using a direct linear correlation found between the MP512 DM and TM measurements.


6.2 Material and methods

6.2.1 Experimental Setup

The MP512T was placed on the movable stand made form a PMMA plastic, as described in chapter 5, so the detector can be positioned between the Linac head and the solid water phantom at various distances from the phantom surface. Compared to other transmission detectors where the detector is mounted on the Linac head, the use of a movable transmission monolithic detector that can be mounted away from the Linac head reduces electron scatter, and therefore the photon energy fluence can be measured more accurately.

The changes in the surface dose observed when placing the MP512T in the air between the Linac head and the phantom has been discussed in the previous chapter. The MP512T detector produced only minimal perturbations and was fully transparent for 6 MV photon beams when placed at a $D_{sd}$ of $\geq 18$ cm above the phantom. The observed surface dose increased by about 5% and the difference was close to zero for small field sizes of less or equal to 5x5 cm$^2$. The transmission coefficient of the MP512T measured at $d_{max}$ is close to 1.00 at $D_{sd}$ of $\geq 18$ cm, and with a deviation of about 1.010-1.020 when decreasing $D_{sd}$ below 18 cm.

The measurement setup is shown in Figure 6.1. The MP512T was connected to a Data Acquisition System (DAQ) designed and built at CMRP. The system is based on a multichannel electrometer chip. The DAQ consists of a Field Programmable Gate Array (FPGA) that is read out in parallel by four analogue-to-digital converters (ADC). The FPGA provides the clock and synchronization circuit for the DAQ allowing its synchronization with the sync signal of the Linac. This guarantees that the charge
generated in the detector is only acquired when the electron gun fires, avoiding any loss of signal or unnecessary integration of dark current or electromagnetic induced noise. For further details regarding the DAQ, refer to Fuduli et al. [201].
Figure 6.1. (a) TM measurement setup of the MP512T allowing various distances between the Linac head and solid water phantom surface, (b) DM measurement setup of MP512T at a depth of $d_{\text{max}}$.

All experiments were performed on a Varian (Model 21XI) accelerator using 6 MV photon beams. Since the MP512T has a thickness of 0.45 mm it is, therefore, possible to operate it in both transmission mode and dose mode. The TM responses were measured by placing the MP512T on the movable stand holder positioned at various $D_{sd}$ ranging from 0.3 cm to 24 cm. The MP512T was covered with a black plastic sheet, 80 µm thick making the detector more light tight.

For the DM response, the MP512T was positioned in a homogenous solid water phantom at depth of $d_{\text{max}}$ (1.5 cm for 6 MV). As discussed in chapter 4, the air gap above the detector
was set to 0.5mm so that electronic equilibrium is fully established for measurements of output factors and PDD (within ±2%) [242], [243].

The TM and DM response measurements were performed for field sizes of $2 \times 2 \text{ cm}^2$, $3 \times 3 \text{ cm}^2$, $5 \times 5 \text{ cm}^2$, $8 \times 8 \text{ cm}^2$ and $10 \times 10 \text{ cm}^2$, and the treatment field size was defined at a source to surface distance (SSD) of 100 cm. Each measurement was obtained delivering 200 MU. All measurements were repeated in triplicate, and the resulting standard error was calculated. To find the correlation between DM and TM measurements the ratio of DM and TM, measurements were evaluated for each experimental setup.

6.2.2 Dose calculation for regular fields

The TM response was measured for regular field sizes of $1 \times 1 \text{ cm}^2$ and $4 \times 4 \text{ cm}^2$ at a $D_{\text{sd}}$ of 4 cm and 24 cm. The DM response at $d_{\text{max}}$ was then calculated using the DM/TM correlation shown in Equation 6.1. Note that these field sizes were not part of the measurement set used to obtain the correlation. The field size of $1 \times 1 \text{ cm}^2$ lies outside the measurement set and the expected response DM was extrapolated. While the $4 \times 4 \text{ cm}^2$ field size lies between 2 of the field sizes of the measurement set and interpolation was used to determine the expected DM response. The calculated doses were compared with the measured dose determined using the MP512T detector placed in solid water at $d_{\text{max}}$.

6.2.3 Dose calculation in IMRT fields

To determine the delivered dose at $d_{\text{max}}$ from MP512T transmission measurements, intensity modulated fields used to treat a malignant base of skull chordoma were delivered with the TM in place. The plan consists of 6 static fields and delivering a nominal dose
1.8 Gy per fraction to the target volume (12.40 cm$^3$). All IMRT fields were delivered with the gantry set to 0° (gantry pointing vertically downward toward the phantom surface) for a $D_{sd}$ of 4 and 24 cm. A gamma evaluation with criteria of 1%/1mm, 2%/2mm and 3%/3mm was used to compare the measured dose distribution at $d_{max}$ predicted from transmission measurements to the dose computed at $d_{max}$ using the TPS and the dose measured by EBT film.

6.3 Results

6.3.1 The ratio of DM and TM response

The ratio of the DM to TM response for the central pixel of the MP512T at various $D_{sd}$ and field sizes is shown in Figure 6.2. The ratio of the DM to TM response for off-central pixel including; -4mm, 4mm, -8mm, 8mm, -10mm, 10mm, -14mm and 14mm is shown in Figure 6.3, Figure 6.4, Figure 6.5, Figure 6.6, Figure 6.7, Figure 6.8, Figure 6.9, and Figure 6.10 respectively.
Figure 6.2. The ratio of DM and TM response at central pixel measured by MP512T at $d_{\text{max}}$ for different $D_{\text{sd}}$, with constant irradiation field sizes.

Figure 6.3. The ratio of DM and TM response at -4mm off-central pixel measured by MP512T at $d_{\text{max}}$ for different $D_{\text{sd}}$, with constant irradiation field sizes.
Figure 6.4. The ratio of DM and TM response at 4mm off-central pixel measured by MP512T at $d_{\text{max}}$ for different $D_{sd}$, with constant irradiation field sizes.

Figure 6.5. The ratio of DM and TM response at -8mm off-central pixel measured by MP512T at $d_{\text{max}}$ for different $D_{sd}$, with constant irradiation field sizes.
Chapter 6: QA of an IMRT clinical scenario using MP512T with variable spatial resolution

Figure 6.6. The ratio of DM and TM response at 8mm off-central pixel measured by MP512T at $d_{\text{max}}$ for different $D_{sd}$, with constant irradiation field sizes.

Figure 6.7. The ratio of DM and TM response at -10mm off-central pixel measured by MP512T at $d_{\text{max}}$ for different $D_{sd}$, with constant irradiation field sizes.
Figure 6.8. The ratio of DM and TM response at 10mm off-central pixel measured by MP512T at $d_{\text{max}}$ for different $D_{\text{sd}}$, with constant irradiation field sizes.

Figure 6.9. The ratio of DM and TM response at -14mm off-central pixel measured by MP512T at $d_{\text{max}}$ for different $D_{\text{sd}}$, with constant irradiation field sizes.
Figure 6.10. The ratio of DM and TM response at 14mm off-central pixel measured by MP512T at $d_{\text{max}}$ for different $D_{\text{sd}}$, with constant irradiation field sizes.

The measurements points for each field size were fitted with a trend line using the least squares method. Table 6.1 shows the resulting slope ($M$) and the DM/TM axis intercept ($B_A0$) for the central pixel of the MP512T at $d_{\text{max}}$ for all field sizes. The uncertainty of the calculated slope was within 0.21% (1SD) and 0.18% (1SD) for the intercept.

Table 6.1. Slope ($M$) and DM/TM axis intercept ($B_A0$) at $d_{\text{max}}$ for the central pixel of the MP512T for various field sizes

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Field Size (cm$^2$)</th>
<th>2</th>
<th>3</th>
<th>5</th>
<th>8</th>
<th>10</th>
</tr>
</thead>
<tbody>
<tr>
<td>$M$</td>
<td></td>
<td>0.0199</td>
<td>0.0210</td>
<td>0.0180</td>
<td>0.0191</td>
<td>0.0201</td>
</tr>
<tr>
<td>$B_A0$</td>
<td></td>
<td>1.1649</td>
<td>1.1437</td>
<td>1.0281</td>
<td>0.9686</td>
<td>0.9530</td>
</tr>
</tbody>
</table>
From the results, the ratio of off-center pixels shows a similar behaviour to that of the central pixel. The average difference between the central pixel slope and off-center pixel slope was within ±1.79% (1 standard deviation).

### 6.3.2 The DM and TM correlation

As can be seen from Table 1, the slope $M$ of the trend lines for DM/TM versus $D_{sd}$ only weakly depends on the field size, and hence we have chosen to work with an average value of the slope $M$. However, the same is not true for DM/TM axis intercept. Using an average value, the slope $M$ and the specific values for the DM/TM axis intercept in Table 1, one can predict the dose at $d_{max}$ using a transmission measurement as follows:

$$DM = (TM)_{D_{sd}}(B_{A0} - MD_{sd}) \tag{6.1}$$

Here $M$ is the average slope for all field sizes, which was found to be 0.01960. DM/TM axis intercept $B_{A0}$ for an arbitrary equivalent field size as a function of the field area can be found from the piecewise polynomial fit to the data shown in Figure 6.11.
Chapter 6: QA of an IMRT clinical scenario using MP512T with variable spatial resolution

Figure 6.11. The relation of the DM/TM axis intercepts ($B_{Ao}$) as a function of field area ($A$).

This data is well described by the following piecewise polynomial fits with each fit having an $R^2$ value of 1:

$$B_{Ao} = \begin{cases} 
-0.00014214286 \cdot A^2 - 0.00239214286 \cdot A + 1.17674285714 & \text{for } 0 \text{ cm}^2 \leq A \leq 25 \text{ cm}^2 \\
0.00001456410 \cdot A^2 - 0.00282184615 \cdot A + 1.08954358974 & \text{for } 25 \text{ cm}^2 < A \leq 100 \text{ cm}^2 
\end{cases}$$

(6.2)

The “kink behaviour” on Fig 6.11 is possibly due to the fact that all measurements were related to $D_{max}$ depth 15 mm that is actually $D_{max}$ for 10cmx10cm field only. Further investigation will be carried out for obtaining relation (6.1) for true $D_{max}$ for all filed sizes, however it is outside of this thesis.
6.3.3 Dose calculation for regular fields

For regular field size dose calculation, the responses at \( d_{\text{max}} \) were calculated when the TM responses were measured at the detector to surface distances of 4 and 24 cm. The DM/TM axis intercept \( B_{A0} \) for the 1x1cm\(^2\) and 4x4cm\(^2\) field size were obtained from Equation 6.2. Note that the \( B_{A0} \) value for the 1x1cm\(^2\) field size results from extrapolation while that for the 4 x 4 cm\(^2\) field size is obtained by interpolation as shown in Table 6.2.

<table>
<thead>
<tr>
<th>A (cm(^2))</th>
<th>( B_{A0} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1.2300</td>
</tr>
<tr>
<td>4</td>
<td>1.0900</td>
</tr>
</tbody>
</table>

The detector responses were multiplied by the calibration factor to convert the measured detector signal to the absorbed dose (Gy). The dose calculation for regular fields is shown in Table 6.3. For field size 1x1 cm\(^2\) the difference between measured dose and calculated dose was -2.18% at 4cm \( D_{sd} \) and 0.63% and 24 cm \( D_{sd} \). For field size 4x4 cm\(^2\) the difference between measured dose and calculated dose was 0.93% at 4cm \( D_{sd} \) and 1.95% and 24 cm \( D_{sd} \).
Table 6.3. The measured and calculated dose response of MP512 for central pixel at $d_{\text{max}}$ for field size 1x1 cm$^2$ and 4x4 cm$^2$

<table>
<thead>
<tr>
<th>$D_{sd}$ (cm)</th>
<th>Field Size 1x1 cm$^2$</th>
<th>Field Size 4x4 cm$^2$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Measured</td>
<td>Calculated</td>
</tr>
<tr>
<td>4</td>
<td>0.8157</td>
<td>0.7979</td>
</tr>
<tr>
<td>24</td>
<td>0.8157</td>
<td>0.8208</td>
</tr>
</tbody>
</table>

The absorbed dose for the 1x1 cm$^2$ field size was also compared with the dose measured by EBT3, the difference being within ±1.73% for both detectors to surface distances.

6.2.4 Dose calculation for IMRT fields

For the IMRT dose calculation, the measurement and calculation were evaluated only at the central fragment of the plan since the active MP512T detector area is 5.2 x 5.2 cm$^2$. The plan consists of 6 delivery gantry angles. The equivalent field size ($A_{eq}$) for each gantry angle was calculated using the standard equivalent square relationship shown in equation (6.3) [244].

$$A_{eq} = \frac{2xy}{x+y}$$  \hspace{1cm} (6.3)

DM/TM axis intercepts, $B_{A0}$, for any square field were obtained from the second order polynomial fits shown in Equation 6.2. Table 6.4 shows the equivalent field for each gantry angle and the calculated $B_{A0}$. 109
Table 6.4. The equivalent field ($A_{eq}$) for each angle and the calculated $B_{A0}$

<table>
<thead>
<tr>
<th>Gantry (°)</th>
<th>Field Size (cm$^2$)</th>
<th>Equivalent Field size (cm$^2$)</th>
<th>$B_{A0}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>150</td>
<td>5x7.5</td>
<td>6x6</td>
<td>1.0068</td>
</tr>
<tr>
<td>100</td>
<td>5x7.5</td>
<td>6x6</td>
<td>1.0068</td>
</tr>
<tr>
<td>60</td>
<td>5x6</td>
<td>5.45x5.45</td>
<td>1.0186</td>
</tr>
<tr>
<td>300</td>
<td>5x7.5</td>
<td>6x6</td>
<td>1.0068</td>
</tr>
<tr>
<td>260</td>
<td>5x5.5</td>
<td>5.23x5.23</td>
<td>1.0233</td>
</tr>
<tr>
<td>210</td>
<td>5x5.5</td>
<td>5.23x5.23</td>
<td>1.0233</td>
</tr>
</tbody>
</table>

All calculated planar doses for each gantry at $d_{max}$ were accumulated and compared with the planar dose extracted from the TPS, the planar dose measured by EBT3 film and the measured dose measured by MP512T at $d_{max}$ by using Matlab.

Figure 6.12 shows the example the MP512T calculated planar dose at $d_{max}$ from MP512T transmission measurement $D_{sd} = 4$ cm compared to the dose from TPS with the gamma evaluation of 3%/3mm using Matlab. All gamma evaluation is shown in Table 6.5. The script used for gamma evaluation is in appendix B.
Figure 6.12. The comparison of MP dose calculated when placing MP512T at $D_{sd} = 4$ cm with the planer dose from TPS.

Gamma criteria: 3\%3mm

Threshold: 10\%

Gamma passrate: 98.14\%

Average gamma: 0.31
Table 6.5. The gamma evaluation for IMRT plan dose calculation at $D_{sd}$ of 4 and 24 cm

<table>
<thead>
<tr>
<th>$D_{sd}$ (cm)</th>
<th>3%/3mm</th>
<th>2%/2mm</th>
<th>1%/1mm</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>MPcal-TPS</td>
<td>MPcal-EBT3</td>
<td>MPcal-MPmeasured</td>
</tr>
<tr>
<td>4</td>
<td>98.14%</td>
<td>96.89%</td>
<td>99.79%</td>
</tr>
<tr>
<td>24</td>
<td>97.22%</td>
<td>97.53%</td>
<td>99.69%</td>
</tr>
</tbody>
</table>
Chapter 6: In vivo QA of IMRT with variable spatial resolution

6.4 Discussion

In this chapter it was demonstrated that by placing the monolithic pixelated silicon detector, MP512T at different positions between the Linac head and the solid water phantom surface, a variable effective spatial resolution is provided. This allows one to probe small radiation fields intersecting the detector at different spatial resolutions. In particular, a functional relationship between the dose in phantom, DM, and the transmission dose, TM, as a function of detector to surface distance \( D_{sd} \) was derived. The \( D_{sd} \) values were varied from 0.3 cm to 24 cm and the area of the radiation field that can be used to derive the dose in phantom for arbitrary radiation fields was found to be based on the transmission dose measurements. The calculated IMRT planar dose at \( d_{max} \) illustrated good agreement when compared with dose predicted by the TPS and measured using EBT3 film. The poor pass rate for the gamma criteria of 1%/1mm can be explained by possible submillimeter misalignment of registration of the film in the MP512T frame of reference in TM and for the case of the TPS, by the voxel size in the dose calculation which is larger than 1mm\(^3\). The registration of the same MP512T in the DM coordinate system relative to the MP512T in TM is more accurate however. In addition, both detectors have a pixel size of 0.5x0.5 mm\(^2\), which is less than 1x1 mm\(^2\). The obtained result confirms that the MP512T can provide dose mapping at \( d_{max} \) with a spatial resolution of order 1 mm and this is important for a small treatment fields such as those used in SRS.

Since SRS and SBRT employ small radiation fields to conform the radiation dose tightly to the target, meaningful treatment verification requires QA tools having a high spatial resolution that accurately provide relevant dose information. The MP512T employed in this work as a transmission detector has the potential to be used for real-time dose
monitoring of small radiation fields, opening the door to real time 3D dose verification, based on transmission measurements as the treatment is being delivered to the patient. Hence, this work represents the first step in the development of real-time 3D dose reconstruction based on TM measurements.

Note that the effective measurement area of measurement \( A_o \) at \( d_{\text{max}} \) can be obtained from the sensitive area \( A_{TM} \) of the MP512T as a function of \( D_{sd} \) as follows, cf. Equation 6.4.

\[
A_o = A_{TM} \left( \frac{L+1.5}{L-D_{sd}} \right)^2 \tag{6.4}
\]

Where, \( L \) is a source to surface distance, which for conventional linear accelerators is 100 cm.

### 6.5 Conclusions

The work described in his chapter demonstrates that the dose in a phantom can, in principle, be calculated at any depth of interest based on the transmission measurements made with the MP512T detector. The calculated dose for regular field sizes and intensity modulated fields for a clinical case have shown good agreement when compared with the dose predicted by the TPS and measured using EBT3 film. This study demonstrates the potential of our pixelated monolithic silicon detector as a transmission detector that can be used for small field QA, and that has the potential to be used for real-time 3D dose reconstruction as therapy is delivered.
CHAPTER 7

CONCLUSION AND FUTURE WORK

Radiation therapy has become one of the main treatment options for cancer over the last several decades with over 50% of patients having some form of radiation therapy as part of their cancer management plan. There is a contribution to modern radiotherapy techniques, such as IMRT, VMAT and SRS/SBRT. These sophisticated treatment techniques enable the high-energy X-ray beam to be adjusted, changing the beam shape and intensity to conform better to the shape of the tumours while sparing the radiation dose to the normal healthy tissue. The concentrated radiation dose techniques use many multileaf collimators that can move independently in the beam path in order to block the beam and generate the high precision in millimetre sized radiation dose voxels.

The movement of the leaves is controlled and optimized by the calculation algorithm from the treatment planning system. There are more parameters that the system uses for computing the expected dose distribution in the patient such as beam energy, delivery time, type of tissue in the beam line, etc. To check if all parameters for this complex plan are correct, sophisticated treatment planning verification is needed. The verification method ensures the treatment plan delivers the radiation dose so as to match well to the dose delivered to the patient.

Online treatment verification is needed during the treatment delivery. Using such a method, any errors can be detected in real-time, and the treatment can be stopped immediately and re-planned to accommodate for the error. The additional software and hardware that are required for this decision are not related to the work presented this thesis.
The CMRP at UOW has developed a 2D monolithic silicon diode detector array known as “Magic Plate512” for in-phantom dosimetry and transmission measurements with millimetre spatial resolution that is variable.

In this thesis, in-phantom dosimetry has been studied. The effect of the air gap size above the detector was investigated and its impact on the measured output factor, wedge beam profile and percentage depth dose for both photon and electron beams. In this study, the suitable air gap was optimized for each beam energy utilized.

The output factor measured with the MP512 reduced with the increase of the detector air gap at the smaller radiation field size. The MP512, with the air gap of 0.5 mm and 1.2 mm, show good agreement to the output factors measured with the EBT3 film and MOSkin within ±2% for 6 MV photon beam fields and 10 MV, respectively.

The optimal air gap size for 6 MV and 10 MV energy was confirmed through similar results for the expected PDD measurement. The PDD measured by MP512 with a 0.5 mm (for 6 MV) and 1.2 mm (for 10MV) air gap size was within ±3% of the EBT3 for the 2 x 2 cm² field for both photon energies. For the larger fields such as 5x5 cm², and 10x10 cm², the PDD measured with the MP512 is within ±1.6% and ±1.5% of that measured using a Markus IC for 6 and 10 MV fields respectively.

The thesis used the optimal air gap size of 0.5 mm for 6 MV and 1.2 mm for 10 MV for wedge beam profile measurements. As expected, the difference between the beams profile measured by the MP512 and EBT3 film increases with increasing the air gap size. The air gap causes a measurable dose reduction for small radiation fields due to the loss of electronic equilibrium.

The thesis also investigated the effect of the air gap size above the MP512 detector for various energies of electron beams. However, the effect on electron beam is not significant due to an electronic equilibrium being fully established and maintained.
The MP512, with a different detector packaging incorporated into the design was called MP512T. The detector is 0.45 mm thick and was then used as a transmission type detector. This thesis came up with, and demonstrated, the idea of the adjustable effective detector spatial resolution by placing the MP512T on a movable stand and moving the detector in the direction between the target in the patient and the Linac head.

By moving the detector between the patient and the Linac head it is possible to change the effective spatial resolution for dosimetry in a monitored radiation field. These detectors allow us to obtain a variable effective spatial resolution from 2 mm to 4 mm for the MP512T detector. For a small tumour size, placement of the detector closer to the patient will improve the spatial resolution in transmission mode, while monitoring of the whole radiation field is still possible.

In this thesis, the influence of the MP512 on the beam was studied. In particular any perturbation measured as a change in the surface dose and beam transmission factors were investigated when operating the MP512T in transmission mode and with the MP512T placed at various distances from the solid water phantom.

When placing the detector at the distance of above 18 cm above the phantom surface produced only a small perturbation of the surface dose, measured as an increase in surface dose by less than 5%. The increase in surface dose was negligible for radiation field sizes of 5 x 5 cm\(^2\). The difference in surface dose between MP512T faced up and faced down showed only a 2% difference. The effect of the PCB without the detector mounted on the surface also indicated that at the distance of more than 18 cm, the percentage difference is close to zero for all irradiated field sizes. The PCB alone increased the surface dose by about 60% compared to the increase in the surface dose from the MP512T. Due to the Compton Effect electrons are mostly created at the MeV range, and this suggests that an
opening or recess in the PCB under the active area of the silicon monolithic detector is recommended to reduce the increase in skin dose for all the considered IFSs.

The measurement of transmission coefficient of the MP512T detector at $d_{\text{max}}$ is close to 1 as the distance between the MP512T detector and the phantom surface decreased below 18 cm. The PCB under the silicon monolithic detector makes the transmission coefficient closer to 1 for any placement of the proposed transmission detectors between patient and the Linac head.

In this thesis, the use of the MP512T as a transmission-type detector was studied in more detail with an example of a real clinical scenario. The correlation of the transmission mode response (TM) and dose mode response (DM) of the MP512T for different detector to surface distances ($D_{sd}$) and treatment field sizes were used to predict the dose at $d_{\text{max}}$.

The optimal air gap size above the detector for 6MV photon beam (0.5 mm) was applied for the dose mode measurement of MP512 at $d_{\text{max}}$. For TM, the MP512T was positioned at various values of $D_{sd}$. The results showed that a correlation between TM and DM existed and could be employed to predict the dose at $d_{\text{max}}$.

When placing the MP512T at a $D_{sd}$ of 4 cm and 24 cm, the predicted dose for a regular field size of 1x1 cm$^2$ fell within 2.18% compared to the measured dose. The predicted dose when placing the MP512T at the same $D_{sd}$ for 4 x 4 cm$^2$ fell within 1.95% of the measured dose.

The performance of the MP512T when used to calculate the dose for the clinical IMRT plan was in good agreement with the TPS and the EBT3 film. The gamma criterion of 3%/3mm pass rate was 98.14% for $D_{sd}$ 4 cm and 90.5% for $D_{sd}$ 24 cm. The pass rate of 2%/2mm was 97.22% for $D_{sd}$ 4 cm and 93.8% for $D_{sd}$ 24 cm.

Future work will be to develop the real-time high spatial resolution 3D dose reconstruction algorithm based on the TM measurements using the MP512T. Moreover,
the mechanical realization of proposed transmission detector will be realized on a
telescopic jig attached to the Linac head block tray slot together with a wireless reader
developed at CMRP.
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Appendix A

Solid works drawing

The detector holder with specific geometries were designed for particular part of this thesis. A 3D drawing was generated by using Solid works (Dassault Systems SolidWorks Corporation, Massachusetts, USA)

A.1 The transmission Magic Plate 512 (MP512T) detector holder

The special holder shown in Figure A.1 was used for the work described in Chapter 5 and Chapter 6.

Figure A.1: The transmission Magic Plate (MP512T) detector holder
A.2 The movable stand detector holder

The movable sand detector holder shows in Figure A.2 was used in the work described in Chapter 5 and Chapter 6.

Figure A.2: The movable stand detector holder
Matlab script was designed by using Matlab (The MATHWORK Inc.) to manage the data involved in the thesis.

### B.1 EBT3 Film Center Location

This script was used to find the center of the EBT3 film.

```matlab
function [x0,y0] = center(A)
if size(A,3)== 3
    A = double(rgb2gray(A)); %Do image to gray scale
end
    [row,col] = size(A);  % Find the size of the image
% Determine location * image value
    Mx = ones(row,1)*(1:col).*A;
    My = (1:row)'*ones(1,col).*A;
% Determine the total value summation
    area = sum(A(:));
% check for zero values
    if area == 0 % central mass on uniform image
        x = row/2;
        y = col/2;
    else
% Calculate centroids
        x = sum(Mx(:))/area; % centroid location of x
        y = sum(My(:))/area; % centroid location of y
    end
if nargin==2 && pix
    x = round(x); %x is the center of coll
    y = round(y); %y is the center of raw
end
% Make in the integer number
    y0=round(x); %switch to X axis when x is a position of column
    x0=round(y); %switch to Y axis when y is a position of row
end
```
B.2 MP512 Pixel Interpolation

There was some dead pixel during the measurement by MP512. This script was used to interpolate between the functional detector pixel and the dead one.

```matlab
function [ newdata ] = InterpolateHelper( data )
warning('off','all')
newdata = zeros(24,24);
[n_rows, n_columns] = size(data);
newdata(1,:) = data(1,:);
newdata(:,1) = data(:,1);
ewdata(24,:) = data(24,:);
newdata(:,24) = data(:,24);

fprintf('Start row pass');
%row pass
for n=2:n_rows-1
  check = 1;
  while(check)
    fprintf('Start rows %d
', n);
    focus = data(n,:);
    line = [1:24];
    [fitresult, gof] = createFit(line, focus);
    prompt = 'How many pixels do you want to interpolate:';
    n_pixels = input(prompt);
    if n_pixels > 0
      fix_pixels = [];
      for n_pixel=1:n_pixels
        fprintf('What is the pixel No. %d that you want to fix',n_pixel);
        prompt = ':';
        fixing_pixel = input(prompt);
        fix_pixels(n_pixel) = fixing_pixel;
      end
      [fitResult, gof ] = createFit2(line, focus, fix_pixels);
      for n_pixel = 1:n_pixels
        pixel = fix_pixels(n_pixel);
        focus(fix_pixels) = fitResult(fix_pixels);
      end
    end
    createFit(line, focus);
    newdata(n,:) = focus;
    prompt = '0 = ok, 1 = redo';
    check = input(prompt);
  end
end

fprintf('Start column pass');
% column pass
```

150
for n=2:n_columns-1
    check = 1;
    while(check)
        fprintf('Start column %d\n', n);
        focus = newdata(:,n);
        line = [1:24];
        [fitresult, gof] = createFit(line, focus);
        prompt = 'How many pixels do you want to interpolate:';
        n_pixels = input(prompt);
        fix_pixels = [];
        if n_pixels > 0
            for n_pixel=1:n_pixels
                fprintf('What is the pixel No. %d that you want to fix', n_pixel);
                prompt = ':';
                fixing_pixel = input(prompt);
                fix_pixels(n_pixel) = fixing_pixel;
            end
            [fitResult, gof] = createFit2(line, focus, fix_pixels);
            for n_pixel = 1:n_pixels
                pixel = fix_pixels(n_pixel);
                focus(fix_pixels) = fitResult(fix_pixels);
            end
        end
        createFit(line, focus);
        newdata(:,n) = focus;
        prompt = '0 = ok, 1 = redo';
        check = input(prompt);
        close all;
    end
end

B.3 Gamma Evaluation

This script is used for calculated the gamma evaluation described in chapter 6.

function [GammaMap numpass avg numWithinField] = GammaCompare(Image1, Image2, x_size, y_size, EPIDppx, Dose_tol, DTA_tol, FE_thresh, rad, varargin)
% Compare Image2 to reference image (TPS) Image1
% rad: radius - in points - need to work out before pass
% GAMMA settings
% x_size = image x size
% y_size = image y size
% Dose_tol = DoseTol; % Percent of maximum dose. Given as a fraction
% DTA_tol = DistTol; % cm
% FE_thresh = ThreshFx; % Given as a fraction
% rad = radius; distance in cm to search
% Note: DTA tolerance and specified X, Y points must be in the same units
% example
% [GammaMap numpass avg numWithinField] = GammaCompare(img1, img2, 1024, 768, 0.0392, 0.03, 0.3, 0.1, 5);

debuglevel = 0;
% EPID pixel size spacing (EPIDppx) = 0.0392 (1024X768); =0.0784 (512X384)
% - AS1000 EPID
xp = (1:x_size)*EPIDppx;
yp = (1:y_size)*EPIDppx;
res_x = xp(2) - xp(1);
res_y = yp(2) - yp(1);

% if radius not specified, compute a sensible one
% Expand DTA tolerance by 50%. Use this as a search radius
if ~exist('rad','var') || isempty(rad)
    radlim = DTA_tol * 1.5;
    rad = min(ceil(radlim/res_x),ceil(radlim/res_y));
end

MaxVal = max(Image1(:));
Mask = zeros(size(Image1));
crit_val = FE_thresh*MaxVal;

% Use the Resampled image for this - EPID no spikes?
Mask(Image1>crit_val) = 1;

Dose_tol = Dose_tol*MaxVal;  % GLOBAL - PERCENTAGE OF MAX DOSE
if debuglevel > 1
    fprintf('Maximum dose for Gamma: %2.1f cGy\n',MaxVal);
    fprintf('Dose Tolerance for Gamma: %2.1f cGy\n',Dose_tol);
    fprintf('DTA: %2.1f\n', DTA_tol);
    fprintf('FE_thresh: %2.1f\n', FE_thresh);
    fprintf('rad: %2.1f\n', rad);
    fprintf('Image 1 val: %2.1f\n',Image1(round(size(Image1,1)/2),
        round(size(Image1,2)/2)));
    fprintf('\n');
end

% VECTORIZED CALCULATION STARTS HERE

% GammaMapsub will carry the calculated gamma values for the truncated % images. GammaMap2 will be the Gamma values for the full image.
GammaMapsub = NaN;
GammaMap = zeros(size(Image1));
% Find the threshold limits for truncation
[validmask_y validmask_x] = find(Mask);
min_x = min(validmask_x)-rad;
max_x = max(validmask_x)+rad;
min_y = min(validmask_y)-rad;
max_y = max(validmask_y)+rad;
if min_x < 1
    min_x = 1;
end
if min_y < 1
    min_y = 1;
end
if max_x > size(Image1,2)
    max_x = size(Image1,2);
end
if max_y > size(Image1,1)
    max_y = size(Image1,1);
end

% Truncate the images to avoid needless calculations
Im1 = Image1(min_y:max_y,min_x:max_x);
Im2 = Image2(min_y:max_y,min_x:max_x);
% Shift the image by varying amounts. Determine the minimum gamma value
% for all shifts
for i=-rad:rad
    for j=-rad:rad
        % circshift function wraps elements from top to bottom as necessary
        % The entire image is shifted at once
        Im2_shift = circshift(Im2,[i j]);
        dist = sqrt((res_y*i)^2 + (res_x*j)^2);
        DoseDiff = Im2_shift - Im1;
        % Compute the gamma map for this particular shift value
        Gamma_temp = sqrt((dist./DTA_tol).^2 +
                           (DoseDiff./Dose_tol).^2);
        % Accumulate the map of the minimum values of gamma at each point
        GammaMapsub = min(GammaMapsub,Gamma_temp);
    end
end
% Put the truncated gamma map back into its proper location within the full gamma map
GammaMap(min_y:max_y,min_x:max_x) = GammaMapsub;
% Remove any edge effects from the circular shifting by multiplying by the
% mask values. This will negate any calculated gamma values around the
% edges of the distribution where this effect would arise
GammaMap = GammaMap .* Mask;
% Ensure that NaN values outside the mask do not affect the calculation
GammaMap(~Mask) = 0.0;

% Compute statistics
numWithinField = nnz(Mask);
numpass = nnz(GammaMap<1 & Mask)./numWithinField;
avg = sum(GammaMap(:))./numWithinField;