High resolution radiation dosimetry in an MRI-linac using novel silicon array detectors

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High Resolution Radiation Dosimetry in an MRI-Linac Using Novel Silicon Array Detectors

A thesis submitted in fulfilment of the requirements for the award of the degree

Doctor of Philosophy

from

University of Wollongong

by

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2019
Abstract

The integration of a linear accelerator with magnetic resonance imaging, or MRI-linac, provides a means to acquire gated images of the tumour and surrounding organs. This may enable more accurate radiation therapy treatment with the potential for motion-tracking using these images. Challenges of this kind of treatment include effects due to the magnetic and radiofrequency (RF) fields on the dose distribution. Magnetic fields influence charged particles via the Lorentz force. The ‘Electron Return Effect’ (ERE) and asymmetry of the point spread kernel are dosimetric effects due to a magnetic field that is transversely orientated with respect to the radiation beam. Narrower penumbral widths result from an inline magnetic field, as well as an increase in skin dose. Furthermore, for validation of this dose delivery, radiation detectors that can operate under magnetic and RF fields are needed. The effect that these fields have on the detector response, as well as the deposited dose, is investigated.

A QA concept named ‘MR dynamic dosimaging’ is introduced with the potential to combine the MagicPlate-512 (M512) silicon array detector with MRI-compatible phantoms on a moving platform. MR visible/compatible detector systems are required for this to be achieved. Preliminary studies towards dosimaging were completed separately on a clinical MRI scanner and standard linac. A type of imaging hydrogel, named ‘gel-water’ was investigated to enable the M512 to be MR visible, and at the same time replace solid water/water as a dosimetry phantom. It was concluded that gel-water is
a feasible option to achieve these aims in an MRI-linac. The system with gel-water and the M512 had an agreement with solid water of $\leq 1.3\%$ for various dosimetric characterisation measurements.

Quantification of the narrowing of beam penumbra due to an inline magnetic field is needed, especially in air cavities, for a progression towards MRI-linac clinical treatments. Profiles on Gafchromic EBT3 film were used to investigate differences in penumbra between solid water and solid lung phantoms, and under 0, 0.9, and 1.5 T conditions using the Australian MRI-linac prototypes. The penumbral width narrowing was greatest at 1.5 T (up to 4.4 ± 0.1 mm), compared to the 0.9 T field (up to 2.5 ± 0.1 mm).

It is important that the effect of magnetic fields on detectors and the dose distribution is quantified. A permanent magnet capable of a magnetic field strength of up to 1.2 T provided a preliminary testing environment prior to investigation of in-house developed silicon array detectors on the MRI-linac. Both transverse and inline orientations of the magnetic field were possible with this permanent magnet and solid water and solid lung phantoms were utilised. A DUO silicon array detector was used to measure beam profiles. The results were compared to film to highlight any intrinsic magnetic field effects in the silicon. The largest changes in penumbral width were observed in the lung phantom. In lung, the greatest penumbral narrowing due to the inline field was 0.7 mm, and the greatest increase of penumbral width due to the transverse field was 2.3 mm. The largest asymmetry due to the transverse magnetic field was 4.6%.

Dosimetric profiles were then successfully measured using the M512 in a 1.0 T inline magnetic field in comparison to both zero field and to profiles taken during an imaging sequence on the Australian MRI-Linac. Good agreement was observed between penumbral widths measured by both the M512 and
film, with differences of up to a maximum of 1.8 mm, and with an average of <1 mm. Concurrent imaging widened the penumbra by up to 1.2 mm due to RF noise affecting the detector, but this was not experienced by the profiles measured with film. Hence, more effective noise reduction will be required for the achievement of dynamic dosimaging. The M512 silicon array detector was used for collecting high resolution dose profiles on the MRI-linac, which represents the first successful introduction of this unique, high resolution device on the first MRI-linac in Australia.
Statement of Originality

This is to certify that the work described in this thesis is entirely my own, except where due reference is made in the text.

No work in this thesis has been submitted for a degree to any other university or institution.

Signed

Sarah Jane Alnaghy
First of February, 2019
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<th>Description</th>
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<tr>
<td>3DCRT</td>
<td>Three-Dimensional Conformal Radiation Therapy</td>
</tr>
<tr>
<td>ABS</td>
<td>Acrylonitrile Butadiene Styrene</td>
</tr>
<tr>
<td>ADC</td>
<td>Analogue-to-Digital Converter</td>
</tr>
<tr>
<td>AFE</td>
<td>Analogue Front End</td>
</tr>
<tr>
<td>Al</td>
<td>Aluminium</td>
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<tr>
<td>CBCT</td>
<td>Cone-Beam Computed Tomography</td>
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<tr>
<td>CEN</td>
<td>Central Electrode Negative</td>
</tr>
<tr>
<td>CEP</td>
<td>Central Electrode Positive</td>
</tr>
<tr>
<td>CMRP</td>
<td>Centre for Medical Radiation Physics</td>
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<tr>
<td>60Co</td>
<td>Cobalt-60</td>
</tr>
<tr>
<td>CPE</td>
<td>Charged Particle Equilibrium</td>
</tr>
<tr>
<td>CT</td>
<td>Computed-Tomography</td>
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<tr>
<td>CTV</td>
<td>Clinical Target Volume</td>
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<tr>
<td>DAQ</td>
<td>Data Acquisition</td>
</tr>
<tr>
<td>DCE-MRI</td>
<td>Dynamic Enhanced Magnetic Resonance Imaging</td>
</tr>
<tr>
<td>DHI</td>
<td>Dose Heterogeneity Index</td>
</tr>
<tr>
<td>DMG</td>
<td>Dose Magnifying Glass</td>
</tr>
<tr>
<td>DVH</td>
<td>Dose Volume Histograms</td>
</tr>
<tr>
<td>DWI</td>
<td>Diffusion Weighted Magnetic Resonance Imaging</td>
</tr>
<tr>
<td>EBRT</td>
<td>External Beam Radiation Therapy</td>
</tr>
<tr>
<td>ERE</td>
<td>Electron Return Effect</td>
</tr>
<tr>
<td>FEG</td>
<td>Frequency Encode Gradient</td>
</tr>
<tr>
<td>FFF</td>
<td>Flattening Filter Free</td>
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<tr>
<td>Abbreviation</td>
<td>Description</td>
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<tr>
<td>FOD</td>
<td>Fibre optic dosimeter</td>
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<tr>
<td>FPGA</td>
<td>Field-Programmable Gate Array</td>
</tr>
<tr>
<td>FSE</td>
<td>Fast Spin Echo</td>
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<tr>
<td>FWHM</td>
<td>Full Width at Half Maximum</td>
</tr>
<tr>
<td>GTV</td>
<td>Gross tumour volume</td>
</tr>
<tr>
<td>ICCC</td>
<td>Illawarra Cancer Care Centre</td>
</tr>
<tr>
<td>IMRT</td>
<td>Intensity-Modulated Radiation Therapy</td>
</tr>
<tr>
<td>LCTC</td>
<td>Liverpool Cancer Therapy Centre</td>
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<tr>
<td>LET</td>
<td>Linear Energy Transfer</td>
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<tr>
<td>Linac</td>
<td>Linear Accelerator</td>
</tr>
<tr>
<td>MARDOS</td>
<td>Magnetic Apparatus for RaDiation Oncology Studies</td>
</tr>
<tr>
<td>M512</td>
<td>MagicPlate-512</td>
</tr>
<tr>
<td>MLC</td>
<td>Multileaf Collimator</td>
</tr>
<tr>
<td>MOSFET</td>
<td>Metal Oxide Semiconductor Field-Effect Transistor</td>
</tr>
<tr>
<td>MP121</td>
<td>MagicPlate-121</td>
</tr>
<tr>
<td>MRI</td>
<td>Magnetic Resonance Imaging</td>
</tr>
<tr>
<td>MRigRT</td>
<td>MRI-Guided Radiation Therapy</td>
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<tr>
<td>MU</td>
<td>Monitor Unit</td>
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<tr>
<td>MV</td>
<td>Megavoltage</td>
</tr>
<tr>
<td>PCB</td>
<td>Printed Circuit Board</td>
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<tr>
<td>PEG</td>
<td>Phase Encode Gradient</td>
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<tr>
<td>PMMA</td>
<td>Polymethyl Methacrylate</td>
</tr>
<tr>
<td>PTV</td>
<td>Planning Target Volume</td>
</tr>
<tr>
<td>QA</td>
<td>Quality Assurance</td>
</tr>
<tr>
<td>RF</td>
<td>Radiofrequency</td>
</tr>
<tr>
<td>SAD</td>
<td>Source-to-Axis Distance</td>
</tr>
<tr>
<td>SBRT</td>
<td>Stereotactic Body Radiation Therapy</td>
</tr>
<tr>
<td>sDMG</td>
<td>Serial DMG</td>
</tr>
<tr>
<td>Si</td>
<td>Silicon</td>
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<tr>
<td>SiO2</td>
<td>Silicon Dioxide</td>
</tr>
<tr>
<td>Abbreviation</td>
<td>Definition</td>
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<td>------------------------------------</td>
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<tr>
<td>SNR</td>
<td>Signal-to-Noise Ratio</td>
</tr>
<tr>
<td>SRS</td>
<td>Stereotactic Radiosurgery</td>
</tr>
<tr>
<td>SSD</td>
<td>Source-to-Surface Distance</td>
</tr>
<tr>
<td>TLD</td>
<td>Thermoluminescent Dosimeter</td>
</tr>
<tr>
<td>TMR</td>
<td>Tissue Maximum Ratio</td>
</tr>
<tr>
<td>TPR</td>
<td>Tissue Phantom Ratio</td>
</tr>
<tr>
<td>TPS</td>
<td>Treatment Planning System</td>
</tr>
<tr>
<td>TrueFISP</td>
<td>True Fast Imaging with Steady State Precession</td>
</tr>
<tr>
<td>VMAT</td>
<td>Volumetric Modulated Arc Therapy</td>
</tr>
<tr>
<td>WED</td>
<td>Water Equivalent Depth</td>
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Conference Proceedings

SJ Alnaghy, T Causer, M Gargett, BM Oborn, M Petasecca, N Roberts, L Holloway, AB Rosenfeld, P Metcalfe, ‘DUO silicon array detector profiles in transverse and inline static magnetic fields’, oral presentation, Aus MRRT, Wollongong, 2018

SJ Alnaghy, M Gargett, T Causer, B. Oborn, M Petasecca, J Begg, G Liney, L Holloway, AB Rosenfeld, P Metcalfe, ‘Small field dose profiles measured in a transverse, static magnetic field with Duo silicon detector’, poster presentation, MR in RT, Ingham Institute for Medical Research, Liverpool, 2017

SJ Alnaghy, J Begg, T Causer, T Alharthi, L Glaubes, A George, L Holloway, AB Rosenfeld, P Metcalfe, ‘Comparison of solid water and solid lung Gafchomic® film profiles in 0 T and 1.5 T longitudinal magnetic field of the Australian MRI-Linac’, poster presentation, EPSM, Sydney, 2016


S Alnaghy, M Gargett, G Liney, M Petasecca, J Begg, A Espinoza, M Newall, L Holloway, M Lerch, AB Rosenfeld, P Metcalfe, ‘MRI-linac image-guided radiation dosimetry: Initial experiments with the MagicPlate-512
silicon array detector’, oral presentation, MRI-Linac Collaborators Meeting, University of Sydney, Sydney, 2015


**S Alnaghy**, ‘Characterisation of linatron in a magnetic field using silicon array dosimeters’, oral presentation, MRI-Linac Collaborators Meeting, University of Sydney, Sydney, 2014
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Awards

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Chapter 1

Introduction

Several tumour sites surrounded by soft tissue are difficult to target with radiation therapy due to current sub-optimal x-ray-based image-guidance systems such as cone beam computed tomography (CBCT). The sites of most clinical interest as identified by the MRI-Linear Accelerator Consortium (2016) were brain, breast, cervix, oesophagus, lung, oropharynx, pancreas, prostate, and rectum [8]. Magnetic Resonance Imaging (MRI) has a superior image quality compared to CT due to multiple sequences providing breadth of image acquisition, allowing for an improvement in the visualisation of soft tissue treatment sites in radiation therapy [9–13]. Hence, combining MRI with a linear accelerator (MRI-linac), potentially provides a more accurate targeting for dose escalation to the tumour, and therefore the potential for improvement of patient outcomes. MRI-linacs enable adaptive radiation therapy (ART) due to the ability to visualise anatomical changes over the treatment course [14–16]. In addition, MRI-linacs have the capacity for motion-tracking, which is expected to improve treatment by reducing the volume of treatment by tracking or gating, of sites that have proven more difficult to treat, such as lung [17]. MRI has the added advantage over x-ray-based image-guidance of not providing radiation dose from imaging.

Magnetic field effects on dose distribution and radiation detector func-
tion must be well understood. The radiofrequency (RF) field effects on
detector function must also be understood. To validate MRI-linac treatments,
detectors for use in vivo and for pretreatment quality assurance (QA) are
required; hence, it is essential that detectors can operate under magnetic
and RF fields, and any changes in the detector or dose deposition should be
quantifiable so that a comparison can be made to measurements taken at zero
magnetic field. This is the focus of this thesis.

The detectors utilised in this study, the MagicPlate-512 and the DUO,
are silicon array detectors developed in-house at the Centre for Medical
Radiation Physics (CMRP), Wollongong, Australia. As a pre-treatment dose
verification tool, these detectors are capable of verifying the dose distribution
to be delivered to the patient. Preliminary studies were completed separately
on a clinical MRI scanner and standard linac. Further studies continued to a
permanent magnet system in conjunction with a standard linac, before pro-
ceeding to the Australian MRI-linac (Ingham Institute for Medical Research,
Liverpool, Australia). An investigation with Gafchroomic® EBT3 film of the
penumbral narrowing due to an inline magnetic field was also completed on
the MRI-linac to compare this effect in different materials. MR images of
stationary detectors during dose acquisition were also taken. Overall, this
project is a precursor to potential future studies of a QA tool named ‘dynamic
dosimaging’, whereby a moving detector can be tracked utilising the MR
imaging of the MRI-linac to verify the dose over time as well as its position.

1.1 Objectives and Overview of this Thesis

The fundamental objectives of this Thesis were as follows:

1. To determine appropriate electronic shielding of silicon array detectors for
use in the RF fields of an MRI-linac, so that the detectors are not affected
by the RF and so that the MR images are not affected by noise from the detector.

2. To characterise gel-water phantom material for use as part of MRI-linac dosimetry phantom and dosimeter for ‘static dosimaging’, which combines detector-tracking in real-time with MR imaging for concurrent position and dose verification. For eventual extension to ‘dynamic dosimaging’.

3. To investigate photon field penumbral narrowing due to a magnetic field in alignment with the radiation beam of two different inline MRI-linac prototypes (1.5 T closed bore and 1.0 T open bore). EBT3 film will be utilised to quantify the effect in solid lung in comparison to solid water.

4. To investigate magnetic field effects on and with the silicon array detector, DUO, in the transverse and inline, static magnetic fields (1.2 T) of a permanent magnet system on a standard linac.

5. To investigate magnetic field effects on and with a novel silicon array detector in the inline magnetic field of an MRI-linac (1.0 T), without MR imaging and during image acquisition.
Chapter 2

Literature Review

2.1 External Beam Radiation Therapy

External Beam Radiation Therapy (EBRT) is a non-invasive treatment, predominantly for cancer, that uses high energy x-rays (usually 6-18 MV) to deliver a radiation dose to the patient. The most common mode of delivery is via a linear accelerator (linac). Three-Dimensional Conformal Radiation Therapy (3DCRT) and Intensity-Modulated Radiation Therapy (IMRT) are techniques usually employed to administer this treatment. Other techniques include Volumetric Modulated Arc Therapy (VMAT), where the radiation dose is continuously delivered with a rotating linac gantry, and TomoTherapy® where the treatment is delivered helically via the rotating narrow fan beam of a TomoTherapy machine. Radiation dose is administered in a fractionated regimen to allow reoxygenation of the tumour and to allow healthy tissue to repair. Stereotactic Body Radiation Therapy (SBRT) and Stereotactic Radiosurgery (SRS) are hypofractionated regimens designed to deliver a large radiation dose in a small number of fractions. The aim of EBRT is to treat the tumour, whilst minimising the irradiation of healthy tissue for the reduction of side-effects [18]. Advantages of EBRT include that patients can carry out their daily activities around treatments and that it is mostly an out-patient procedure [19]. There are several disadvantages of EBRT, which include the
length of treatments (i.e. daily treatments over many weeks), damage to surrounding organs and the risk of a secondary malignancy [20].

3DCRT uses multileaf collimators (MLCs) to conform the treatment field to the target. 3D treatment planning is utilised. IMRT utilises many beam segments to modulate the intensity of the beam within each treatment field. Inverse treatment planning achieves optimisation from user-provided dose objectives and then the calculation of a dose distribution from these [21]. Static and dynamic options for MLCs are available in IMRT. Dynamic IMRT has been found to require 15-20% more monitor units (MUs) and 15% more time than static IMRT, increasing the risk of secondary malignancies from radiation exposure [22–24]. However, dynamic IMRT with its continuous MLC motion allows the intensity that is delivered to resemble the optimal fluence of the plan more closely [22]. IMRT has been shown to reduce the dose to critical structures in comparison to 3DCRT, whilst being able to escalate the target dose [25–27]. Advantages of 3DCRT include that this technique requires less MUs than static IMRT (up to 4.9% less depending on the energy), and hence less than dynamic IMRT [22,28].

SBRT delivers hypofractionated radiation doses to the patient, and in the case of SRS, only one fraction is utilised to deliver the entire prescribed dose. Small field sizes of 4 cm or less are treated, with large doses prescribed per fraction [29]. Previously, rigid frames have been utilised to immobilise the patient more precisely to deliver these large doses to small fields. Newer technologies that deliver SBRT and SRS include linacs such as TrueBeam, or other technologies such as TomoTherapy and Cyberknife [30].

### 2.1.1 Image-Guidance and Tracking

An important limitation of EBRT to consider is the over- or under-dosing of the treatment target that may occur due to patient or organ motion, either between treatment fractions (interfraction) or during fractions (intrafraction).
As radiation therapy treatment margins decrease and target doses increase, verification in radiation therapy has become increasingly important. Image-guidance is integral to current radiation therapy practice, with present forms of x-ray-based image-guidance including kilovoltage (kV) cone beam computed tomography (CBCT) or megavoltage (MV) portal imaging. These are aligned to bony anatomy, soft tissue or fiducial markers for position verification. Tracking with dynamic multi-leaf collimators (DMLCs) or gated treatments are the methods currently used to account for intrafraction motion. The kV intrafraction monitoring (KIM) study utilises fiducial markers implanted into a patient to monitor movement for gating using kV imaging [31]. Other markers utilised for this purpose are optical surface markers which can be integrated with CBCT to correlate internal and external patient motion [32]. Markerless tumour tracking by matching kV projection images with digitally reconstructed radiographs (DRRs) is another method that has been investigated to monitor the motion of tumours [33]. A DMLC motion-tracking study utilising the MagicPlate-512 (M512) silicon array detector with Calypso Radiofrequency (RF) tracking device has already been published [34], while a concept similar to this, with DMLC motion-tracking of silicon array detectors is possible for MRI-linacs. The MRI-linac concept will be discussed in the next section.

### 2.1.2 Magnetic Resonance Imaging - Linear Accelerator Concept

A combination of MRI with a linac, or ‘MRI-linac’, allows for real-time imaging of treatment sites so that these sites can be tracked during radiation therapy. The superior image quality of soft tissue with MRI is expected to improve the delineation of tumours, and thus reduce the irradiation of healthy tissue due to tighter treatment margins. Another benefit of MRI-linacs is the absence of extra radiation dose due to imaging. The enhanced soft tissue visualisation and motion-tracking capability is expected to improve sites that have proven difficult to treat, such as lungs, kidneys, and the pancreas [17].
In an MRI-linac, the magnetic field lines can be either inline with (longitudinal orientation) or perpendicular to (transverse orientation) the radiation field (Figure 2.1). Some systems have the capacity for both orientations. Several MRI-guided radiation therapy (MRIgRT) facilities have been developed in different parts of the world. These MRIgRT systems were developed at the University Medical Centre in Utrecht, The Netherlands [35], the Cross Cancer Institute at the University of Alberta in Edmonton, Canada [36], ViewRay at Washington School of Medicine in St Louis, USA [37], and the Princess Margaret Cancer Centre in Toronto, Canada [38]. The facility where this study took place was the Ingham Institute for Medical Research at Liverpool Hospital in Liverpool, Australia [39]. Each facility has a variation in the alignment of the radiation beam and magnetic field, MRI bore size, magnetic field strength, and energy of the radiation beam [35–40].

The facility in Utrecht originally developed a system with a modified, closed-bore, 1.5 T Achieva Philips MRI (Philips, Amsterdam, The Netherlands) and an Elekta 6 MV linac (Elekta AB, Stockholm, Sweden). This system has an accelerator mounted on a ring so that it can rotate in the midtransversal plane.
of the scanner [35]. It has now developed into a clinical machine by Elekta AB called Unity that has a 7 MV standing wave linear accelerator [41]. This system now has CE mark for clearance in Europe [42] and has started treating patients with 32 machines currently sold [41,43,44].

The Cross Cancer Institute has a system capable of both in-line and perpendicular orientations of the radiation beam with respect to the magnetic field. Their phase II system is a 6 MV linac combined with a 0.6 T superconducting, whole-body, open-bore magnet and their phase III system is a clinical system [36].

ViewRay had the first clinical system available, called the ViewRay System (ViewRay, Ohio, USA), which originally utilised three Cobalt-60 ($^{60}$Co) sources and an open, split-solenoid, 0.35 T scanner [37, 45]. The $^{60}$Co sources have since been replaced with a linac. The ViewRay system is now called the MRIdian and is both FDA cleared and CE marked [46]. The first patient was treated in 2014 and is currently operating at 18 facilities worldwide [45,47].

The MRIgRT system at Princess Margaret Cancer Centre is a different set-up entirely, with the 1.5 T, open-bore MRI being mounted on a rail to allow it to be manoeuvred into three different suites including an MRI simulation suite, a high dose rate (HDR) brachytherapy suite, and external beam (6 MV linac) suite [38,48].

The radiation component of the Australian MRI-linac was provided by a Varian industrial linac called a Linatron MP (Varian Medical Systems, Inc., CA, USA), which delivers a horizontal beam. It has dual-energy capability providing nominal flattening filter free (FFF) energies of 4 and 6 MV, with peak energies of 3.3 and 4.3 MV respectively. Its respective dose rates are 400 and 800 cGy/min. The frequency, or trigger rate, is capable of ranging
between 50 and 400 Hz. The linatron does not have a flattening filter. FFF clinical linacs have produced dosimetrically equivalent plans in preclinical studies [49,50]. An increase in dose rate and thus, a reduction in dose delivery time by more than 50% has also been observed [49, 51, 52]. The beam of the linatron is collimated by a circular aperture collimator and Varian Millennium 120 multileaf collimator (MLC). The linatron has a variable source-to-surface distance (SSD) achievable by a rail system to allow its position relative to the magnetic field to be changed. The first prototype of the Australian MRI-linac consisted of a 1.5 T Siemens Magnetom Sonata Whole-body MRI System (Siemens, Munich, Germany), previously a clinical system, which was coupled with the linatron in the inline orientation. The second prototype replaced the Sonata magnet with a 1.0 T Agilent split-bore magnet (Agilent Technologies, CA, USA), also with the magnetic field in-line to the radiation beam. The Australian MRI-linac will be further described in Section 4.2).

The physics behind the two major components that comprise an MRI-linac will be discussed in Section 2.2 and 2.3.

### 2.2 Linear Accelerator Physics

The delivery of high energy x-rays from a linear accelerator is achieved by bombarding a target of high atomic (Z) number with electrons. In the electron gun of the linac, the electrons required to produce the photon beam are generated from the cathode by thermionic emission, then accelerated to the anode, during which time they are focused by a focusing electrode. A gun grid inserted in front of the cathode can also be included to control the current of the electron gun by altering the voltage to the grid with respect to the cathode. The electron gun injects electrons into an accelerating waveguide. The electrons are accelerated in the waveguide by microwaves, which are generated by a klystron or magnetron. Operation of the klystron or magnetron requires
high-voltage pulses from a modulator circuit. A transmission waveguide allows the microwaves to pass to the accelerating waveguide. The two waveguides are separated by a beryllium or ceramic window as the transmission waveguide is filled with pressurised gas, whereas the accelerating waveguide is a vacuum. The electrons are pulsed in synchronisation with the microwaves into the waveguide [5,53].

Accelerating waveguides contain cavities since electrons and microwaves travel at different velocities, and a hollow tube would allow the microwaves to travel faster than the electrons. The cavities first allow the electrons to be ‘bunched’ on the electric field wave, and then afterwards to gain energy and to be accelerated. In most linacs (i.e. not those that have vertical gantries), a bending magnet assembly is employed to deflect the beam [5,53].

When the electrons strike the target, they interact with the Coulomb field of the atomic nuclei in the target material and lose energy by emission of a photon. This process is called bremsstrahlung and results in continuous x-rays. The target is made of tungsten, copper or a copper-tungsten laminate, which have a relatively high Z number for a high bremsstrahlung output. Bremsstrahlung output also depends on target thickness and kinetic energy of the incident electrons. After the target, due to the non-uniform shape of the x-ray beam; a flattening filter is used to produce a uniform shape (i.e. to ‘flatten’ the beam). Primary collimators (usually tungsten) are mounted above this flattening filter but below the target. For an electron beam, scattering foil replaces the target and flattening filter; its purpose being to widen the beam [5,53]. The secondary collimators, usually made of tungsten or a lead alloy are positioned below. Finally, the beam can be formed into irregular shapes by the MLCs [5].

The linatron of the Australian MRI-Linac Program differs in design from
these standard clinical linacs. In industry, the linatron is typically utilised for applications such as cargo-inspection or food and medical equipment sterilisation [54,55].

### 2.3 Magnetic Resonance Physics

The physics of MRI is based on the alignment of the magnetic moments of protons within a patient’s body with a static magnetic field. The protons resonate with the Larmor frequency, which is dependent of the strength of the magnetic field. The Larmor frequency for a proton in a 1 T field is approximately 42.6 MHz. For image formation in MRI, these protons precessing with a certain frequency are excited by a matching radiofrequency (RF) field; whereby they move to a higher (or lower) energy state (i.e. opposite spin). When the RF pulse is removed, the magnetic moments realign with the static magnetic field. The resulting RF signal created by the decay is detected for image formation [56].

The RF coils, which are electromagnetic coils, transmit and receive the RF pulses. Proximity RF coils are specific to a certain region of the body, such as the head or extremities, while surface coils improve resolution in smaller areas. The magnetic field is provided by either permanent magnets or superconducting magnets [56]. Typical magnetic field strengths for MRI scanners are 1.5-3 T, with higher field strengths being investigated; currently, the highest magnetic field for small animal scanners is 21.1 T [57]. The higher the magnetic field strength, the greater the signal-to-noise (SNR) ratio. The receive coil arrays specific to the body part to be imaged include 32-64 receive channels connected to each coil [17].

Magnetic gradient fields are generated in coil wires and are utilised to localise an object spatially in 3D by linearly varying in strength in proportion
to the distance. These gradient fields run in the x-, y-, and z-direction, and are added to the main magnetic field, where they are altered by varying the coil current. The steepness of these gradient fields is determined by their peak amplitude, and the slew rate is the time required to reach this peak amplitude. The slice thickness is determined by bandwidth and gradient steepness. For positional information along the Frequency Encode Gradient (FEG), the proton spins are frequency encoded by position. A short duration gradient can introduce phase to determine position along the Phase Encode Gradient (PEG). MRI data are stored in the frequency domain, or k-space, before the Fourier Transform of the k-space is plotted as an image [56].

Relating MR imaging back to radiation therapy; high resolution and contrast between the tumour and normal tissue is ideal due to the requirement for accurate delineation of the tumour in treatment planning. Different tissues have different relaxation rates; hence alteration of the pulse sequence parameters can generate contrast; the main types being T1, T2 and T2* relaxation, and water diffusion [17, 58, 59]. T1 relaxation is where the energy from the RF pulse is transferred back into the surroundings during realignment with the static magnetic field (longitudinal plane). T2 relaxation refers to the loss of phase coherence in the transverse plane, which may be increased due to field inhomogeneities (T2* relaxation) [60].

MRI, while beneficial for radiotherapy when not present during dose delivery, causes problems for dosimetry when it is present during dose delivery due to the magnetic and RF fields, as discussed in Section 2.5 and 2.7. Conversely, dosimetry can have an effect on MR images due to noise from detector electronics.
2.4 Dosimetry Needs for EBRT

Dosimetry is an integral part of safe EBRT delivery. It is used for measuring the linac output for commissioning and monitoring in daily quality assurance, as well as verification of the dose delivered to the patient, either before a treatment begins or in vivo.

Commissioning of a clinical linac is a procedure in which the mechanical and dosimetric characteristics of the linac are tested, and the data collected is utilised in the treatment planning system (TPS). During the commissioning process, a full set of beam data is acquired. This differs from acceptance testing, in which only a small subset of beam data for verification is collected [61, 62]. Commissioning can reveal discrepancies between vendor data (referred to as ‘golden beam data’) and data measured at the institution itself. Golden beam data should be verified and tested, as it may not be acceptable for clinical use [63].

During commissioning, it is essential to understand the limitations of the linac, detectors and other equipment, and to understand the measuring conditions. Some examples of considerations when measuring, as highlighted by TG-106, include that correct phantom and phantom size are used, gantry tilt is avoided, and the detector type, shape, orientation, size and polarity can affect results [63].

Commissioning of the linatron for the Australian MRI-linac is required, both with and without the presence of a magnetic field, due to variations that may arise from standard linac data from the magnetic field. The linatron has a different spectrum to standard clinical systems. Other challenges may arise due to the non-standard nature of this commissioning work. Green, Goddu and Mutic (2012) discuss the commissioning process of the ViewRay system. The
authors highlighted that it was not only the presence of the magnetic field that provided a more challenging commissioning process but that many available QA devices and techniques were not compatible with the system’s geometry [64].

Procedures for dosimetry, treatment planning and treatment delivery should be developed in a department, the accuracy of these verified, and then quality control procedures established [65]. Although the output of the machine is measured and verified during commissioning, continual quality assurance (QA) is required. At the level of the patient, dosimetry is also very important. In vivo dosimetry is a method of monitoring the actual dose delivered to the patient rather than the planned dose, and can be used as another safety precaution in treatment delivery. A dosimeter is placed on or inside the patient where the dose measurement is needed [66]. According to the ICRU report 24 (1976), an ultimate check of the actual delivered dose can only be performed through in vivo dosimetry [67–69].

Dosimetry systems are required which can accurately perform in both magnetic field and RF fields. Hence, dosimeters are required that contain non-ferromagnetic components and that can be adequately shielded from RF fields. The effect on dosimeters under a magnetic field should be quantified. This is further discussed in Section 2.5.

2.4.1 Small Field Dosimetry

Small fields are required to be considered differently than larger fields due to their altered dosimetric characteristics. There are two main effects that need to be considered when using small fields. The first is source occlusion, whereby the collimator blocks part of the direct beam source. The second is charged particle disequilibrium, which leads to a decrease in central axis dose due to the field width being smaller than the maximum range of lateral electron scatter. The minimum radius is 1.3 cm for lateral equilibrium in a
Due to the uncertainty surrounding small field dosimetry, it has been recommended to have more than one person independently verify the measurements, to use more than one detector, and to apply correction factors [63,70].

It is important to consider which detector is appropriate for small field dosimetry. Larger detectors are subject to volume averaging and an inability to resolve penumbra. Smaller detectors should, therefore, be used, ideally with an energy dependence close to water, as well as with good dose linearity and a minimum dose rate dependence [63]. Differences in small from large fields are particularly noticeable in output factor measurements; Das et al. (2000) found a 12% difference in output factors for field sizes less than 20 mm [70,71]. Other aspects to consider in small field dosimetry are that the full width half maximum (FWHM) does not represent the actual field size, and that mean beam energy increases the smaller the field sizes become due to the stereotactic collimator. There is difficulty in calculating the dose to patients and modelling the beams is also challenging [70].

The type of detector also matters. While minimising detector size is very important for small fields, if minimising ionisation chamber size to a mini- or micro-chamber, SNR will be low. Silicon detectors have been shown to be useful in small field dosimetry with their high SNR and high resolution [70].

2.5 Impact on Dosimetry within a Magnetic Field

Magnetic fields influence charged particles via the Lorentz force, which acts perpendicular to the direction of a particle, as well as the direction of the magnetic field. Hence, secondary electrons in the treatment beam are affected by this force; experiencing changes in direction via arc-shaped trajectories. Electrons
moving from a higher density to a lower density material, such as at air-skin interfaces, may undergo the ‘Electron Return Effect’ (ERE), where a particle’s trajectory when exiting the higher density material leads them to return into this material, causing a higher dose at this point [1,72]. Another dose effect is the resulting asymmetry of the point spread kernel; reducing the build-up distance of the radiation beam and causing an asymmetrical penumbra, shifted in the direction of the Lorentz force [1,73]. Spiralling contaminant electrons can also cause increased skin dose before being absorbed in the first few millimetres [74]. These dose effects are dependent on the magnetic field strength and the energy of the charged particle. Better MRI quality increases the dose effects due to the magnetic field; however, these effects can be modelled [1,75,76].

Raaijmakers et al. (2008) investigated ERE with GEANT4 simulations for four cases: ERE at the distal (Figure 2.2) and lateral sides of the beam, in cylindrical air cavities (1-30 mm in diameter) and in lung tissue (8 cm thickness), and for four magnetic field strengths: 0.2 T, 0.75 T, 1.5 T and 3 T. For the distal side, ERE was found to increase with increasing field size until the field size was wider than the largest electron trajectory diameter and the penumbral widths. As can be visualised in (Figure 2.2), distal ERE also increased with decreasing field strength down to 0.75 T. This was due to the reduced penetration depth of the secondary electrons at higher field strengths and the smaller volume segment from which the exit electron fluence originated. At 0.2 T, the dose decreased again due to most of the dose being deposited outside the field borders due to the larger secondary electron radius. For the lateral side, ERE increased with increasing field strength due to the greater fluence of electrons exiting through the lateral side from the deformation of the point spread kernel in the direction of the force. Near the distal side, the lateral ERE was added to the distal ERE, thus increasing the dose [1].

In air cavities, there was increased dose before the cavity due to electrons returning into the phantom, and therefore a region of decreased dose after
the cavity. The decreased dose region started at the bottom of the cavity for 0.2 T, but then rotated gradually to the side of the cavity for increased field strengths. This was due to the direction of the secondary electrons changing due to the Lorentz force. The rotation of the area of increased dose demonstrated a greater rotation due not only to the secondary electron direction change but also due to the radii of these electrons. This effect occurred until the radius of the air cavity was smaller than the radii of the secondary electrons, in which case, the electrons crossed the cavity as if there was no cavity present and the dose distribution was unaffected. This excluded higher field strengths, where an even smaller cavity than what was tested would be required as the radii of secondary electrons was still smaller than the smallest cavity. Finally, in lung tissue, the ERE increased the dose before the cavity and decreased the dose after the cavity. At 1.5 T the dose increase before the cavity was +49%, and the decrease after the cavity was -36%.

Raaijmakers et al. further demonstrated the dose effects on lung tissue by simulation of opposing beams to determine if the inhomogeneity due to the magnetic field could be negated. Opposing beams were found to improve the dose distributions, with the dose increase/decrease on either side of the cavity reducing to between -12% and +8% across all measured field strengths [1].
Dose effects are reduced by utilisation of a longitudinal magnetic field instead of a transverse field. The magnetic field is aligned to the direction of the electrons, and hence there is no force acting upon those electrons travelling forward. A focusing effect is observed instead, with any components of the momentum vector of each particle being brought into alignment with the magnetic field by ‘spiralling’, rather than a full circular trajectory [77–79]. This causes dose kernels to become more focused and elongated, and the effect is dependent on the irradiated material type. Narrower penumbral widths also result from the longitudinal field and consequently, a more conformal dose distribution is achievable [80].

However, there are disadvantages of the longitudinal field. The effect of non-purged electron contamination, where electron contamination scatters into the radiation beam from the flattening filter, secondary collimators or air column above the patient, rather than being scattered out of the field (by transverse fields), is problematic in longitudinal fields. Oborn et al. (2012) modelled both an integrated MRI-linac system (no air gap between linac collimation and MRI coil) and a separated MRI-linac system (air gap between collimation and coil) using GEANT4 Monte Carlo simulations. Different MRI magnetic field strengths from 0 T up to 3 T were simulated, as well as different fringe field strengths outside the magnet coils. Fringe field strengths included 0 T (a fully shielded collimation system), 0.06 T (limit of MLC motor operation), 1/r^5 drop off (non-shielded collimation system with active shielding from reverse coils), 1/r^2 drop off (non-shielded collimation and MRI system) and 3 T (if coil encompassed collimation system). An increase in skin dose by up to 1000% was found in some cases without magnetic shielding of the collimators. When collimators were fully shielded and using the lowest field strength of 0.25 T, skin dose still demonstrated an increase of 40-80%. The skin dose was dependent on the magnetic fringe field, MRI-coil thickness and linac isocentre distance. Increasing the fringe field lead to higher
electron contamination and less lateral scatter of this electron contamination, as did increasing the coil thickness. In general, the increase in radiation field size increased the electron contamination, except at higher fringe fields where the opposite effect occurred due to less collimator electron contamination falling within the field. Bolus would be required to lower the skin dose [81].

Spiralling contaminant electrons can also affect the surface dose in transverse magnetic fields. The transverse field deflects these electrons, reducing the surface dose within the field but depositing this dose to surfaces perpendicular to the magnetic field. A study was done on the 1.5 T/7 MV MRI-linac at Utrecht to investigate this contamination dose outside the field using EBT3 film. The film was placed both perpendicular and parallel to the magnetic field, at the surface of a water phantom and then at depths of up to 6.5 mm. The distance from the field edge was increased from 5 cm up to 20 cm. Different field sizes between $3 \times 3$ cm$^2$ and $15 \times 15$ cm$^2$ were investigated. For the perpendicular film, $5.6 \pm 0.2\%$ of the maximum central axis dose was recorded at the surface film 5 cm from the beam edge for a $10 \times 10$ cm$^2$ field size. For the parallel film, $1.4 \pm 0.1\%$ and $4.1 \pm 0.2\%$ of the maximum central axis dose was measured on either side of the field edge [74].

In terms of effects on the linac itself, the electron path inside the waveguide of the linac can also be affected by the magnetic field. This can lead to changes in energy fluence and dose distribution [82]. An extended isocentre distance (i.e. $>100$ cm) is required for the Australian MRI-linac design due to the beam collimation system not being able to fit inside the MRI bore. This is an advantage in terms of reducing the magnetic field effects on the linac (in particular, the electron gun, target, and MLCs), but a disadvantage in terms of decreased dose to the treatment target and larger collimation positioning error due to the greater distance [81].
The Utrecht group magnetically decoupled their linac and MRI to reduce the distortions on the MRI from the linac. This was achieved by utilising active magnetic shielding of the electron gun to produce a zero magnetic field at this location. The accelerator tube was also shielded to minimise the magnetic field as much as possible at this point [83].

Another study of note when considering magnetic field effects is that by Yang et al. (2015) [84]. This study examined the possibility of the more rotationally symmetric treatment modality of TomoTherapy for operation in a transverse magnetic field, with GEANT4 simulations. Their aim was to demonstrate that the rotational mode of dose delivery would reduce the dose perturbation effects by averaging, in contrast to the restriction of beam angles proposed in other studies where the aim to reduce the effects of ERE worsened the quality of the treatment plan [72, 84–86]. Three patient plans were investigated: a head-and-neck, lung, and prostate plan, each in field strengths of 0.35, 0.7, 1.5, and 3 T, and compared to zero magnetic field. The dose volume histograms (DVH) and dose heterogeneity index (DHI) of the simulations were compared to those of the treatment plan [84].

At zero magnetic field, the simulated plans agreed with the plans produced in the treatment planning system. For the plans simulated under a magnetic field, in the head and neck and the prostate plan, the largest dose perturbations were mostly ±3 Gy, with a few larger deviations of ±4 Gy. The DHI was approximately 3% in all magnetic field strengths. Skin dose decreased for the prostate case by up to 1 Gy for field strengths between 0.7 and 1.5 T due to electrons generated in the air around the patient moving in circular trajectories instead of reaching the patient. However, for the head-and-neck case, skin dose increased with field strength, with a 3-6 Gy increase observed at 3 T. As in the prostate case, the contamination electrons are kept circling the magnetic field lines of the patient, rather
than reaching the patient, however, in this case, the electrons from regions adjacent to the neck, accumulate around the neck. Dose perturbations were within $\pm 3$ Gy, excluding air cavities. DHI values for the head-and-neck plan simulation in 1.5 and 3.0 T were 5.3%. The lung case was the most affected by the presence of the magnetic field, with overdosing to regions in the planning target volume (PTV). DHIs of up to 20.1% (largest for 1.5 T) were observed. A dose reduction to the skin was observed for the 25% of the skin receiving the highest dose; however, the remainder received an increased dose [84].

It was found that TomoTherapy was useful to treat homogeneous sites, such as the prostate, in terms of reducing skin dose and perturbations in the beam penumbra, due to the low ERE present. Sites with low density regions, such as the lung, experienced compromised PTV coverage in a magnetic field. The authors recommend that, when creating a treatment plan in which the dose deposition will be influenced by a magnetic field, extra optimisation structures could be added to the plan to visualise magnetic field effects particular to that structure [84].

2.5.1 Modelling of Magnetic Field Effects in Treatment Planning Systems

Treatment planning systems used for MRI-linacs can now accurately model the effect of magnetic fields on charged particles. Hissoiny et al. (2011) investigated a fast graphics processing unit (GPU)-based Monte Carlo dose calculation platform for dose calculations in a magnetic field. The TPS calculations were validated by experimental measurements. Dose distributions passed a gamma analysis to with 2%/2 mm [75]. Chen et al. (2019) found that relative and absolute dosimetry measurements using various dosimeters on a 1.5 T/ 7 MV MRI-linac agreed with the beam model calculations of a prototype, Monte-Carlo-based TPS. The depth dose measurements and beam profile measurements (outside the penumbra region) were within 1% of the TPS. QA plans
successfully passed gamma analysis with various pass rates within clinical criteria. The study concluded that the TPS accurately accounted for the effect of the magnetic field [76].

2.5.2 Effects Due to Radiofrequency

Radiofrequency, or RF, interference is an issue for MRI, and hence also for MRI-linac systems. RF from external sources, such as power supplies or networks, can interfere with the signal received from the patient for image formation. Image artefacts can result in the reconstructed image [87].

When considering an MRI-linac, the RF from the linac needs to be accounted for, as incomplete RF shielding in a linac has been shown to degrade MR images [88]. A large source of noise results from the pulse power modulator, which has been investigated by Lamey et al. (2010). The authors suggest that the modulator should be housed in an RF shield, at a maximum distance from the MRI, or if possible, to position the modulator in a separate room from the MRI [89]. In addition to this, RF from the MRI can have an effect on electronics from devices such as radiation detectors. These devices need to therefore be appropriately shielded [90].

2.6 Radiation Detectors

There are various types of radiation detectors available for dosimetry. These include ionisation chambers, radiochromic film, thermoluminescent dosimeters (TLDs), fibre optic dosimeters, metal oxide semiconductor field-effect transistor (MOSFET), diamond and diode detectors. As well as point detectors, detectors are also available in array form; either in one, two, or three dimensions.
2.6.1 Ionisation Chambers

Ionisation chambers are used for reference measurements. An ionisation chamber consists of two electrodes separated by a gas-filled cavity. Leakage current is reduced by an insulator between the two electrodes, as well as the presence of a guard electrode, which directs the leakage current to ground so it is not collected by the collecting electrode. Ionisation chambers can be directly read out and do not have much variation in sensitivity over their lifetime [61]. Raw chamber charge readings \( M_{\text{raw}} \) are corrected by several correction factors, before conversion to absorbed dose \( D_w^Q \) by a calibration coefficient that was determined using \(^{60}\text{Co} (N_{D,w}^{^{60}\text{Co}})\) and a quality conversion factor \( k_Q \). \( k_Q \) is used to convert the calibration coefficient from \(^{60}\text{Co} \) to a beam of quality Q [82]:

\[
M = M_{\text{raw}} \times P_{\text{tp}} \times P_{\text{ion}} \times P_{\text{pol}} \times P_{\text{elec}}
\]  \hspace{1cm} (2.1)

where

- \( P_{\text{tp}} \) is the correction for temperature, air pressure and other ambient conditions
- \( P_{\text{ion}} \) is the correction for incomplete ion collection and other non-ambient conditions
- \( P_{\text{pol}} \) is the correction for polarity effects
- \( P_{\text{elec}} \) is the electrometer calibration factor [82].

\[
D_w^Q = M_{\text{raw}} k_Q N_{D,w}^{^{60}\text{Co}}
\]  \hspace{1cm} (2.2)

A correction factor \( k_B \) has been introduced to account for magnetic field effects [91], which will be discussed in Section 2.7.
Ionisation chambers are divided into two types, cylindrical and parallel plate. Cylindrical ionisation chambers can be utilised for both absolute and relative measurements of dose. These chambers consist of a central collecting electrode of aluminium and a conductive outer wall of graphite [61, 92]. Cylindrical chambers are utilised with a build-up cap for charged particle equilibrium (CPE) if in-air measurements are taken [92]. Parallel-plate chambers are of a flat, disc-like shape, with the central axis parallel to the direction of the beam. They have a shallow front window thickness of ≤1 mm, and a wider guard ring (to negate the in-scattering perturbation effect) [92–95]. These chambers are more advantageous than cylindrical chambers in electron beam absolute dose measurements of <15 MeV [92].

Attix chambers [96] are an example of parallel plate chambers, constructed mostly from solid water to reduce chamber perturbations [95]. These chambers contain a thin entrance window of 0.025 mm with a 48 µm deep effective point of measurement [97]. They have been shown to have a high accuracy in surface dosimetry measurements between +0.5% and −0.5% for 6 MV and 24 MV photon beams, respectively, due to their large-diameter guard ring [98–100]. These chambers have a small demonstrated polarity effect within the build-up region, with the highest effect observed at the surface (3.8% for a 10 × 10 cm² field with energies ranging from 6 to 24 MV [98, 101] and up to 6.2% for 6 and 18 MV beams over the range of field sizes between 5 × 5 cm² and 40 × 40 cm² [102]).

Ionisation chambers have been utilised in MRI-linac dosimetry studies [2, 82, 103], which will be discussed in Section 2.7.
2.6.2 Radiochromic Film

Radiochromic film is used as a radiation detector by measuring the change in its colour after exposure to ionising radiation. This change occurs when the photo monomer molecules in the film, which are colourless, undergo a chemical change due to the energy transfer from the incoming photons/particles. Film is scanned on a flatbed scanner at least 24 hours after irradiation to allow for density changes, maintaining consistent orientation and position of the film on the scanner to reduce uncertainties. The colour change is what is measured to determine net optical density \((OD_{net})\), which is ultimately proportional to the dose deposited \([104,105]\). \((OD_{net})\) is determined by:

\[
OD_{net} = \log_{10}\left(\frac{I_o}{I}\right)
\]  

(2.3)

where \(I_o\) is the background intensity and \(I\) is the final intensity. The dose is determined from the \(OD_{net}\) by a 3rd or 4th order film calibration curve.

Advantages of EBT3 film include its high spatial resolution (<25 µm), near tissue-equivalence and ability to self-develop [106]. Film can be used for commissioning measurements and patient-specific QA, and is useful in small field dosimetry and resolving beam penumbra [105]. EBT3 film was chosen as the benchmark detector in this study as it is relatively unaffected by a magnetic field [107], which will be discussed in Section 2.7.

2.6.3 Fibre Optic Dosimeters

Fibre optic dosimeters (FODs) detect radiation via changes induced in the optical signal [108]. These dosimeters include plastic optical fibres and organic scintillators, which use water-equivalent materials [109]. Their advantages include a real-time readout capacity, the fact that they use an optical signal instead of an electrical signal, and do not cause imaging artefacts. Therefore
these dosimeters are useful in an MRI-linac environment, as they are not af-
fected by electromagnetic interferences, do not interfere with imaging, and are 
made with non-magnetic materials [108,110].

2.6.4 Thermoluminescent Dosimeters

TLDs are read by the emission and detection of thermoluminescence after the 
dosimeter is heated, and the subsequent conversion of this light into electrical 
signal, which is proportional to photon fluence. This emission of light is due to 
the electrons being freed from their traps, which they were trapped in during 
irradiation. A ‘glow curve’ is a graph utilised to display the photon intensity 
of the thermoluminescence varying with temperature. The glow curve displays 
peaks, which correspond to the different energy traps of the TLD materi-
al, requiring different temperatures to release electrons from these traps [5,111].

Advantages of TLDs are their small size and variety of forms available, 
for example, chips, powders or rods, which allows for easier placement. Other 
advantages include that tissue-equivalent material can be used (for example, 
LiF:Mg,Ti), no cables are required, and that these dosimeters are inexpensive 
and are reusable after annealing, which also reduces the cost. Disadvantages 
include that the dose cannot be read out immediately due to the heating 
process required for thermoluminescence, and that read-out can also be lost in 
this process. The read-out and annealing process require a long period [61]. 
The effects of a 2.5 T magnetic field on TLD response have not been found 
to be significant at high doses [112] but still requires investigation at low 
doses [113].
2.6.5 Silicon Detectors

2.6.5.1 Semiconductor Physics

Silicon (Si) semiconductors can be either n- or p-type. N-type Si semiconductors are doped with impurities of a donor element. The donor contains a free electron as it is a pentavalent element, and hence its majority carriers are electrons, and minority carriers are positive charges (or ‘holes’). P-type Si semiconductors are instead doped with a trivalent element meaning it accepts an electron. It, therefore, has the opposite majority and minority carriers to n-type semiconductors [66].

When an n- and p-type semiconductor join, a p-n junction is created. By applying a reverse bias across the junction, a depletion layer with a thickness ranging from approximately 10 µm to 5 mm is formed by the majority carriers diffusing across this junction to the opposite side. A potential is then created, resulting in an electric field. When ionising radiation is introduced, electron-hole pairs are created, and these carriers can then diffuse across the depletion layer, and the charges can be collected and measured. The charges collected are proportional to the deposited energy from the ionisations. This is linear energy transfer (LET)-independent for lighter particles such as the electron. Indirect recombination can be used to determine the charges that are the result of the ionising radiation, and which charges are from the leakage current from any offset voltage of the electrometer [66,114].

Silicon detectors have greater dose measurement efficiency than ionisation chambers, allowing for much smaller detector volumes and therefore can be manufactured at a higher spatial resolution. These detectors can therefore measure the dose in high dose gradient regions. Real-time read-out is also an advantage. However, a few problems arise in their use. Long-term irradiation effects include the reducing of the diffusion length of the carriers and hence the
sensitivity of the detector is reduced. A change in temperature can affect the recombination of electrons and holes and the diode response [114, 115]. Other disadvantages of these detectors include dependence on temperature, dose rate, angle and energy [5, 61, 111]. Some examples of point and array Si detectors are discussed below.

2.6.5.2 **MOSkin™** Detectors

**MOSkin™** detectors are MOSFET detectors that were originally developed to measure the dose at air/skin interfaces [116]. MOSFET detectors are based on field effect transistors, which are made up of source, gate and drain terminals, that are then connected to the silicon dioxide (SiO$_2$) layer (gate terminal) and Si substrate. For an n-type MOSFET (which are the most common MOSFET detectors), the source and drain are of n-type material, with the Si being the opposite (i.e. p-type) [117]. When ionising radiation strikes the gate oxide of the MOSFET, electron-hole pairs are created, and the holes that do not undergo recombination then move towards the interface between the SiO$_2$ and the p-type Si substrate. A positive build-up charge is created when some of these holes are trapped near the interface. At a critical gate bias, known as the threshold voltage, an inversion layer (inversion of conductivity between drain and source) is formed, allowing a fixed current to flow in the n-type channel between the drain and source. This threshold voltage shift is proportional to the absorbed dose in the gate oxide [117–119]. MOSFETs can undergo irradiation in either active or passive mode, which means with or without a positive gate bias voltage, respectively. Active mode is employed to increase the sensitivity and linearity of the MOSFET’s response by reducing recombination of the electron-hole pairs; hence more holes are trapped, hence a larger difference in threshold voltage [118, 119].

**MOSkin** detectors have a reproducible Water Equivalent Depth (WED) of 70 µm, which allows for measurements at this depth [120], consistent with
ICRU recommendations for skin dosimetry [100, 121, 122]. This is the depth that corresponds to the basal layer, which is the first radiosensitive layer of the epidermis [123–125]. To achieve this depth measurement, a thin kapton film overlays the gate, which acts as a build-up layer. The film also protects electronics from damage from, for example, moisture [125].

2.6.5.3 Commercial Silicon Array Detectors

Commercial silicon array detectors are available in 2D and 3D arrangements. The ArcCHECK and Delta$^4$ are examples of quasi-3D array detectors that are commercially available. The ArcCHECK is of cylindrical design with 1386 silicon diodes spaced 1 cm apart. Both the array length and diameter are 21 cm. A PMMA phantom is built into the system [126, 127]. The Delta$^4$ has two crossed planes of silicon diodes inside a PMMA phantom. There are a total of 1069 diodes that have a pitch of 5 mm in the central $6 \times 6$ cm$^2$ region and 10 mm in the outer region [128]. These quasi-3D array detectors are used primarily for patient QA, for example, in IMRT and VMAT verification [129–132]. These current commercial systems are limited in their resolution; high resolution detectors will now be discussed.

2.6.5.4 High Resolution Silicon Array Detectors

Several high resolution silicon detectors described here (and in Figure 2.3) are of various design, differing in the number of diodes, the layout, pitch and the use of either separate epitaxial diodes or monolithic substrate technology. The MagicPlate is a two-dimensional array detector available in different configurations, as the MagicPlate-121 (MP121) and MagicPlate-512 (M512). The MP121 has 121 separate epitaxial diodes in an $11 \times 11$ array embedded in a kapton substrate, whereas the MagicPlate-512 has 512 channels in a monolithic Si detector. The pitch of the MP121 is 1 cm compared to the smaller 2 mm pitch of the M512. “Drop-in” technology was utilised to mount the diodes of the MP121 in the kapton, to both minimise energy and angular dependence.
The MagicPlate can be utilised as a transmission detector and to measure the in-phantom dose [133–135]. The Dose Magnifying Glass, or DMG, also has various configurations with diodes arranged in one or more linear arrays, such as the original DMG, serial DMG (sDMG), Octa and DUO. The DMG has 256 channels and is a p-type Si strip, 1D detector. The sensitive area is $20 \times 2000 \mu m$, with a $375 \mu m$ thick Si wafer. The sDMG was designed to measure the penumbral dose for small fields used in SRS and SBRT. The original DMG and sDMG both have a pitch of $200 \mu m$. The Octa is a monolithic silicon pixelated detector, with eight linear arrays starting at the centre and extending outwards. The pitch for Octa is $0.3 \text{ mm}$ and $0.43 \text{ mm}$ for the vertical/horizontal arrays and diagonal arrays, respectively [134–136].

The M512 and DUO will be discussed in more detail as they were utilised for this project. These detectors were chosen for this study due to their smaller sensitive volume, higher spatial resolution and larger dynamic range compared to ionisation chambers [29], as well as the fact that they do not have a large air
cavity, which would be affected by ERE to a lesser extent [82]. Their real-time readout capability is also an advantage over film.

The M512 and DUO are $52 \times 52 \text{ mm}^2$ in dimension and wire-bonded to a printed circuit board (PCB). These detectors are based on a monolithic, p-type silicon substrate with ion-implanted diodes (pixels) connected to pads along the perimeter. The detectors are available in both bulk silicon and epitaxial forms. The M512’s 512 pixels are arranged in a $22 \times 22$ array with another 7 pixels on each side of the array. Each sensitive volume has dimensions of $500 \times 500 \times 100 \mu\text{m}^3$. The DUO has 505 diodes arranged in two orthogonal, linear arrays. Each sensitive volume has dimensions of $40 \times 800 \times 100 \mu\text{m}^3$, except for nine smaller central diodes with dimensions of $180 \times 180 \times 100 \mu\text{m}^3$. The detector pitch is 200 $\mu\text{m}$. For protection, the detector is encased between two 5 mm thick slabs of PMMA. There is a small air gap just above the detector to negate the dose enhancement of silicon that is especially apparent at small fields and to protect the very fine substrate wire-bonding [29,137].

These detectors are operated in passive mode. The electronic readout for the M512 is a Data Acquisition (DAQ) System based on two analogue front end (AFE0064) chips per PCB. The chips are charge integrators, with 64 parallel inputs integrating the charge simultaneously and two differential outputs. After the charge has been converted to a voltage level it is stored in an analogue buffer, where it is sampled and sent to the analogue-to-digital converter (ADC). A dead time of about 70 $\mu\text{s}$ is observed during output read-out. The field-programmable gate array (FPGA) both drives and synchronises the AFE and ADC. The readout system can be timed (frequency can be adjusted up to 5 KHz) to match the electron gun pulse (3.7 $\mu\text{s}$) of the linac to avoid any loss of charge by allowing the dead time to fall between linac pulses (i.e. integrating the charge only when the beam is switched on). In this way,
electronic noise and leakage current are also minimised [138].

The M512 was previously characterised by Aldosari et al. (2014). The tests performed were uniformity, linearity (between 50 and 500 MU), repeatability, dose per pulse (DPP, from $0.9 \times 10^{-5}$ to $3.4 \times 10^{-4}$ Gy/pulse), output factors (for field sizes of $1 \times 1$ cm$^2$ up to $10 \times 10$ cm$^2$), percentage depth dose (from 1.5-30 cm depth), and beam profiles for 0.5-10 cm$^2$ field sizes. The effect of the M512 packaging on the response was also evaluated. The radiation stability of the M512 was investigated and found to be only 1%/10 kGy$_{H20}$ for a Co-60 source and 0.9%/10 kGy$_{H20}$ for an 18 MV linac. Uniformity was 0.25% deviation from the mean value after equalisation (equalisation will be described in Chapter 3). Good linearity and repeatability were determined, with a $R^2$ (regression coefficient) of 0.9988 and 0.2% deviation respectively. In comparison to output factors measured by a PTW diamond, Scanditronix point EDD-2 diode and MOSkin$^{TM}$ detector, and Gafchromic EBT3 film, the M512 agreed with these detectors to within 2%. This was excluding the $1 \times 1$ cm$^2$ field size, where an over-response of <4% was observed for the M512, due to the increased percentage of scattered electrons from both the silicon and packaging at this field size. When compared with EBT3 film for beam profile measurements, the M512 agreed with the film measurements to within 1.36%. Depth dose measurements were within ±1% of ionisation chamber measurements. The detector packaging was only found to cause a deviation of +0.64% of the response. Dose per pulse measurements were within 5% over the range measured [29].

The DUO was also previously characterised by Shukaili et al. (2017). The characterisation tests completed were DPP dependence (from $2.8 \times 10^{-5}$ to $1.2 \times 10^{-4}$ Gy/pulse), percentage depth dose (down to a depth of 25 cm), beam profiles (0.5-5 cm$^2$ field sizes), and output factors (0.5 $\times$ 0.5 cm$^2$ up to 30 $\times$ 30 cm$^2$). DPP was 5% between $2.7 \times 10^{-4}$ to $1.2 \times 10^{-4}$ Gy/pulse.
but increased to 23% when extending the range of measurements to $2.8 \times 10^{-5}$ Gy/pulse. The DUO agreed to ionisation chamber percentage depth dose measurements to within 1.5%. FHWM and penumbral widths of profiles were within 1% and 0.5 mm of EBT3 film, respectively. The DUO output factors agreed to within 1.8% of EBT3 film and MOSkin detectors if a 0.5 mm air gap was present above the sensitive volumes of the DUO [137].

The M512 and DUO were found to be suitable for SRS/SBRT field size measurements. The effect of magnetic fields on these detectors will be investigated in this work.

### 2.7 Use of Detectors within a Magnetic Field

Several detectors have previously been tested in the presence of a magnetic field, either experimentally and/or with simulations. These include film (EBT2 and EBT3) [107, 139], various ionisation chambers [2, 3, 82, 91], an ionisation chamber array (IC PROFILER\textsuperscript{TM}) [103], diamond detector (PTW 60003) [140], a single diode detector (IBA PFD) [140], and silicon array detectors (MP121 and M512 [4, 80], ArcCHECK-MR diode array [141] and MR-Delta4 [142]). In most cases, correction factors were applied to the detector response to account for the influence of the magnetic field. For example, for ionisation chambers, the magnitude of these correction factors depended upon the magnetic field strength and orientation, radiation beam orientation, and the ionisation chamber geometry and orientation [3, 82, 103].

As mentioned previously, EBT3 film was the benchmark detector for this work. EBT2 film has shown a decrease in OD of up to 15% due to changes in the orientation of monomer crystals within the active layer under the influence of a 0.35 T magnetic field [139]. For EBT3, the difference between 0 T and 1.42 T was only just outside of uncertainty: a decrease in OD of 0.8%,
as well as a 2.1% shift towards higher doses for corresponding OD values [107]. For this reason, film will be a useful dosimeter in the MRI-linac to assist in distinguishing between intrinsic and extrinsic effects of a magnetic field on other detectors.

For ionisation chambers, the raw charge reading \( (M_{raw}) \) depends directly on the number of electrons entering the chamber and the electron track length. Ionisation chambers are also an air cavity. Hence, the magnetic field would have an effect on response due to the Lorentz force acting upon the secondary electrons [2, 82]. The response of a Farmer NE2571 ionisation chamber in a transverse magnetic field was simulated utilising GEANT4 Monte Carlo simulations by Meijsing et al. (2009). This chamber had a sensitive volume of 0.69 cm\(^3\). Two chamber orientations with respect to the radiation beam and the magnetic field were modelled (Figure 2.4). Experimental measurements were performed in a magnetic field of strength 0 T up to 1.2 T, and simulations were performed in a magnetic field of strength 0 T up to 2 T. When the chamber was perpendicular to the radiation beam (configuration I), there was close agreement between simulations and measurements. When the chamber was parallel to the radiation beam (configuration II) there was a slight difference between simulations and measurements. However, a slight misalignment between the magnetic field and radiation beam of 3°, which would more greatly affect configuration II because of its rotational asymmetry (unlike configuration I), was believed to be the cause of the discrepancies in this case. Hence, for both orientations, the measurements agreed with the simulations to within the uncertainty (i.e. to within 2.5% standard deviation). The chamber geometry affected the response, with the response from configuration I increasing up to 1 T and then decreasing, and the response from configuration II decreasing at first and then increasing after 1 T [2].

Monte Carlo simulations have also been utilised by Reynolds et al. (2013) to model the response of a NE2571 ionisation chamber (to compare to the results
of the Meijsing et al. 2009 study) and a PR06C ionisation chamber (due to its widespread clinical use) in the presence of both transverse and longitudinal magnetic fields of various strengths (0-1.5 T). The orientation of the chamber was varied between being perpendicular and parallel to the radiation beam (Figure 2.5). The response of the PR06C chamber (with a sensitive volume of 0.69 cm$^3$) was also measured experimentally in various magnetic field strengths ranging from 0 T to 0.21 T, using a transverse magnetic field only. The longitudinal orientation was not possible for experimental measurement due to the magnet set-up available at the time of the experiment. For the simulations of the transverse field, an 11% change in response was observed at approximately 1.0 T. The responses for the transverse field followed the same trend as those responses acquired by Meijsing et al. (2009). For the longitudinal field, there was no noticeable reliance on chamber orientation, with 1% change up to 1.0 T, and a 2% increase in response at 1.5 T. This was due to the focusing of electrons into the chamber when utilising the longitudinal field. The experimental
measurements matched closely with the transverse field simulations (average deviation of 0.45%), for the range of field strengths over which measurements were taken. This was expected as both the NE2571 and PR06C have a similar design. For the use of these ionisation chambers in a magnetic field, a correction factor of 0.935 was recommended. However, this was dependent upon the various factors already mentioned, such as chamber orientation and magnetic field strength. Ion recombination and polarity correction factors were also investigated by both simulations and measurements, but were found to be unaffected: the results were the same to within uncertainty of the results acquired when no magnetic field was present [3].

Smit et al. (2013) investigated the NE2571 Farmer ionisation chamber in terms of optimal chamber orientation with respect to the magnetic field (1.5 T), as well as the effect of the magnetic field on linearity and repeatability. The effect on ion recombination and polarity correction factors was also investigated but
found to be negligible. Linearity and repeatability were found to be unaffected by the magnetic field; with a difference of 0.2% observed for linearity and a 0.1% difference for repeatability, when compared to ideal linear values. In terms of chamber orientation, the total change in chamber response at 1.5 T was <0.2% at 90° chamber orientation; hence this would be preferable for dosimetry measurements. Although a larger correction would be needed for this orientation (due to an over-response of 8.8%). The total effect of the 1.5 T magnetic field on the chamber reading was measured to be an average of 4.9% increase, with a correction factor of 0.953 required. The authors estimated an uncertainty in the correction factor to be 1%; however, they recognised that more data was needed to provide a proper uncertainty analysis. The determined correction factor holds only for the set-up utilised in this study [82].

In a recent study by Pojtinger et al. (2018), magnetic field correction factors, $k_B$, for eight different ionisation chambers – both thimble and parallel-plate – were determined. The magnetic field strengths and beam energies used were to replicate those of the Unity and ViewRay systems, i.e. 1.5 T/7 MV and 0.35 T/6 MV, respectively. A transverse magnetic field was used, as this too is utilised in both systems. Experimental results were compared to simulated results. The thimble chambers investigated were: PTW 30013, PTW 30015, PTW 30016 and NE2571. The parallel-plate chambers investigated were: Roos Chamber, NACP-02, Markus and Advanced Markus. The magnetic field effects on the dose distribution and effects on the chamber itself were separated by dividing the total change of dose in a chamber by the relative change of dose caused by the shifted profile. Both a parallel (180°) and perpendicular (90°) orientation of the thimble chamber with respect to the magnetic field were utilised, while the parallel-plate chambers were positioned in their standard orientation. For the Unity set-up, the highest $k_B$ for the thimble chambers was found to be for PTW 30013, where the response increased by up to 4.9% for the 90° chamber orientation. For the 180° orientation, changes in response of
<0.8% were observed. For the parallel-plate chambers, a greater increase in
dose response was observed, the highest being for the Roos Chamber of up to
7.9%. The ViewRay set up showed similar results for the thimble chambers
and only some differences for the Roos and NACP-02 parallel-plate chambers.
Overall, the maximum increase in response was observed at 1 T. There was
<2% difference between experimental and simulated results. It was concluded
that changes in response differ due to magnetic flux density, chamber geometry
and orientation [91].

The air gap between ionisation chamber wall and the phantom insert
has also been investigated in a 1.5 T MRI-linac to determine its influence on
chamber response [143]. The responses measured with an air gap present were
compared to those measured when the air gap was filled with water. The
measured responses were 0.7-1.2% higher when the air gap was filled with
water when compared to air; the difference was 0.3% on a conventional linac.
The authors concluded that existing solid phantoms with air gaps present
surrounding the chamber are inappropriate for dosimetry on an MRI-linac [143]

The IC PROFILER, a multi-axis ionisation chamber array, was investi-
gated for performance in a magnetic field by Smit et al. (2014) and validated
with film. The IC PROFILER has 251 parallel plate ionisation chambers, each
with a sensitive volume of 0.05 cm³, which is smaller than the sensitive volumes
of those chambers used for reference dosimetry. Hence, it was expected that
these ionisation chambers would be affected to a lesser extent by the magnetic
field, and that the readings could be normalised to a reference chamber (NE2571
Farmer ionisation chamber), thus eliminating the requirement for a magnetic
field correction factor. The detector pitch on the x- and y-axes was 0.5 cm, and
0.71 cm on the diagonal axes. The power supply on the panel was replaced with
an external power supply to make it MR-safe. This array performed well in a
1.5 T field. The central ionisation chamber was used to normalise the signal
for chambers on all axes for normalised profile measurements. Unnormalised readings can be used for measurements such as output factor measurements, however, prior to this, the authors recommend allowing a 15-minute standby time, after which, a background correction should be completed. The readings were shown to be unaffected by transverse magnetic fields in terms of linearity, reproducibility and pulse rate frequency dependence. Ionisation chamber shape and orientation required some consideration but were found to not significantly alter the readings. Rotating the panel by 5°C around the central axis demonstrated <0.2% deviation (and 1% in the penumbral region) between the case where no magnetic field was present and the case where a magnetic field was present. Deviations between film and IC PROFILER results were due to noise from the film [103].

For in-air simulations of the responses of a PTW60003 diamond detector and IBA PFD diode detector by Reynolds et al. (2014), correction factors for use in a magnetic field were recommended. The simulated magnetic field strengths were 1-1.5 T. Measurements were also performed for the field strengths available experimentally, which were 0 T up to approximately 0.18 T. The diamond detector had a sensitive volume of 1.47 mm in radius and 0.25 mm in thickness. The diode detector had a sensitive volume of 1.25 mm in radius and 0.5 mm in thickness. As in their study on the responses of ionisation chambers in a magnetic field, Reynolds et al. investigated both orientations of the magnetic field and two orientations of the detectors: both parallel and perpendicular to the radiation beam. The detectors utilised here were asymmetric along their long axes, hence another orientation for each detector was also investigated, with the detectors perpendicular to the radiation beam but rotated 180° around their long axis in the transverse field. A sixth orientation was also analysed, where the magnetic field was parallel to the long axis of the detector, but with both magnetic field and chamber long axis perpendicular to the radiation beam [140].
A larger effect on the response in a transverse field was observed (up to almost 20% deviation at 1.5 T for the perpendicularly-orientated detectors), when compared to the longitudinal field orientation (only 0.5-1% deviation at 1.5 T). Hence, for the longitudinal field, no correction factor was needed up to 1 T and minimal correction was required after this up to 1.5 T. Measurements agreed with the simulations to within 0.5% [140].

The responses of both detectors were also simulated in a water tank for both magnetic field orientations at 0.5 T; however, these simulations could not be verified experimentally due to the physical size of the water tank not being able to fit in the electromagnet. The diode detector displayed a closer match between in-air simulations and water tank simulations. However, in the penumbral region; both detectors failed to accurately measure the change in dose to water in transverse magnetic fields, although the authors did not provide a reason as to why this occurred. As with ionisation chambers; the response of the detector was dependent upon its orientation to the magnetic field. Because different energies were simulated in this study; it was found that the photon beam energy affected the response of the detector. The correction factor recommended overall was 0.935. From modelling two different solid state detectors; it was demonstrated that detector composition also influenced the responses. The authors recognised that their study presented the overall effect of the magnetic field on detector response rather than separating their analysis into the intrinsic response of the detector and the effect on secondary electrons depositing dose in the sensitive volume [140].

Houweling et al. investigated the performance of the ArcCHECK-MR diode array under transverse magnetic field conditions on a 1.5 T MR-linac. This detector can be used for patient-specific QA. The ArcCHECK-MR has 1386 diodes with 10 mm pitch on a cylindrical water phantom. It differs from
the non-MR-compatible version of the ArcCHECK by extending the distance of
the power supply out of the high-field region. Comparisons were made between
a conventional linac and MR-linac; hence the energy spectra were different
at 6 MV and a nominal beam energy of approximately 8 MV, respectively [141].

Short term reproducibility demonstrated a standard deviation of 0.05%
and 0.06% for the conventional linac and MR-linac, respectively. Dose linearity
differed by 2.1% for the conventional linac and 0.7% for the MR-linac. Both
linacs demonstrated a maximum dose rate difference of 1.0%. Field size
dependence was 2.4% for a conventional linac and 2.6% for the MR-linac for
field sizes above $5 \times 5 \text{ cm}^2$, which is similar to previously reported results
by Li et al. (2013) without a magnetic field [141, 144]. The dose per pulse
dependence utilised a method of comparing a diode on the top and the base
of ArcCHECK-MR due to the limited space in the MR bore, and hence,
the source-to-surface distance (SSD) approach could not be used. The ratio
between the two diodes was found to be 2.7 for the conventional linac and 2.9
for the MR-linac. For both linacs, the inter-diode dose response variation was a
standard deviation of $<3\%$. The characteristics measured for the ArcCHECK-
MR were found not to be affected by a 1.5 T transverse magnetic field. The
authors recognise that the MR-linac has an unflattened field (in comparison to
the flattened field of the conventional linac). They do however also argue that
the MR-linac beam profile is different from conventional unflattened beams
due to the MR cryostat, which behaved somewhat as a flattening filter, and
hence there was no similar beam with which to compare their results. Another
difference between the two linacs was the SAD, which was 142.7 cm for the
MRI-linac, in comparison to the standard 100 cm of the conventional linac [141].

The magnetic field-compatible version of the Delta4 detector system,
named MR-Delta4, was characterised on a 7 MV/1.5 T MRI-linac with a
transverse magnetic field, as well as on a 6 MV conventional linac. Like the
standard Delta4, the MR-compatible version has a total of 1069 diodes across two orthogonal planes. Modifications were made to the original device to make it MR-compatible, such as extending the power supply cables so that the power supply could be positioned in the 5 gauss region. On the conventional linac, the MR-Delta4 and a standard Delta4 were also compared for the various characterisation measurements. The short-term reproducibility and dose linearity were investigated, as well as field size, dose rate, dose-per-pulse and angular dependency. Short-term reproducibility was found to vary by \(<0.1\%\) and dose linearity by \(\leq0.5\%\). Field size dependency was \(<2.0\%\) for field sizes larger than \(5 \times 5\) cm\(^2\) and dose rate dependency was \(<1.0\%\). For angular dependency, the largest difference between any two angles was 11.5\%. The dose-per-pulse dependency was \(<0.8\%\) and was found to be significantly different between the two devices. However, the maximum difference was only 0.5\%. The MR-Delta4 was therefore considered to be suitable for use in the 1.5 T MRI-linac [142].

Gargett et al. (2015) utilised GEANT4 to simulate the response of the MP121 inside a \(30 \times 30 \times 30\) cm\(^3\) water equivalent phantom at \(d_{\text{max}}\), in both longitudinal and perpendicular magnetic fields and a 6 MV photon beam. Results were compared to a water-equivalent volume. Dose spread arrays generated by tracking particles originating in a \(1\) mm\(^3\) voxel at 1.5 cm depth were assessed for 0, 1, 1.5 and 3 T. A \(2 \times 2\) cm\(^2\), \(4 \times 4\) cm\(^2\) and \(10 \times 10\) cm\(^2\) jaw-defined field were investigated at 1 T. Various single-segment MLC patterns of different shapes and size were tested to investigate the detector response at varying distances from the edge of the beam, also at 1 T. A \(4 \times 4\) cm\(^2\) jaw-defined field was tested at 1.5 and 3 T. Finally, a 12-segmented IMRT beam was also simulated at 1 T to investigate in-field dose gradients [80].

The dose spread arrays became more elongated with increasing field strength for the longitudinal magnetic field. This effect was more prominent in water
(at 3 T: 1.5 mm narrowing, 3 mm lengthening in the forward-scattered direction, lengthening in backscattered direction) compared to the silicon (0.5 mm narrowing at 3 T, negligible lengthening for all field strengths). For the perpendicular field, the dose spread arrays displayed a lateral shift, with dose spread arrays also shortening. This effect was more pronounced in water, with arrays shortening by up to 9.5 mm at 3 T and shifting laterally by a maximum of 4.8 mm at 1.5 T. In silicon the shortening was less: 1.8 mm at 3 T, and also a lesser lateral shift of 1.8 at 3 T [80].

For the various single-segment MLC-defined fields, less than ±1% difference was observed between the silicon array and water for the longitudinal field for all beams and magnetic field strengths. There was an increase in surface dose for the 10 × 10 cm² field size due to the electron contamination being focussed towards the interface between air and water. A dose shift was observed in the penumbral regions for the perpendicular orientation of the magnetic field, due to the dose increase resulting from the smaller radius of the low-energy electrons in these regions. A maximum of 4.3% over-response and maximum of 1.8% under-response; displaced in the direction of the Lorentz force was observed. This effect decreases as field size increases due to lateral electron equilibrium. In-field dose gradients of the IMRT beam did not display this dose shift [80].

Gargett et al. (2018) further investigated silicon array detectors under a magnetic field, this time with the M512. Monte Carlo simulations and experimental measurements were completed on a permanent magnet system (described in Chapter 5). Inline and perpendicular orientations of the magnetic field were investigated. The experimental component comprised small field size beam profile measurements ranging from 0.75 × 0.75 cm² to 2.25 × 2.25 cm². Field strengths of 0.95 T and 1.20 T were investigated for inline and transverse fields, respectively. FWHM and penumbral widths of profiles measured with
the M512 agreed with EBT3 film to within 0.5 mm. A decrease in maximum
dose under the perpendicular field of up to 10% was observed with increasing
air gap size above the detector (up to 2 mm). Under inline field conditions,
this increase was only 2% [4].

Simulations involved modelling the M512 inside of a $30 \times 30 \times 30$ cm$^3$
water equivalent phantom, and comparing to a water-only set-up to distinguish
between intrinsic magnetic field effects on the detector and extrinsic effects
on the dose distribution. Magnetic field strengths were 1 T and 3 T and
compared to zero field. For the water-only geometry, the transverse magnetic
field changed FWHM by $\leq 0.1$ mm, however, there was a slight disparity in
penumbral widths between left and right sides for the smallest field sizes,
which decreased as the field size increased. For the M512 geometry, penumbral
widths were within 0.5 mm of water except at 3 T, which showed a difference
of 0.9 mm. The inline magnetic field did not significantly affect FWHM or
penumbral widths in the water-only geometry. There was an increase in dose
maximum, most prominent at 3 T. For the M512 geometry, FWHM was within
0.2 mm of water for field sizes larger than $1 \times 1$ cm$^2$. The inline field affected
the penumbra by only 0.3 mm, however under 1 T, the M512 differed from
water by 1 mm. The M512 was found to be suitable for use in measuring
FWHM and penumbral widths under a magnetic field [4].

A summary of the detectors investigated under magnetic fields has been
summarised in Table 2.1. Characterising detectors for use in magnetic fields
is an integral part of improving treatment outcomes, to determine which
detectors are suitable for QA in treatment delivery of the MRI-linac. As well
as ionisation chambers, semiconductor detectors present a promising option
for utilisation in magnetic fields. From current studies, it can be seen that
the breakdown of the change in intrinsic response of the detector due to the
magnetic field, and the extrinsic effect of the field on secondary electrons also
requires analysis.
Table 2.1
Summary of detectors investigated in the presence of a magnetic field. Detector and field orientations are written with respect to radiation beam direction. Abbreviations/symbols used in table: w.r.t. = with respect to, $\vec{B}$ = magnetic field, CF = correction factor, IC = ionisation chamber, $\perp$ = perpendicular, $\parallel$ = parallel, expt = experiment, sim = simulations, CAX = central axis.

<table>
<thead>
<tr>
<th>Detector Type</th>
<th>Publication</th>
<th>Detector Orientation</th>
<th>$\vec{B}$ Strength</th>
<th>$\vec{B}$ Orientation</th>
<th>CF Recommended</th>
</tr>
</thead>
<tbody>
<tr>
<td>IC: NE2571</td>
<td>Meijsing et al. 2009 [2]</td>
<td>$\perp$, $\parallel$</td>
<td>0-1.2 T (expt), 0-2 T (sim)</td>
<td>$\perp$</td>
<td>None</td>
</tr>
<tr>
<td></td>
<td>Reynolds et al. 2013 [3]</td>
<td>$\perp$, $\parallel$</td>
<td>0-1.5 T (sim)</td>
<td>$\perp$, $\parallel$</td>
<td>0.935 (1 T)</td>
</tr>
<tr>
<td></td>
<td>Smit et al. 2013 [82]</td>
<td>Changes incrementally from $\perp$ to $\parallel$</td>
<td>1.5 T (expt)</td>
<td>$\perp$</td>
<td>0.953 (1.5 T)</td>
</tr>
<tr>
<td></td>
<td>Pojtinger et al. 2018 [91]</td>
<td>$\perp$, $\parallel$</td>
<td>0-2.5 T</td>
<td>$\perp$</td>
<td>0.9700 ($\perp$, 0.35 T), 0.9638 ($\perp$, 1.5 T), 0.9995 ($\perp$, 0.35 T), 0.9993 ($\perp$, 1.5 T)</td>
</tr>
<tr>
<td>IC: PR06C</td>
<td>Reynolds et al. 2013 [3]</td>
<td>$\perp$, $\parallel$</td>
<td>0-1.5 T (sim), 0-0.21 T (expt)</td>
<td>$\perp$, $\parallel$ (sim), $\perp$ (expt)</td>
<td>0.935 (1 T)</td>
</tr>
<tr>
<td>IC: PTW 30013</td>
<td>Pojtinger et al. 2018 [91]</td>
<td>$\perp$, $\parallel$</td>
<td>0.25 T</td>
<td>$\perp$</td>
<td>0.9684 ($\perp$, 0.35 T), 0.9535 ($\perp$, 1.5 T), 0.9976 ($\perp$, 0.35 T), 0.9993 ($\perp$, 1.5 T)</td>
</tr>
<tr>
<td>IC: PTW 30015</td>
<td>Pojtinger et al. 2018 [91]</td>
<td>$\perp$, $\parallel$</td>
<td>0.25 T</td>
<td>$\perp$</td>
<td>0.9694 ($\perp$, 0.35 T), 0.9788 ($\perp$, 1.5 T), 0.9980 ($\perp$, 0.35 T), 0.9993 ($\perp$, 1.5 T)</td>
</tr>
<tr>
<td>IC: PTW 30016</td>
<td>Pojtinger et al. 2018 [91]</td>
<td>$\perp$, $\parallel$</td>
<td>0.25 T</td>
<td>$\perp$</td>
<td>0.9770 ($\perp$, 0.35 T), 0.9554 ($\perp$, 1.5 T), 0.9977 ($\perp$, 0.35 T), 0.9993 ($\perp$, 1.5 T)</td>
</tr>
<tr>
<td>IC: Roos</td>
<td>Pojtinger et al. 2018 [91]</td>
<td>$\perp$</td>
<td>0.25 T</td>
<td>$\perp$</td>
<td>0.969 (0.35 T), 0.9272 (1.5 T)</td>
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<tr>
<td>IC: NACP-02</td>
<td>Pojtinger et al. 2018 [91]</td>
<td>$\perp$</td>
<td>0.25 T</td>
<td>$\perp$</td>
<td>0.9765 (0.35 T), 0.9772 (1.5 T)</td>
</tr>
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<td>IC: Markus</td>
<td>Pojtinger et al. 2018 [91]</td>
<td>$\perp$</td>
<td>0.25 T</td>
<td>$\perp$</td>
<td>0.9993 (0.35 T), 0.9720 (1.5 T)</td>
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<tr>
<td>IC: Adv. Markus</td>
<td>Pojtinger et al. 2018 [91]</td>
<td>$\perp$</td>
<td>0.25 T</td>
<td>$\perp$</td>
<td>0.9920 (0.35 T), 0.9869 (1.5 T)</td>
</tr>
<tr>
<td>IC PROFILER™</td>
<td>Smit et al. 2014 [103]</td>
<td>Panel $\perp$ and rotated around CAX between $\pm5^\circ$</td>
<td>1.5 T (expt)</td>
<td>$\perp$</td>
<td>None</td>
</tr>
<tr>
<td>PTW 6003 Diamond</td>
<td>Reynolds et al. 2014 [140]</td>
<td>$\perp$, $\parallel$, rotating long axis in $\perp \vec{B}$</td>
<td>0-1.5 T (sim), 0-0.18 T (expt)</td>
<td>$\perp$, $\parallel$</td>
<td>0.935</td>
</tr>
<tr>
<td>IBA PFD Diode</td>
<td>Reynolds et al. 2014 [140]</td>
<td>$\perp$, $\parallel$, rotating long axis in $\perp \vec{B}$</td>
<td>0-1.5 T (sim), 0-0.18 T (expt)</td>
<td>$\perp$, $\parallel$</td>
<td>0.935</td>
</tr>
<tr>
<td>ArcCHECK-MR</td>
<td>Houweling et al. 2016 [141]</td>
<td>N/A</td>
<td>1.5 T (expt)</td>
<td>$\perp$</td>
<td>None</td>
</tr>
<tr>
<td>MR-Delta4</td>
<td>De Vries et al. 2018 [142]</td>
<td>N/A</td>
<td>1.5 T (expt)</td>
<td>$\perp$</td>
<td>None</td>
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<tr>
<td>MP121</td>
<td>Gargett et al. 2015 [80]</td>
<td>$\perp$</td>
<td>0-3 T (sim)</td>
<td>$\perp$, $\parallel$</td>
<td>None</td>
</tr>
<tr>
<td>M512</td>
<td>Gargett et al. 2018 [4]</td>
<td>$\perp$</td>
<td>0-3 T (sim), 0-1.20 T (expt)</td>
<td>$\perp$, $\parallel$</td>
<td>None</td>
</tr>
</tbody>
</table>
2.8 Conclusion

Magnetic field effects on dose distribution such as ERE, point spread kernel asymmetry, penumbral width narrowing and increase in skin dose, have been discussed. Part of this thesis focusses on how these effects differ in phantoms of solid water compared to the lower density of solid lung. For the dose to be measured accurately on an MRI-linac, it is integral that the effect of the magnetic field on detector response and the dose distribution is quantified. Several studies that measured these effects on various ionisation chambers, diode and diamond detectors were discussed. Adjustment to the calibrated dose under a magnetic field means that a new dosimetry protocol is required, or at least a modification of the correction factor, \( k_Q \) \((k_B)\), for existing protocol. High resolution silicon array detectors are investigated in this work due to their closer tissue-equivalence and higher resolution than ionisation chambers, and their instantaneous readout when compared to film.

MRI-linacs can also track treatment sites with MR imaging to deliver the dose more accurately to a moving target. A method is needed to perform a pretreatment check, to ensure that the dose distribution of a patient’s treatment plan is the dose delivered to the patient. MRI-linacs allow these QA checks to be performed not just in space but in time as well, by tracking the detector, so that the moving dose delivery can be measured. The novel silicon array detectors investigated in this work are ideal for this kind of QA measurement due to their high spatial and temporal resolution. A preliminary investigation of this in the static case is also carried out in the course of this work.
Chapter 3

Experiments with Gel-Water: Towards MRI-Linac Dosimetry and Imaging

Sections of this chapter are presented in the publications:


3.1 Introduction

To validate 4D tumour tracking, a QA concept which combines the M512 with MRI-compatible phantoms on a moving platform is currently being developed. This QA concept has been named ‘MR dynamic dosimaging’ and
it would enable a shifting detector to undergo DMLC tracking with imaging, whilst simultaneously measuring dose for plan verification, i.e. simultaneously verifying the dose placement and dose value. To achieve dynamic dosimaging, accurate MR visible/compatible detector systems are required. MRI requires phantoms with a strong enough signal for imaging, which is difficult. MRI cannot easily image solids due to their short T2 components (unless using non-standard methods such as ultra-short TE) [145]. Dosimetry requires phantoms that are tissue-equivalent in terms of effective atomic number ($Z_{eff}$) for MV x-ray beams [5]. Dosimetric and geometric measurements should fall within clinical tolerances; these are generally within 2% and 1 mm, respectively [146]. The substantial water content of hydrogel materials may make them ideal materials to meet these two requirements. One such hydrogel was investigated for concurrent use with the M512 and in place of solid water or water. This hydrogel is a proprietary Zerdine formulation (CIRS Inc., Norfolk, VA, USA) that was originally designed to simulate a full bladder in ultrasonic imaging phantoms; it was chosen for the study because its MR imaging properties provide an excellent match to water [147], and phantoms consisting of this material are referred to as ‘gel-water’ phantoms. In the current study, these phantoms were investigated for suitability regarding accurate dosimetry when teamed with the M512 and visibility on MRI. $Z_{eff}$ and the relative electron density ($\rho_{e}^{w}$) of gel-water were determined in comparison to water and other common materials used for radiotherapy phantoms, using a method detailed in [148].

This chapter comprises initial studies for dosimaging. At this stage of the project, the MRI-linac was not ready for use, and these experiments were completed separately on a clinical MRI scanner and linear accelerator. MR safety and radiofrequency effects on the M512 and its DAQ were also necessarily investigated prior to any further experiments.
3.2 MR Safety, Radiofrequency Effects and Data Acquisition System

3.2.1 Methodology

3.2.1.1 Modification of a Data Acquisition System for MRI Safety

The M512 DAQ system is comprised of four general components: the detector, the adaptor board, AFE boards and the FPGA (Figure 3.1). The M512 is connected via three pin connectors to an adaptor board, and this board is joined to the four AFE boards through eight 64 pin connectors. Four digital cables then join the AFE boards to the FPGA. The FPGA is connected via USB 2.0 interface. The software was developed in-house.

The detector and electronics originally contained some ferromagnetic materials. These were made MRI-safe before their introduction into an MRI scanner. All screws were replaced with nylon screws. Ferromagnetic components that were not able to be replaced without a complete redesign of the system were assessed for the amount to which they would be affected in a magnetic field. This included the aluminium (Al) wire bonding of the detector to the PCB, whereby a soft gold finish on the PCB is necessary to make an effective electrical contact. Soft gold requires a small amount of nickel, which is a ferromagnetic material. The pin connectors were also considered. These consist of an Al
alloy, with a section containing tin.

At Liverpool Cancer Therapy Centre (LCTC), Liverpool, New South Wales, the M512 was introduced into the magnetic fringe field of a Siemens 3 T Wide Bore Skyra MRI scanner (Siemens, Munich, Germany). The 1 T position was located according to a Siemens Magnetom Skyra magnetic field map [149] to be just outside the bore of the scanner. The detector was positioned at the 1 T position, and the adaptor board and AFE boards were attached. The magnetic field strength at the position of the AFE boards was approximately 0.7 T. The FPGA was placed at a greater distance from the bore depending on digital cable length.

### 3.2.1.2 Radiofrequency Effects and Image Quality

The extent to which the RF noise from the RF coils of a 3 T MRI scanner affected the functionality of the DAQ was then investigated. The DAQ was positioned in the MRI, as described above. A cylindrical water phantom (Siemens, Munich, Germany) was positioned at the imaging isocentre in the bore of the MRI to simultaneously investigate the effect of image noise due to the DAQ. MR images of the water phantom were taken while the M512 was acquiring a reading. RF noise was expected to cause interference with the DAQ, and the DAQ was expected to introduce noise to the MR images.

The whole detector system - except for the digital cables and FPGA - was shielded by a large aluminium case, 2 mm in thickness. The 2 mm thickness allowed for 99% shielding from RF noise, which is approximately 50 MHz, and was previously calculated to shield the detector system from the RF noise generated by the Calypso RF tracking system [34]. The experiment was first completed without digital cable shielding, for both 30 cm cables and 1 m cables, and then repeated with cable shielding for the 1 m cables. The cable shielding comprised of grounded Al tape, encasing each cable completely. The
3.2.2 Results

3.2.2.1 Modification of a Data Acquisition System for MRI Safety

When moving the DAQ components slowly to the 1 T position of the 3 T MRI scanner, none of these components experienced the magnetic force, despite the small amount of ferromagnetic materials present. Hence, the DAQ was deemed to be MRI-safe at 1 T, which is the planned magnetic field strength of the MRI-linac.

3.2.2.2 Radiofrequency Effects and Image Quality

The electronic readout desisted to function after approximately 3 seconds when 1 m unshielded digital cables were utilised. For the 30 cm unshielded cables, however, the DAQ functioned correctly. When the 1 m cables were then shielded, the DAQ functioned normally.
Figure 3.3 MR image of water phantom, DAQ switched off.

Image noise due to the DAQ could only be evaluated when the cables were shielded because the detector did not work when the cables were not shielded. No difference in image noise was observed by qualitative evaluation between acquisitions with the DAQ switched first off then on (Figures 3.3 and 3.4). For a quantitative assessment, 1D profiles were taken through the centre of the water phantom MR images. It can be seen in Figures 3.5 and 3.6 that the difference in profiles is small. A subtraction of the reference image from the image taken during detector acquisition resulted in an average difference of ±1.1%.

3.2.3 Discussion

The DAQ was made to be MR safe, and those components that could not be readily changed were found to be MR safe at 1 T. The results demonstrated that shielding was required to enable the DAQ to function, depending on the
Figure 3.4 MR image of water phantom, DAQ acquiring reading during imaging.

Figure 3.5 1D profile through centre of MR image of water phantom; effect of DAQ on image noise.
cable length. The wavelength of the RF from the scanner allowed information to be conveyed to the electronics from the detector at the 30 cm cable length without interference, but not at 1 m. Hence, shielding is required so that cables could potentially be made at any length, depending on what is later needed on the MRI-linac.

### 3.3 Fiducial Tracking During MRI Acquisition

#### 3.3.1 Methodology

For dosimaging to be possible, the detector needs to be MRI visible. However, the detector could not be visualised on MRI at this stage without an appropriate phantom (a phantom was acquired and investigated in the next section). Hence, to ensure that its position could be tracked during a dynamic MRI acquisition, fiducial markers were attached to the detector (as shown in Figure 3.7). The
dynamic position of these markers was observed with MR imaging.

For MRI safety, the functioning M512 was not utilised for this experiment as it would have to be placed in the centre of the MRI bore at the 3 T position. A functioning detector was not necessary for this experiment as only imaging of the fiducials was required. To reduce risk to the detector system, a non-functioning silicon array detector was utilised instead. A cylindrical water phantom was utilised to give adequate signal for fiducial marker visualisation on the MR images. The detector was attached to the phantom, as displayed in Figure 3.8.

A True Fast Imaging with Steady State Precession (TrueFISP) MRI acquisition was the sequence selected. This sequence has a relatively high temporal resolution [150] and is the standard sequence utilised for lung imaging at LCTC. It is also the sequence employed by the MRIdian MRI-linac system [151]. The phantom and detector were manually moved inside the bore during the acquisition.
3.3.2 Results/Discussion

Figure 3.9 demonstrates that the fiducial markers were able to be visualised during a TrueFISP sequence while undergoing a breathing-like motion. If the detector and/or fiducial markers are required to be MRI visible for dynamic dosimaging, then an adequate phantom for both imaging and dosimetry is needed.
3.4 MRI Visualisation with Gel-Water Phantoms

3.4.1 Methodology

A phantom for the MRI-linac not only requires a material that produces a strong enough signal for MR imaging, it must also be tissue-equivalent and will not damage the detector, and since the M512 is not waterproof, a water phantom or standard hydrogel phantom could not be utilised. A CIRS bladder hydrogel phantom (called gel-water in Figure 3.10) was selected due to its elemental composition (see Table 3.1) for MRI signal, containment to avoid damaging the detector, and tissue-equivalence (this was tested in subsequent experiments).

Three gel-water phantoms were utilised: one was 1.5 cm thick and two were 10 cm thick, while the cross section of the hydrogel phantom was 20.0 × 20.2 cm². The hydrogel was surrounded by side walls (2.5 cm thick) constructed from acrylonitrile butadiene styrene (ABS) and by a thin saran wrap 560 plastic film (DOW Chemical USA, Midland, MI, USA) that covered the top and bottom surfaces. The total area of all the phantoms (including the walls) was 22.5 × 22.7 cm².

This investigation aimed at determining various properties of the gel-water phantoms and ABS phantom walls such as the mass density ($\rho$), $\rho_{\text{w}}^w$ and $Z_{\text{eff}}$, and to image these phantoms with both MRI and CT. The MR images were used to determine whether impurities such as bubbling were present, and whether the detectors could be visualised adequately with the signal from these phantoms. A CT scan was also utilised to measure the CT number and hence the $\rho_{\text{w}}^w$ of the gel-water and ABS, and these results were compared with the calculated values.
3.4.1.1 Relative Electron Density and Effective Atomic Number Calculations

The $\rho_w$ and $Z_{\text{eff}}$ of gel-water were calculated using $\rho$, where $\rho_w$ is the electron density of a substance ($\rho_{\text{es}}$) normalised to the electron density of water ($\rho_{\text{ew}}$). For the purpose of comparison, the $\rho_w$ and $Z_{\text{eff}}$ were also calculated for water, solid water and polymethyl methacrylate (PMMA), the elemental compositions of which are shown in Table 3.1. The electron density was calculated by the equation [5,152]:

$$\rho_w = \frac{N_A \rho}{\rho_{\text{ew}}} \left[ \sum_{i=1}^{N} f_i \left( \frac{Z_i}{A_i} \right) \right]_s$$  \hspace{1cm} (3.1)

where $N_A$ is Avogadro’s number = $6.023 \times 10^{23}$ mol$^{-1}$, $\rho$ is the density of the substance ($s$), $\sum_{j=1}^{N}$ is the sum over all elements composing $s$, $f_i$ is the fractional mass of an element composing $s$, and $\left( \frac{Z_i}{A_i} \right)_i$ is the ratio of atomic number ($Z$) to atomic mass ($A$). $Z_{\text{eff}}$ was calculated from the following relation [5]:

$$Z_{\text{eff}} = \left[ \sum_{i=1}^{N} \alpha_i (Z_i)^a \right]^{\frac{1}{a}}$$  \hspace{1cm} (3.2)
Table 3.1
Elemental composition of gel-water (CIRS Inc., Norfolk, VA, USA) in comparison to various substances [5–7].

<table>
<thead>
<tr>
<th>Elemental Composition</th>
<th>Water</th>
<th>Gel-Water</th>
<th>Solid Water</th>
<th>PMMA</th>
</tr>
</thead>
<tbody>
<tr>
<td>H</td>
<td>12.0</td>
<td>10.7</td>
<td>8.09</td>
<td>8.05</td>
</tr>
<tr>
<td>C</td>
<td>4.20</td>
<td>67.22</td>
<td>59.98</td>
<td></td>
</tr>
<tr>
<td>N</td>
<td>1.40</td>
<td>2.40</td>
<td></td>
<td></td>
</tr>
<tr>
<td>O</td>
<td>88.0</td>
<td>83.6</td>
<td>19.84</td>
<td>31.96</td>
</tr>
<tr>
<td>S</td>
<td>0.10</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ca</td>
<td></td>
<td></td>
<td></td>
<td>2.32</td>
</tr>
<tr>
<td>Cl</td>
<td></td>
<td></td>
<td></td>
<td>0.13</td>
</tr>
<tr>
<td><strong>Total Density</strong></td>
<td><strong>1.00</strong></td>
<td><strong>1.01</strong></td>
<td><strong>1.04</strong></td>
<td><strong>1.19</strong></td>
</tr>
</tbody>
</table>

$\rho (\text{g/cm}^3)$:

where $a = 2.94$, which accounts for the photoelectric effect present at the energies encountered in CT. $\alpha_i$ can be determined from [5,148]:

$$\alpha_i = \frac{f_i \left( \frac{Z}{A} \right)_i}{\sum_{j=1}^{N} \left( \frac{Z}{A} \right)_j}$$

(3.3)

### 3.4.1.2 Mass Density Measurement

The $\rho$ of gel-water was then measured experimentally. The phantoms were weighed and their volumes measured to calculate $\rho$. Since this gel-water could not be separated from the ABS, the mass of the ABS was subtracted from the mass of the whole phantom by measuring its volume and taking $\rho$ as reported by the manufacturer ($1.040 \text{ g/cm}^3 [153]$). The mass of the saran wrap 560 film was not accounted for due to its negligible mass (thickness = 152 µm, density...
= 1.80-1.94 g/cm³ depending on the sample).

3.4.1.3 CT Verification of Electron Density

CT scans of the gel-water were completed to verify the electron density calculations. Imaging was carried out with a Philips Brilliance Big Bore CT scanner at 120 kV and 300 mAs per slice. A head and neck scan protocol was chosen as the gel-water was small, and 2 mm thick slices were utilised. Each phantom was imaged separately, and depending on its size, between 10 and 20 CT numbers were measured from each scan. CT-density conversion data was acquired from the Pinnacle TPS for the Philips CT scanner. The gradient of the regression line ($R_\gamma$) of the CT density conversion data was determined by plotting this data and fitting a regression line. $\rho_e^w$ was determined by [5]:

$$\rho_e^w = R(\frac{1}{1000}N_{CT}) + 1$$ (3.4)

where $N_{CT} = \text{CT number} + 1000$ (to ensure integers are unsigned to conserve memory in the Pinnacle TPS [5]).

The error for $R_\gamma$ was determined by the Linest function in Microsoft Excel and the error in $N_{CT}$ was determined as the 95% confidence limit of $N_{CT}$. These uncertainties were combined to find the error in $\rho_e^w$ that was determined by substituting the average $N_{CT}$ into the regression line equation.

3.4.1.4 Gel-Water MR Images

MR images were acquired at 3 T of the gel-water phantoms with and without a silicon array detector on the top surface of the phantom to determine the homogeneity of the gel-water, as well as detector visualisation. The three phantoms were imaged with each phantom positioned in the centre of the MRI bore Figure 3.11). One large phantom was imaged by itself and then with the detector, while the other large phantom and the 1.5 cm thick phantom were
Figure 3.11 Stacked gel-water phantoms (of 1.5 cm thickness and 10 cm thickness) inside coil for MR imaging.

imaged with the smaller phantom atop the larger. A T2 Fast Spin Echo (FSE) sequence was utilised to reduce any image artefacts induced by the detector. Each slice was 3 mm thick.

3.4.2 Results

3.4.2.1 Relative Electron Density and Effective Atomic Number Calculations

The values calculated for $\rho^w_e$ and $Z_{eff}$ are shown in Table 3.2, and indicate that the gel-water is closer in $\rho^w_e$ and $Z_{eff}$ to water than what solid water or PMMA are to water.

3.4.2.2 Mass Density Measurement

The $\rho$ of the gel-water was measured to be $1.06 \pm 0.07$ g/cm$^3$, which is within experimental uncertainty of the value given by the manufacturer (1.007 g/cm$^3$ [153]). The uncertainty was determined by combining the uncertainty of both
Figure 3.12 Gel-water phantom with silicon detector.

Table 3.2
Calculated values for relative electron density ($\rho_w^e$) and effective atomic number ($Z_{eff}$) utilising Equation 3.1 to 3.3.

<table>
<thead>
<tr>
<th>Substance</th>
<th>$\rho_w^e$</th>
<th>$Z_{eff}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water</td>
<td>1.00</td>
<td>7.38</td>
</tr>
<tr>
<td>Gel-Water</td>
<td>1.01</td>
<td>7.39</td>
</tr>
<tr>
<td>Solid Water</td>
<td>1.02</td>
<td>7.54</td>
</tr>
<tr>
<td>PMMA</td>
<td>1.16</td>
<td>7.55</td>
</tr>
</tbody>
</table>
mass and volume. This $\rho$ value gives a $\rho_w^e$ value of $1.06 \pm 0.07$, when it is rearranged and substituted into Equations 3.1-3.3.

### 3.4.2.3 CT Verification of Electron Density

From the CT scan data, $\rho_w^e$ was determined to be $1.001 \pm 0.042$ for the gel-water and $1.001 \pm 0.042$ for the ABS, while $\rho$ was $1.005 \pm 0.114$ g/cm$^3$ for the gel-water and $0.957 \pm 0.114$ g/cm$^3$ for the ABS.

### 3.4.2.4 Gel-Water MR Images

The images were assessed qualitatively for any impurities in the gel-water which appeared to be homogeneous, with only a small number of bubbles present (Figure 3.13). The bubbles were approximately 2 mm or less, and were obtained by measuring the scale on the MR images and then measuring each bubble. The 1.5 cm thick phantom contained no observable bubbles, one of the 10 cm thick phantoms contained 4 (referred to as phantom 1 in the dosimetric results in Section 3.5), and the other 10 cm thick phantom (referred to as phantom 2) contained 20 bubbles. The bubbles are not concentrated in any one particular position within the phantoms but are distributed randomly. Furthermore, small sections of the outer walls of the phantoms had been hollowed out, most likely due to the hydrogel inside the phantom being filled during construction (Figure 3.15). However, even if the largest hollowed out section (approximately $30 \times 28 \times 12$ mm$^3$) is considered, its effect on the electron density calculations was negligible. The detector was observed on the MR image due to artefacts that the copper tracks on the PCB produced in the phantom (Figures 3.12 to 3.16).

### 3.4.3 Discussion

The measured and CT-determined value for $\rho_w^e$ for the gel-water phantom of $1.06 \pm 0.07$ and $1.001 \pm 0.042$ respectively, were both within the experimental error range of the calculated 1.01 value, and the gel-water provided an adequate signal in the MRI. In fact, the images showed a resolution that allowed the gel-
Figure 3.13 MR image of stacked gel-water phantoms (of 1.5 cm thickness and 10 cm thickness); the larger phantom displaying a small bubble (<2 mm in diameter) indicated by the arrow.

Figure 3.14 MR image of 10 cm thick gel-water phantom displaying hollowed out section of side wall.
Figure 3.15 2D MR image of 10 cm thick gel-water phantom with artefact due to detector positioned on top surface.

Figure 3.16 3D MR image of 10 cm thick gel-water phantom with artefact due to detector positioned on top surface.
When gel-water is positioned on either side of a detector, it will be observable (if unshielded), but if the detector requires aluminium shielding from the electromagnetic field of radiofrequencies from the MRI-linac, then anything inside the shielding will be invisible on MRI. So, if the shielding is fitted to the detector properly, it will be observable where the detector is absent from the MR image.

Despite the presence of bubbles in the two larger phantoms they appeared to be mostly homogeneous. These phantoms are not cast or poured by traditional methods so they are susceptible to the occasional formation of tiny air bubbles. One source of uncertainty was that the gel-water phantoms produced slightly different results so their compositions would differ slightly and cause this discrepancy. However, this effect was negligible when the gel-water is compared to solid water, and it did not correlate with the number of bubbles as one phantom did not give consistently higher or lower results than the other. It is therefore recommended that each new gel-water phantom be characterised after manufacture. The characterisation could be done as a quality assurance exercise prior to shipment if the manufacturer has access to MRI (as is done with solid water with CT and other methods).

### 3.5 Tissue-Equivalence Comparison Between Gel-Water and Solid Water

A tissue-equivalency of gel-water is needed for dosimetry, so a depth dose, tissue phantom ratio (20/10) (TPR\textsubscript{20/10}) and a tissue maximum ratio (TMR) comparison between the gel-water and solid water were measured on a standard 6 MV linac. A comparison between the M512 and RMI Attix chamber
(serial no: 1076) was also made to measure the percentage depth dose in solid water. Beam profiles, output factors and gamma analysis for the M512 data were carried out in both materials.

3.5.1 Methodology

The dose for the linac measurements was delivered by a 6 MV Varian 2100C Linac at a dose rate of 600 cGy/min, and 100 MU were delivered in every case. The Attix chamber was set to a bias of −300 V and a PTW Unidos Webline electrometer (serial no: 000228) was utilised for the charge readout. Chamber polarity was then reversed for repeated surface and \( d_{\text{max}} \) measurements, to observe how the charge was affected by polarity at these depths.

The M512 was set to the least sensitive gain setting where a suitable signal could be obtained for sampling. The frequency was adjusted to 360 Hz to match the frequency of the linac and synchronised with the electron gun firing, with an integration time of 52 \( \mu \text{s} \). An equalisation procedure was carried out first to determine a calibration factor for the array. Each pixel has a slightly different response due to its individual sensitivity and preamplifier gain. The equalisation corrects for the individual response of each diode to allow for an equalised response over the whole array [29, 133]. This procedure was performed at 100 cm SSD with a \( 20 \times 20 \text{ cm}^2 \) field size and at a depth of 10 cm in solid water, where the profile is approximately flat. The equalisation factor vector is determined by dividing the acquired data by the average response from all pixels. These equalisation factors are then applied to the measured data. For percentage depth dose, TPR and TMR measurements, the Attix chamber and M512 measurements were compared to standard linac commissioning data for that particular linac, utilising a CC13 ionisation chamber and water tank.
3.5.1.1 Percentage Depth Dose

To measure the percentage depth dose, a $10 \times 10 \text{ cm}^2$ field size was set and measurements taken with the M512 and Attix chamber. A source-to-surface distance (SSD) of 100 cm was set. For measurements in solid water with the Attix chamber, it was first positioned at the surface of the phantom and then at subsequent build-up and fall-off depths. 20 cm thick solid water was inserted for backscatter material. The gel-water measurements were completed at depths of 15, 100 and 200 mm as they were the only depths available. The backscatter material was solid water, as the Attix chamber had to be placed into its solid water holder.

For depth dose measurements utilising the M512, an extra 5 mm of PMMA had to be accounted for when measuring depth due to the PMMA casing surrounding the detector. This was inclusive of a 1.2 mm airgap directly above the silicon detector, as shown in Figure 3.17. Gel-water measurements were completed at depths of 15, 100 and 200 mm of actual gel-water, but the extra 5 mm of PMMA meant that the total depths were actually 20, 105 and 205 mm. For this reason, measurements in solid water were taken at depths of 15, 20, 100, 105, 200 and 205 mm to cover all potential comparisons (against both gel-water and Attix chamber). A density correction was not used for the 5 mm of PMMA as this would result in $<1$ mm difference (i.e. $5 \text{ mm} \times 1.16 = 5.8 \text{ mm}$). At depths of 10.5 cm and 10.6 cm for a $10 \times 10 \text{ cm}^2$ size field, this would result in approximately 0.3% difference between these depths. This difference increases as the size of the field decreases, but there is only approximately 0.4% difference for a $3 \times 3 \text{ cm}^2$ field. This difference is less than the uncertainty measured for the M512, which will be discussed in the results section.

As Figure 3.18 shows, the linac and couch were both positioned at $270^\circ$ due to the gel-water bowing out slightly when positioned on top of the Attix
chamber in a standard set-up at 0, and the compression of the chamber volume leads to a change in the chamber reading. All the depth dose measurements were normalised to the dose at dmax for 6 MV in solid water for corresponding field sizes, while uncertainty was calculated as the normalised 95% confidence level of three repeated measurements.

3.5.1.2 TPR\textsubscript{20/10} and TMR

Although Attix chambers are not generally utilised for TPR and TMR measurements, no chamber could be inserted into the gel-water so the Attix chamber was utilised because its effective point of measurement could be positioned closer to the gel-water than other chambers due to its thin entrance window. Other chambers would need to have a position correction. The TPR\textsubscript{20/10} and TMR measurements were completed at a standard field size of 10 × 10 cm\textsuperscript{2} at the isocentre and the source-to-axis (SAD) distance was set at 100 cm. For TPR, the dose at a depth of 20 cm (D\textsubscript{20}) was compared to the dose at 10 cm depth (D\textsubscript{10}) using the equation:
Figure 3.18 Experimental set-up: Gel-water sample positioned on its side in front of Attix chamber due to bowing of gel water onto chamber in standard orientation, with solid water as backscatter inside solid water holder. Purpose of the extra solid water beneath the gel-water was to raise it to correct height. Linac and couch were positioned at 270°.
\[ TPR_{20/10} = \frac{D_{20}}{D_{10}} \] (3.5)

For TMR, the dose at the reference depth of 10 cm was compared to the dose at \( d_{\text{max}} \) \( (D_{1.5}) \), shown by the equation:

\[ TMR = \frac{D_{10}}{D_{1.5}} \] (3.6)

The gel-water, solid water and standard CC13 water tank data were compared, and the 95% confidence limit uncertainties from each measurement were combined to determine the final uncertainty.

### 3.5.1.3 Profiles

Profiles were taken across the M512 data in the x- and y-directions to compare gel-water and solid water for jaw-defined jaw-defined field sizes ranging from 0.5 \( \times \) 0.5 cm\(^2\) up to 4 \( \times \) 4 cm\(^2\). The SAD was 100 cm and the profiles were measured at a depth of 10.5 cm. Again, a depth of 10.5 cm was required rather than 10 cm due to the reasons stated above. The profile data was normalised to the central pixel, and the normalised 95% confidence limit of three acquisitions was taken as the uncertainty for each pixel. The full width at half maximum (FWHM) was measured for each profile by utilising the shape-preserving interpolant fit in MATLAB.

### 3.5.1.4 Output Factors

Output factors were acquired with the M512 for field sizes ranging from 0.5 \( \times \) 0.5 cm\(^2\) up to 10 \( \times \) 10 cm\(^2\) (jaw-defined) and normalised to the 10 \( \times \) 10 cm\(^2\) reading; these measurements were performed in gel-water and solid water for comparison. These factors were measured at a depth of 10.5 cm and with a SAD of 100 cm, and the normalised 95% confidence limit from three measurements was the uncertainty.
3.5.1.5 Gamma Analysis

Gamma analysis was completed for various field shapes using the M512 to investigate, in particular, differences in the penumbral region between solid water and gel-water. The measured fields were a $3 \times 3 \text{ cm}^2$ MLC-defined square field, $3 \times 3 \text{ cm}^2$ jaw-defined square field, $3 \text{ cm}$ diameter MLC-defined circle field, and two dynamic fields: one sliding window and one pyramid-shaped. Jaw sizes for the MLC fields were: $5 \times 5 \text{ cm}^2$ for the square and circle fields and $10 \times 10 \text{ cm}^2$ for the two dynamic fields. The passing criteria examined was $3\%/3 \text{ mm}$, and then further reduced to $3\%/2 \text{ mm}$ and $2\%/2 \text{ mm}$.

A gamma analysis of the various field shapes was carried out using the M512 to investigate any differences in the penumbral region between solid water and gel-water. The fields consisted of a $3 \times 3 \text{ cm}^2$ MLC-defined square field, a $3 \times 3 \text{ cm}^2$ jaw-defined square field, a $3 \text{ cm}$ diameter MLC-defined circle field, and two dynamic fields: one sliding window and one pyramid-shaped. The jaw widths for the MLC fields were $5 \times 5 \text{ cm}^2$ for the square and circular fields and $10 \times 10 \text{ cm}^2$ for the two dynamic fields. A global gamma analysis was utilised and the passing criteria examined were $3\%/3 \text{ mm}$, which were then reduced to $3\%/2 \text{ mm}$ and $2\%/2 \text{ mm}$. Pixels that were $<20\%$ of the dose were excluded from this analysis; this not only included the out-of-field dose, it also included any dead pixels present in the prototype device.

3.5.2 Results

3.5.2.1 Percentage Depth Dose

The depth dose measurements are shown in Figures 3.20 and 3.20. The maximum differences found between materials over all measured depths are shown in Table 3.3. The difference in the Attix chamber and M512 data could only be determined for solid water because the data were not available at corresponding depths in the gel-water. The maximum discrepancy between
Figure 3.19 Percentage depth dose comparison for gel-water, solid water and water using Attix chamber, M512 and CC13 chamber. Gel-water and solid water measurements normalised to 1.5 cm depth in solid water for a $10 \times 10 \text{ cm}^2$ field size at 6 MV. Water measurements normalised to 1.5 cm depth in water.

detectors over all depths measured was 1.2% for the $10 \times 10 \text{ cm}^2$ field size and 0.7% for the $20 \times 20 \text{ cm}^2$ field size. The uncertainty was small because both detectors were very stable when acquiring data; the Attix chamber data had a maximum error of $\pm 0.07\%$ and the M512 had a maximum of $\pm 1.0\%$ error.

After the polarity in the chamber changed to CEP, the $d_{\text{max}}$ dose displayed almost no difference to CEN. However, the surface measurement differed by 3.5%, so a polarity correction was made for the surface dose measurement. Differences in the polarity of the Attix chamber for surface measurements were reported previously [98, 101].
Figure 3.20 Percentage depth dose comparison for gel-water, solid water and water using Attix chamber, M512 and CC13 chamber. Gel-water and solid water measurements normalised to 1.5 cm depth in solid water for a $20 \times 20 \text{ cm}^2$ field size at 6 MV. Water measurements normalised to 1.5 cm depth in water.
Table 3.3
Percentage depth dose maximum differences between materials over all depths measured.

<table>
<thead>
<tr>
<th>Detector Type</th>
<th>Comparison Between</th>
<th>Difference (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Phantom Type</td>
<td>Phantom Type</td>
</tr>
<tr>
<td>Attix Chamber</td>
<td>Gel-water</td>
<td>Solid water</td>
</tr>
<tr>
<td></td>
<td>Solid water</td>
<td>Water tank</td>
</tr>
<tr>
<td></td>
<td>Water tank</td>
<td>Gel-water</td>
</tr>
<tr>
<td>M512</td>
<td>Gel-water</td>
<td>Solid water</td>
</tr>
<tr>
<td></td>
<td>Solid water</td>
<td>Water tank</td>
</tr>
<tr>
<td></td>
<td>Water tank</td>
<td>Gel-water</td>
</tr>
</tbody>
</table>

3.5.2.2 TPR_{20/10} and TMR

The TPR_{20/10} and TMR results are displayed in Table 3.4. The value measured for TPR_{20/10} by the CC13 chamber in water was 0.675, although the Attix chamber measurement in solid water differed from the value of the CC13 water tank by 1.3%. Phantom 1 of the gel-water differed by 1.4%, and phantom 2 by 2.1%. The gel-water phantom 1 TPR value agrees with the solid water TPR within experimental error, while phantom 2 deviates from the solid water by 0.8%.

The TMR value given by the CC13 data was 0.770, but the Attix chamber measurement in solid water differed by 0.4%, and the gel-water measurements differ from the standard data by 1.0 and 0.2% for phantom 1 and 2, respectively. The gel-water phantom 1 differed from the solid water measurement by 1.4% and phantom 2 by 0.6%.

3.5.2.3 Profiles

Profiles were taken across the M512 data in the x- and y-directions and the FWHM was measured for each. Examples of profiles across a 0.5 × 0.5 cm² and
Table 3.4
Tissue phantom ratio 20/10 (TPR\textsubscript{20/10}) and tissue maximum ratio (TMR) values measured at 6 MV with Attix chamber.

<table>
<thead>
<tr>
<th>Substance</th>
<th>TPR\textsubscript{20/10}</th>
<th>TMR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Solid Water</td>
<td>0.662 ± 0.001</td>
<td>0.774 ± 0.002</td>
</tr>
<tr>
<td>Gel-Water (Phantom 1)</td>
<td>0.661 ± 0.001</td>
<td>0.760 ± 0.001</td>
</tr>
<tr>
<td>Gel-Water (Phantom 2)</td>
<td>0.654 ± 0.001</td>
<td>0.768 ± 0.001</td>
</tr>
</tbody>
</table>

Table 3.5
Difference between FWHM of solid and gel-water profiles measured using the M512.

<table>
<thead>
<tr>
<th></th>
<th>Difference Between Solid and Gel-Water FWHM (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>X Profiles</td>
</tr>
<tr>
<td></td>
<td>Phantom 1</td>
</tr>
<tr>
<td>Field Size (cm(^2))</td>
<td>0.5 × 0.5</td>
</tr>
<tr>
<td></td>
<td>1 × 1</td>
</tr>
<tr>
<td></td>
<td>2 × 2</td>
</tr>
<tr>
<td></td>
<td>3 × 3</td>
</tr>
<tr>
<td></td>
<td>4 × 4</td>
</tr>
</tbody>
</table>

1 × 1 cm\(^2\) are shown in Figure 3.21. The differences in FWHM between the gel-water and solid water for each phantom are shown in Table 3.5; the table indicates that the average difference between the FWHM of each material over every size field and in the x- and y-directions was 0.1 mm, and the maximum overall FWHM deviation was 0.3 mm.

3.5.2.4 Output Factors

The output factors (Figure 3.22) taken with the M512 demonstrated excellent agreement between solid and gel-water; gel-water phantom 1 demonstrated a maximum discrepancy of 1.4% from the solid water output factors, and phantom 2 a discrepancy of 0.6%.
Experiments with Gel-Water: Towards MRI-Linac Dosimetry and Imaging

Figure 3.21 Profiles across M512 comparing gel-water and solid water (a)-(c) 0.5 × 0.5 cm² and (d)-(f) 1 × 1 cm². (a), (d) are the beam’s eye view as displayed on the M512 readout software. All profile data normalised to central pixel. Dead/noisy pixels are visible on prototype detector.

3.5.2.5 Gamma Analysis

The results of the gamma analysis are shown in Figure 3.23 and 3.24. Here the jaw-defined square field had a gamma pass rate of 100% within 3%/2 mm, while all the other fields were MLC-defined and had a pass rate of 100% within 2%/2 mm. Whereas penumbral regions have greater discrepancies than in-field regions, they were still within tolerance (i.e. gamma value ≤1), but the penumbral region of the square jaw-defined field failed when the tolerance was reduced to 2%/2 mm.

3.5.3 Discussion

The maximum overall deviation in the percentage depth dose measurements between gel-water and solid water was 2.8% for the Attix chamber, while the largest discrepancy between materials for the M512 was 1.1%, and the largest difference between the Attix and M512 was 1.2% in solid water; in fact most
Figure 3.22 Output factors using central pixel of the M512, comparing gel-water and solid water. All measurements normalised to $10 \times 10$ cm$^2$ field size in solid water for a 6 MV beam.
Figure 3.23 (a)-(c) Beam’s eye view of static field shapes as acquired by the M512, (d)-(f) gamma analysis between gel-water and solid water M512 acquisitions displaying pass rates of 100% within the criteria specified. The jaw-defined field in (a), (d) had a pass rate of 100%.

Figure 3.24 (a)-(b) Beam’s eye view of dynamic field shapes (MLC-defined) as acquired by the M512, (c)-(d) gamma analysis between gel-water and solid water M512 acquisitions displaying pass rates of 100% within 2%/2 mm. Dead/noisy pixels are visible on prototype detector.
differences were much less than these values. There was a maximum deviation of 2.1 and 1.0% for the TPR and TMR measurements between the gel-water (with Attix chamber) and water tank data (CC13). The largest deviation between the FWHM of small radiation fields in gel-water and solid water was a minimal 0.3 mm. The output factors between materials also fell closely, with 1.3% being the greatest deviation. All the field shapes passed gamma analysis with 100% of the data within 3%/2 mm, and all the MLC-defined fields passed to within 2%/2 mm. The failure of the penumbra in the jaw-defined field when the tolerance is reduced to 2%/2 mm is because the penumbra of the jaw-defined field is sharper than an MLC-defined field. A region like this with a higher dose gradient is more likely to display a discrepancy over a region with a lower dose gradient. The Attix chamber and M512 were both stable and hence the experimental error was small.

### 3.6 Conclusion

This chapter first investigated the effect of radiofrequency on the DAQ of the M512, and then conversely investigated the effect of the DAQ on the MR images. Aluminium-shielded digital cables allowed the DAQ to function properly in a radiofrequency field, and the M512 was found to be MRI-safe in a 1 T magnetic field as it experienced no effect due to the magnetic force. The electronics of the M512 did not have any significant effect on the image quality of the MR images acquired. Tracking of fiducial markers attached to the detector was found to be possible utilising a TrueFISP MRI sequence.

It was concluded that the imaging hydrogel ‘gel-water’, is a feasible option to replace solid water in an MRI-linac, especially for dynamic dosimaging. The system with gel-water and the M512 had a discrepancy from solid water of $\leq 1.1\%$ for percentage depth dose measurements, $\leq 0.3$ mm for profiles, and $\leq 1.3\%$ for output factors. A comparison by gamma analysis between these two
materials resulted in a 100% pass rate to within 2%/2 mm for MLC-defined fields. These results are within clinical tolerances. The current MRI-guided dynamic dosimaging set up is successful in detector visualisation and tissue-equivalence, but it is recommended that new gel-water be characterised after manufacture. This initial study will first lead into static dosimaging on the Australian MRI-linac within the timeframe of this thesis. Following this thesis, future studies can be completed with an MRI-compatible motion platform paired with the M512 to simulate the intrafraction motion of a patient in 4D.
Chapter 4

Penumbral Width Trimming in Solid Lung Profiles for MRI-Linac Prototypes

Sections of this chapter are presented in the publication:


4.1 Introduction

Longitudinal magnetic fields have a focusing effect on electrons and may narrow the penumbra [78,79,81]; this can tighten the lateral spread of secondary electrons in air cavities, including lung tissue [79]. Quantification of the focusing effect on the beam penumbra is important, especially in air cavities, for a progression towards MRI-linac clinical treatments. In this study, profiles on Gafchromic EBT3 film were used to investigate differences in penumbra between Gammex-rmi solid water (Gammex Inc., Middleton, WI, USA) and CIRS solid lung (CIRS Inc., Norfolk, VA, USA), and under 0, 0.9, and 1.5 T conditions.
using both the first and second Australian MRI-linac prototype configurations. Comparisons were also made between the entrance water/lung interface, mid-depth, and exit lung/water interface dose profiles.

4.2 Methodology

The linatron of the Australian MRI-linac was described in Section 2.1.2. Further details of the MRI-linac will now be provided.

The linatron was fitted with a circular aperture collimator and has a Varian Millennium 120 multileaf collimator (MLC) separated from the linatron head. The linatron and MLC were attached to a rail system to allow for a variable source-to-surface distance (SSD); the purpose of this was so that the position of electron gun, waveguide, target, and MLCs in the magnetic field could be altered to determine the optimum position for normal functioning of all components, including the MRI. For the Sonata magnet (Figure 4.1), this position was determined to be that providing a distance of 276.9 cm from the source to the centre of the bore. For the Agilent magnet (Figure 4.2), the fringe field was much larger, and therefore the linatron had to be moved to the rail position furthest from the magnet, as the electrons in the waveguide were affected if moved closer. However, the same SSD needed to be maintained for the experiment, so in this case, the experiment was not performed at the centre of the magnet. The magnetic field at this position was measured with a Gaussmeter to be 0.9 T. The field uniformity was measured to be ±4.05 ppm at the surface of a 30 cm DSV sphere and projected to a 20 cm DSV sphere to be ±0.47 ppm. Magnetic field maps of the Agilent magnet have been previously published [154,155], which show that the field is homogenous at this location.

Due to changes in the set-up between the measurements, a copper sheet required for the Faraday cage (Figure 4.3 was in position for measurements
in the Sonata magnet but not for measurements in the Agilent magnet. The experiment with this magnet was therefore performed with the window open. A repeat of a subset of the measurements at a depth of 10 mm was completed with the window closed, as it was expected to have an effect at this depth.

The field sizes for the measurements in 0.9 T magnetic field (second prototype) could not be matched to the 0 and 1.5 T measurements (first prototype) due to the change in source-to-MLC end distance (source-to-collimator end distance, SCD), which could not be changed. The first prototype had an SCD of 50.3 cm, and the second prototype, 52.2 cm. The impact of this difference was assessed by determining the difference in penumbra with this change in SCD.

4.2.1 Comparison of Penumbral Widths with Change in Magnetic Field

Dose measurements were completed on the MRI-linac utilising EBT3 film in solid water and solid lung, with magnetic field strengths of 0, 0.9, and 1.5 T.
**Figure 4.2** Agilent Magnet: the magnetic field component of the second prototype of the Australian MRI-Linac.

**Figure 4.3** Schematic of MRI-linac and Faraday cage (NB: SCD1, source-to-collimator end distance of first prototype; SCD2, SCD of second prototype).
The thickness of the solid lung was 50 mm, and the density was 0.30 g/cm³. Profiles were taken across the film with the aim of comparing differences in the penumbral regions between the following:

- 0, 0.9, and 1.5 T
- Solid water and solid lung
- Entrance water/lung interface, mid-depth, and exit lung/water interface profiles.

The solid water and solid lung were positioned vertically inside a Perspex holder so that it would be perpendicular to the central axis of the horizontal linatron beam (Figure 4.4 and 4.5). The phantom was set up in the centre of the MRI bore at the position of the greatest magnetic field in the Sonata magnet. In the Agilent magnet, the phantom was positioned in the magnet bore nearer to the linatron, with the front of the phantom 50 cm from the centre of the bore. The solid lung slabs were inserted between the solid water slabs to mimic a basic lung cavity and a comparison made to the set-up with solid water only. Film was positioned at the entrance and exit interface positions of solid lung, and at depths corresponding to those positions in solid water. In addition, film was placed at 20 mm depth in the solid lung (corresponding to a depth of 30 mm in solid water), to observe any differences in the penumbra at these regions. Care was taken to ensure no air gaps were present around the film to minimise magnetic field effects in air.

An equivalent dose of 15 Gy, as measured by the ionisation chamber in the head of the linatron, was delivered to each piece of film. A trigger rate of 200 Hz was set on the linatron console. Low mode (4 MV) was selected as this was the energy that had been commissioned at the time of the experiments. There was a difference in beam quality between the two energy modes of approximately 3.3 to 4.3 MV (converted to photon energies using the BJR 25 lookup table [156]).
Figure 4.4 Experimental set-up: Gafchromic EBT3 film positioned at various depths in (a) solid water, corresponding to entrance interface, 20 mm depth, and exit interface position of (b) solid lung.

The energies of 3.3 and 4.3 MV correspond to average energies of approximately 1.1 and 1.4 MeV, respectively. The continuous-slowing-down approximation (CDSA) range for electrons of these two energies in a tissue-equivalent plastic is very close; approximately $3 \times 10^4 \, \text{g/cm}^2$ and $5 \times 10^4 \, \text{g/cm}^2$, respectively, [157] and thus, the difference in refocusing of electrons due to the magnetic field and difference in penumbral shape is not large. The MLC-defined field sizes were set to $3 \times 3 \, \text{cm}^2$ and $10 \times 10 \, \text{cm}^2$ at the extended SSD of the linatron of 276.9 cm, which corresponded to $1 \times 1 \, \text{cm}^2$ and $3.2 \times 3.2 \, \text{cm}^2$ at 100 cm SSD.

For analysis, the film was scanned at a resolution of 150 dpi and the dose was determined from the net optical density using a 3rd order polynomial using the method described in Section 2.6.2. The uncertainty in the film was determined from the equation:

$$\Delta OD = \frac{1}{\ln(10)} \sqrt{\left(\frac{\sigma_{I_o}}{I_o}\right)^2 + \left(\frac{\sigma_I}{I}\right)^2} \quad (4.1)$$

where the variables $\sigma_{I_o}$ and $\sigma_I$ are statistical errors associated with $I_o$ and $I$, respectively. The profiles were normalised to the 50% field edge because an FFF
beam was used. From normalised profiles plotted against the distance from the beam central axis (CAX), 80-20% penumbral widths were determined.

\begin{figure}[h]
\centering
\includegraphics[width=0.8\textwidth]{experimental_setup.png}
\caption{Experimental set-up: (a) Solid lung slab placed at a depth of 10 mm in solid water, and (b) Gafchromic EBT3 film at entrance of solid lung.}
\end{figure}

\section*{4.2.2 Variation in Penumbral Width Over Different Magnetic Fields for Different Depths of Lung Tissue}

The experiment was then repeated utilising solid lung of thickness 20 and 30 mm. For the 0.9 T measurements, a lung slab 100 mm thick was also considered. Film was placed at a depth of 20 mm in the 100 mm thick solid lung (corresponding to a depth of 30 mm in solid water). The experimental set-up is shown in Figure 4.6.
Figure 4.6 Experimental set-up: Gafchromic EBT3 film positioned at the entrance and exit positions of solid lung of thicknesses ranging from 20 mm to 100 mm. In the 50 mm and 100 mm thick solid lung slabs, film was also placed at a depth of 20 mm (corresponding to a depth of 30 mm in solid water).

4.3 Results

4.3.1 Comparison of Penumbral Widths with Change in Magnetic Field

The profiles were aligned to the 50% field edge and displayed from -1 to +1 mm on either side of this edge for clarity (Figures 4.11 to 4.15). Profiles are only displayed of the RHS, also for clarity. The maximums of the profiles are at central axis, which are not shown in Figures 4.11 to 4.12. Figure 4.13 shows an example of a full profile in solid lung at different magnetic field strengths. The differences in 80-20% penumbral width between each magnetic field strength and zero field are displayed in Figures 4.14 to 4.15. A repetition of some measurements at 0 and 1.5 T was needed yet not possible due to the decommissioning of the Sonata magnet and the further unavailability of zero magnetic field; hence, these results have not been included. The average uncertainty overall was calculated to be $\pm 1.2\%$, with an overall maximum uncertainty of $\pm 4.7\%$. 
Figures 4.11 to 4.13 show a general trend of the 1.5 T magnetic field narrowing the penumbra to a greater extent than the 0.9 T field; the 0.9 T field resulted in slightly tighter penumbral widths than the zero field. Figures 4.14 to 4.15 display this difference numerically for both sides of the beam profile. Overall, the largest penumbral narrowing due to the 1.5 T field was $4.4 \pm 0.1$ mm, and $2.5 \pm 0.1$ mm for the 0.9 T field. These values were both for exit interface profiles. Penumbra narrowing increased with depth, which did not greatly differ between the solid lung and solid water; the maximum differences in each material between the entrance and exit doses were $2.6 \pm 0.1$ mm and $2.2 \pm 0.1$ mm respectively.

An exception to the penumbral narrowing due to the magnetic field occurred at the $10 \pm 10$ cm$^2$ field size in solid water at 0.9 T; all profiles for this configuration had penumbral widths greater than the 0 T profiles with the exception of one side of an exit profile. However, the difference in penumbral widths was $<1$ mm.

The magnetic field brought the penumbral width of the solid lung profiles closer in width to the solid water profile penumbra. At 0 T, there was a maximum difference between the penumbra of each phantom material of $2.5 \pm 0.1$ mm; this was an entrance profile. At 0.9 T, the maximum decreased, although not to a large extent, to $2.2 \pm 0.1$ mm. At 1.5 T, the maximum difference decreased to $0.4 \pm 0.1$ mm. The latter two maximums were observed in exit profiles.

For the second prototype, the SCD (52.2 cm) was different than for the first prototype (50.3 cm), which could not be changed. Hence, a slightly larger field size resulted and is presented in Table 4.1. The resulting difference in penumbral width between the two SCDs is shown by the equation [158]:

Table 4.1
Differences in field size and penumbral width between the first and second prototype of the Australian MRI-linac (NB: SCD, source-to-collimator end distance).

<table>
<thead>
<tr>
<th>Prototype</th>
<th>SCD (cm)</th>
<th>Field Size (cm²)</th>
<th>Penumbral width (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1st</td>
<td>50.3</td>
<td>3 × 3 cm²</td>
<td>10 × 10 cm²</td>
</tr>
<tr>
<td>2nd</td>
<td>52.2</td>
<td>3.1 × 3.1 cm²</td>
<td>10.4 × 10.4 cm²</td>
</tr>
</tbody>
</table>

\[ P = D \left( \frac{SSD - SCD}{SCD} \right) \]  \hspace{1cm} (4.2)

where \( P \) = geometric penumbral width, \( D \) = source diameter, and \( SCD \) = source-to-collimator end distance. Substituting in the SCD for each prototype and source diameter of 3 mm, this gives the same value of 0.2 cm (to within one decimal place) for the penumbral width in each prototype.

In each case, i.e. 0, 0.9, and 1.5 T, the linatron was realigned and this was done manually as no permanent lasers could be introduced into the magnetic field at this time; hence a slight asymmetry in the full profile is visible. A small increase in dose was observed in the centre of the profiles taken at 10 mm depth for the 10 × 10 cm² field size at 1.5 T. This increase was approximately 6.5%. When investigating the effect of the copper window plate for the 0.9 T configuration there was no increase in dose along the central axis when comparing profiles with and without the window in place.

4.3.2 Variation in Penumbral Width Over Different Magnetic Fields for Different Depths of Lung Tissue

Figures 4.16 to 4.17 display the penumbral widths for increasing solid lung thickness. The profiles at the entrance to the solid lung were compared at each
Figure 4.7 Entrance interface RHS profiles for $3 \times 3$ cm$^2$ field size, aligned to 50% geometric penumbral edge: (a) solid lung, and (b) solid water (NB: SL, solid lung; SW, solid water).
Figure 4.8 Exit interface RHS profiles for $3 \times 3$ cm$^2$ field size, aligned to 50% geometric penumbral edge: (a) solid lung, and (b) solid water (NB: SL, solid lung; SW, solid water).
thickness and plotted for each magnetic field strength. The profiles at a depth of 20 mm in the solid lung were also plotted, although these measurements were only acquired at 0.9 T. No trend was observed with increasing solid lung thickness for either the entrance profiles or profiles at depth. All differences between the penumbral widths with increasing solid lung thickness was <1 mm.

4.4 Discussion

It was expected that the 0.9 and 1.5 T magnetic fields would cause narrowing of profiles in solid water and solid lung due to their longitudinal orientation. The magnetic field narrowed the penumbra by up to $4.4 \pm 0.1$ mm for 1.5 T and up to $2.5 \pm 0.1$ mm for 0.9 T. It was shown that in both phantom materials, exit interface profiles experienced more tightening due to the 0.9 and 1.5 T
Figure 4.10 Entrance interface RHS profiles for $10 \times 10 \text{ cm}^2$ field size, aligned to 50% geometric penumbral edge: (a) solid lung, and (b) solid water (NB: SL, solid lung; SW, solid water).
Figure 4.11 Exit interface RHS profiles for $10 \times 10 \text{ cm}^2$ field size, aligned to 50% geometric penumbral edge: (a) solid lung, and (b) solid water (NB: SL, solid lung; SW, solid water).
Figure 4.12 RHS profiles at a depth of 30 mm for $10 \times 10 \text{cm}^2$ field size in solid water and a depth of 20 mm in solid lung, aligned to 50% geometric penumbral edge (NB: SL, solid lung; SW, solid water).

Figure 4.13 Example of full profile in solid lung at different magnetic field strengths, $3 \times 3 \text{cm}^2$ field size, exit interface profile (NB: SL, solid lung).
Figure 4.14 Difference in 80-20% penumbral widths between each magnetic field strength and 0 T, for profile measurements across film in solid lung and corresponding depths of solid water, 3 × 3 cm$^2$ field size: (a) At entrance of solid lung, (b) At a depth of 30 mm (i.e., a point 20 mm into solid lung), and (c) At exit of solid lung (NB: LHS, left-hand side; RHS, right-hand side; SW, solid water; SL, solid lung).
Figure 4.15 Difference in 80-20% penumbral widths between each magnetic field strength and 0 T, for profile measurements across film in solid lung and corresponding depths of solid water, 10 × 10 cm² field size: (a) At entrance of solid lung, (b) At a depth of 30 mm (i.e., a point 20 mm into solid lung), and (c) At exit of solid lung (NB: LHS, left-hand side; RHS, right-hand side; SW, solid water; SL, solid lung).
Figure 4.16 80-20% penumbral widths as solid lung thickness changes for a $3 \times 3$ cm$^2$ field size: (a) Entrance profiles, and (b) Profiles at a depth of 20 mm into solid lung (i.e. 30 mm in solid water) (NB: LHS = left-hand side, RHS = right-hand side).
Figure 4.17 80-20% penumbral widths as solid lung thickness changes for a $10 \times 10$ cm$^2$ field size: (a) Entrance profiles, and (b) Profiles at a depth of 20 mm into solid lung (i.e. 30 mm in solid water) (NB: LHS = left-hand side, RHS = right-hand side).
fields than the entrance interface profiles and profiles at a depth of 30 mm; this was expected due to the wider lateral spread of electrons at this depth, which enabled the magnetic field to further lessen this spread. The 1.5 T magnetic field brought the penumbral widths more in alignment between materials, with the maximum difference being reduced from $2.5 \pm 0.1$ mm at 0 T to $0.4 \pm 0.1$ mm at 1.5 T. In some cases, the 0 T penumbra was narrower than for 0.9 T; however this difference was $<1$ mm; within the uncertainty of the measurement.

Previous to the installation of the MRI, the Faraday cage had been open via a window between the linatron (outside the cage) and where the magnet would be positioned inside the cage. This window was closed for the 1.5 T measurements, which would, in part, contribute to the focused electron contamination observed in the slight increase in the centre of profiles taken at a depth of 10 mm, before the profiles flattened out with depth. This effect was not visible on profiles taken in 0.9 T, either with or without the window closed. The effect was therefore largely attributed to the increased field strength focusing the electrons [159], with a small contribution due to electrons being freed from the copper. As other studies have observed [81], the longitudinal magnetic field can result in an increase in skin dose and therefore needs to be considered for patient treatments.

The field sizes for the measurements in 0.9 T magnetic field could not be matched to the 0 and 1.5 T measurements due to the change in source-to-MLC end distance between the first and second prototypes of the MRI-linac. However, penumbral width differences between the two distances were negligible, agreeing to within one decimal place of a width of 0.2 mm. The beam energy was the same for all measurements; hence, this would not impact upon the penumbral differences.

In this study, the dose enhancement expected in a longitudinal field could not
be quantified due to limits in the reproducibility of the linatron at high doses, which was 15 Gy per measurement in this instance. The average uncertainty overall was small and the maximum uncertainty was within an acceptable limit for film data.

A limitation of this study was that the experiment in the Agilent magnet could only be carried out at the 0.9 T position and not at the 1.0 T position where a patient would be treated. This was due to the fringe field affecting the linatron components if it were to be moved closer to the bore centre. This study utilised solid lung that was 0.30 g/cm$^3$ in density, but it is appreciated that healthy lung tissue density can alter between 0.1 and 0.4 g/cm$^3$ depending on patient age, respiration stage, and spatial position within the lung, and this will have an impact upon penumbral flaring [160–162]. Tracking of lung tumours, therefore, may become more critical under the influence of a longitudinal magnetic field due to the tighter penumbra, whereby changes in density within the lung tissue could have a larger effect.

### 4.5 Conclusion

The small changes in penumbral width due to the application of a 0.9 and 1.5 T magnetic field were quantified; the narrowing was greater due to the 1.5 T field (up to $4.4 \pm 0.1$ mm), compared to the 0.9 T field (up to $2.5 \pm 0.1$ mm), as expected. A comparison of profiles measured at the interfaces and mid-depth of a solid lung cavity, and equivalent solid water depths demonstrated that the 1.5 T magnetic field brought the penumbral widths between these phantom materials more into alignment by a maximum difference of $0.4 \pm 0.1$ mm. The larger lateral spread of electrons with depth resulted in exit interface profiles becoming more greatly narrowed by the magnetic field than entrance interface or mid-depth profiles, by up to $2.6 \pm 0.1$ mm.
Chapter 5

Silicon Array Detector Performance in a Permanent Magnet System

Sections of this chapter are presented in the publication:


5.1 Introduction

A permanent magnet named Magnetic Apparatus for RaDiation Oncology Studies (MARDOS) capable of a magnetic field strength of up to 1.2 T was constructed in collaboration between CMRP and Illawarra Cancer Care Centre (ICCC), Wollongong, Australia. MARDOS was designed to be paired with a standard linac and can be positioned with either the magnetic field in the transverse or inline orientation with respect to the radiation beam. The benefit of MARDOS was to provide similar magnetic field strengths to the MRI-linac, yet with the option of a different magnetic field orientation with respect to the
radiation beam. It maintains static magnetic field conditions, in comparison to the changing magnetic field conditions of the MRI-linac. It provided an ideal preliminary set up for observing the magnetic field effects on silicon array detectors and film before the MRI-linac was available for testing with these detectors. The M512 was previously utilised in MARDOS to investigate these magnetic field effects on that detector [4].

This current work will investigate the utility of the DUO silicon array detector as a magnetic-field compatible device, in the magnetic field of MARDOS permanent magnet. The DUO silicon detector, described in Section 2.6.5.4, has a higher resolution (200 µm) than the M512 (2 mm), and hence should be better suited to measuring changes in the dose distribution due to the magnetic field. The DUO was therefore utilised in this study. These results were compared to previously published M512 data [4]. Due to the anticipated magnetic field effects in detector air gaps apparent in other studies [2–4,82,91], the size of the air gap above the sensitive volumes of the DUO detector was investigated in the transverse orientation. Experiments were completed in solid water (Gammex Inc., WI, USA) and then solid lung (constructed in-house, density: 0.3 g/cm³ [163]) phantoms. EBT3 Gafchromic film was the benchmark dosimeter for this work to distinguish between intrinsic and extrinsic effects of a magnetic field on the silicon detector; effects of a magnetic field on EBT3 film have been shown to be minimal [107].

This was a preliminary study to determine the feasibility of this detector for future implementation on the Australian MRI-linac for pre-treatment QA.
5.2 Methodology

5.2.1 MARDOS

MARDOS (Figure 5.1) consists of neodymium-iron-boron magnets with steel cones attached. A cone spacing of 29 mm is required to provide the 1.2 T magnetic field. A complete magnetic field map of MARDOS utilising a Gaussmeter has been previously completed [164]. To compare to data acquired under 0 T conditions, a non-magnetic replica of MARDOS was constructed and included steel cones and steel supporting structure to maintain similar scatter conditions as the magnetic version. The benefit of MARDOS was to provide similar magnetic field strengths to the Australian MRI-linac, yet with the option of a different magnetic field orientation with respect to the radiation beam.

Measurements were performed on MARDOS and a Varian 2100C Linac at 6 MV with a nominal dose rate of 600 cGy/min. Some profiles acquired under 10 MV have been included for comparison of the profile distortion under a transverse magnetic field. The MARDOS magnet was positioned next to the linac couch, and the linac head rotated to 270°. The DUO detector was inserted into a custom-designed aluminium frame so that it would fit between the magnet cones (Figure 5.2). A separate aluminium frame was constructed with a Perspex window to house the film and position it in the same set-up as the DUO. Solid water phantoms were utilised for the first set-up and then the measurements were repeated with solid lung phantoms.

The detector is normally encased between two 5 mm thick slabs of PMMA with a small air gap milled out of the PMMA just above the detector to negate the dose enhancement of silicon and to protect the wire-bonding [29,137]. This was not possible due to space restrictions inside MARDOS, hence, the air gap was instead milled straight into the phantoms, which were placed flush with the detector.
Figure 5.1 Magnetic Apparatus for RaDiation Oncology Studies (MARDOS): a permanent magnet capable of both inline and transverse orientations between magnetic field and radiation beam, and of field strengths up to 1.2 T.

Figure 5.2 The DUO silicon array detector inserted into aluminium frame to enable positioning in magnet cones.
5.2.2 Inline

For the inline orientation, the DUO detector was inserted into the gap between the magnet cones, perpendicular (i.e. face-on) to the cones and radiation beam (Figure 5.3(a)). The closest that the cones could be positioned to one another in this orientation was 30 mm and allowed for a magnetic field strength of 0.95 T. The centre of the cones, and hence, the sensitive volumes of the detector, was positioned at an extended SSD of 150 cm. Custom phantoms were cut from solid water and solid lung to give 1.7 mm build-up material and 1.2 mm backscatter material, which were the maximum phantom thicknesses possible between the cones. Phantoms were 50 mm wide to cover the sensitive volumes of the detector. A 2 mm air gap was milled from the phantom in the section directly above the sensitive volumes. Profiles were measured in 0 T and 0.95 T magnetic fields. Three field sizes were investigated: $0.5 \times 0.5 \text{ cm}^2$, $0.9 \times 0.9 \text{ cm}^2$, $1.3 \times 1.3 \text{ cm}^2$ at isocentre, corresponding to $0.8 \times 0.8 \text{ cm}^2$, $1.4 \times 1.4 \text{ cm}^2$, and $2.0 \times 2.0 \text{ cm}^2$ at the extended SSD. Field sizes were constrained by the aperture in the magnet cones, through which the beam had to pass. DUO was then replaced with film in the same set-up but with no air gap.

5.2.3 Transverse

For the transverse orientation, a different set of cones was manufactured with a slot machined into the cones so that the DUO could be inserted, edge-on to the centre of each (Figures 5.3(b) and 5.3(c)). The coaxial axis of cones was perpendicular to the radiation beam, and the DUO was also perpendicular (i.e. face-on) to the radiation beam. The highest magnetic field (1.2 T) achievable for MARDOS could be utilised in this orientation; hence there was a cone separation of 29 mm. The extended SSD of 150 cm was also used for this orientation. The set up allowed the phantoms to be thicker in this orientation; hence 50 mm build up and 50 mm backscatter was attached to the detector. Due to the cone separation, the phantoms could only be 29 mm wide, which would
not allow for full scatter conditions, and therefore would add to the uncertainty in the measurement. Profiles were measured in 0 T and 1.2 T magnetic fields. Three field sizes were investigated: $0.5 \times 0.5 \text{ cm}^2$, $1.0 \times 1.0 \text{ cm}^2$, $1.5 \times 1.5 \text{ cm}^2$ at isocentre, corresponding to $0.8 \times 0.8 \text{ cm}^2$, $1.5 \times 1.5 \text{ cm}^2$, and $2.3 \times 2.3 \text{ cm}^2$ at the extended SSD. Field sizes were constrained by the gap between magnet cones. Measurements were taken with three different air gap sizes above DUO: 0.8 mm, 1.4 mm, and 2.0 mm. A bolus material of near-water equivalence was also inserted into the air gap to create a 0 mm air gap. Solid water could not be used for this purpose as only a more malleable material could be placed against the sensitive detector and connectors so as not to damage these components. As in the inline case, profiles were repeated with film in the same set-up but with no air gap.

### 5.2.4 Calibration and Analysis

Film was calibrated utilising the methods outlined in Section 2.6.2. The DUO was calibrated by converting the raw reading to charge by scaling the raw reading by the maximum of a specific gain for the AFE0064. The charge was then converted to dose based on a linear relationship. The DUO was also equalised by the method outlined in Section 3.5.

Dose per pulse correction factors for an extended SSD of 150 cm were applied to DUO, ranging between 1.0-3.7% for differing field sizes, as recommended from the characterisation [137].

FWHM, 80-20% penumbral widths and maximum dose differences between detectors and between the presence and absence of a magnetic field were investigated for the inline and transverse set-ups. For the transverse set-up, profile symmetry was used to investigate the skewness of the profiles under the magnetic field in comparison to zero field. Symmetry was calculated as [5]:

\[ \text{Symmetry} = \frac{D_{\text{max}} - D_{\text{min}}}{D_{\text{max}} + D_{\text{min}}} \]
Figure 5.3 The DUO inserted into MARDOS: (a) in the inline orientation with phantom attached to face of detector, (b) in the transverse orientation without phantom attached, and (c) in the transverse orientation with phantom attached.
\[
%\text{symmetry} = \frac{\text{area}_L - \text{area}_R}{\text{area}_L + \text{area}_R} \times 100
\]  

(5.1)

where \(\text{area}_R\) and \(\text{area}_L\) are the right and left areas under the profile measured from the central axis out to the distance corresponding to FWHM.

## 5.3 Results

The absolute dose profiles and normalised profiles are shown in Figures 5.4-5.18. Tables 5.1-5.4 display 80-20% penumbral width and FWHM data and Table 5.5 displays symmetry data. Previously published M512 data [4] are displayed on the graphs and in the tables for comparison. Average uncertainties were within \(\pm 2.7\%\) for DUO and within \(\pm 3.1\%\) for film. M512 uncertainty was previously reported as \(\pm 3\%\) [4]. Error bars are only displayed for M512, and not DUO or film, for clarity on the graphs due to the higher resolution of the latter two.

### 5.3.1 Inline

Profiles were normalised to 0 T to demonstrate the difference in maximum dose between 0 T and 0.95 T. Absolute dose profiles were included to observe the difference between detectors.

The DUO measured the penumbral narrowing of the solid water profiles due to the magnetic field, which was a maximum of 0.2 mm. The effect was more pronounced in lung at 0.7 mm. Overall, the maximum difference measured between the DUO and film profiles was 1.0 mm, and between the DUO and the M512 was 0.9 mm. The average difference for each was 0.5 mm.

The DUO measured the narrowing of FWHM to be a maximum of 0.4 mm for both solid water and solid lung. Overall, the maximum difference between the DUO and film was 0.9 mm and was 2.0 mm between the DUO and the M512. The average difference was 0.3 mm and 0.5 mm, respectively.
Figure 5.4 Absolute dose profiles in solid water for inline orientation of MARDOS comparing the DUO to film and previously published M512 data [4], 0.8 × 0.8 cm² field size.

Figure 5.5 Absolute dose profiles in solid water for inline orientation of MARDOS comparing the DUO to film and previously published M512 data [4], 1.4 × 1.4 cm² field size.
Figure 5.6 Absolute dose profiles in solid water for inline orientation of MARDOS comparing the DUO to film and previously published M512 data [4], 2.0 × 2.0 cm² field size.

Figure 5.7 Profiles in solid lung, normalised to 0 T, for inline orientation of MARDOS comparing the DUO to film and previously published M512 data [4], 0.8 × 0.8 cm² field size.
Figure 5.8 Profiles in solid lung, normalised to 0 T, for inline orientation of MAR-DOS comparing the DUO to film and previously published M512 data [4], 1.4 × 1.4 cm² field size.

Figure 5.9 Profiles in solid lung, normalised to 0 T, for inline orientation of MAR-DOS comparing the DUO to film and previously published M512 data [4], 2.0 × 2.0 cm² field size.
Table 5.1
FWHM and 80-20% penumbral widths for profiles measured in solid water under 0 T and 0.95 T static inline magnetic fields in MARDOS. Uncertainty: ±1 mm.

<table>
<thead>
<tr>
<th>Field Size (mm²)</th>
<th>DUO 0 T</th>
<th>0.95 T</th>
<th>Film 0 T</th>
<th>0.95 T</th>
<th>M512 [4] 0 T</th>
<th>0.95 T</th>
</tr>
</thead>
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<tr>
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<td></td>
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<td></td>
</tr>
<tr>
<td>Penumbral Width</td>
<td>8 × 8</td>
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<td>2.8</td>
<td>3.3</td>
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<td>3.9</td>
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<td></td>
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</tr>
<tr>
<td>Penumbral Width</td>
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<td>3.0</td>
<td>2.5</td>
<td>3.3</td>
<td>2.9</td>
<td>2.6</td>
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<tr>
<td>(mm)</td>
<td>14 × 14</td>
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<td><strong>FWHM</strong></td>
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<td>(mm)</td>
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<td>8.9</td>
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<td>14.6</td>
<td>14.4</td>
<td>13.8</td>
<td>14.2</td>
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<tr>
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<td>20.2</td>
<td>20.5</td>
<td>20.1</td>
<td>20.3</td>
<td>20.6</td>
</tr>
</tbody>
</table>

Table 5.2
FWHM and 80-20% penumbral widths for profiles measured in solid lung under 0 T and 0.95 T static inline magnetic fields in MARDOS. Uncertainty: ±1 mm.

<table>
<thead>
<tr>
<th>Field Size (mm²)</th>
<th>DUO 0 T</th>
<th>0.95 T</th>
<th>Film 0 T</th>
<th>0.95 T</th>
<th>M512 [4] 0 T</th>
<th>0.95 T</th>
</tr>
</thead>
<tbody>
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<td></td>
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</tr>
<tr>
<td>Penumbral Width</td>
<td>8 × 8</td>
<td>3.3</td>
<td>3.2</td>
<td>3.2</td>
<td>3.1</td>
<td>2.9</td>
</tr>
<tr>
<td>(mm)</td>
<td>14 × 14</td>
<td>3.9</td>
<td>4.1</td>
<td>3.7</td>
<td>3.6</td>
<td>3.6</td>
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<tr>
<td></td>
<td>20 × 20</td>
<td>4.6</td>
<td>4.6</td>
<td>4.0</td>
<td>4.1</td>
<td>3.9</td>
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<tr>
<td><strong>Right</strong></td>
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<td></td>
</tr>
<tr>
<td>Penumbral Width</td>
<td>8 × 8</td>
<td>2.9</td>
<td>3.2</td>
<td>3.2</td>
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<td><strong>FWHM</strong></td>
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<td>14.1</td>
<td>13.9</td>
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<td>14.3</td>
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<tr>
<td></td>
<td>20 × 20</td>
<td>20.4</td>
<td>20.4</td>
<td>20.0</td>
<td>20.0</td>
<td>20.5</td>
</tr>
</tbody>
</table>
Central axis dose increase due to the inline magnetic field was measured by the DUO to be 1.8% in solid water, which was within the uncertainty of the 0 T measurements. It was also within the uncertainty of film and the M512. Measurements with the DUO in the solid lung phantom demonstrate a central axis dose increase due to the magnetic field of 11.5%. DUO measured a higher dose than film under 0.95 T by a maximum difference of 7.5%, and M512 by 6.1%.

### 5.3.2 Transverse

Profiles measured in solid water were normalised to 0 T to observe how the bolus brought the 1.2 T DUO profile into alignment with film. All profiles were then normalised to demonstrate the asymmetric shape of the profiles taken under the 1.2 T transverse field compared to 0 T.
Figure 5.11 Normalised profiles in solid water for transverse orientation of MARDOS comparing the DUO to film and previously published M512 data [4], 1.5 × 1.5 cm$^2$ field size.

Figure 5.12 Normalised profiles in solid water for transverse orientation of MARDOS comparing the DUO to film and previously published M512 data [4], 2.3 × 2.3 cm$^2$ field size.
Figure 5.13 Normalised profiles in solid lung for transverse orientation of MARDOS comparing the DUO to film and previously published M512 data [4], 0.8 × 0.8 cm$^2$ field size.

Figure 5.14 Normalised profiles in solid lung for transverse orientation of MARDOS comparing the DUO to film and previously published M512 data [4], 1.5 × 1.5 cm$^2$ field size.
Figure 5.15 Normalised profiles in solid lung for transverse orientation of MARDOS comparing the DUO to film and previously published M512 data [4], 2.3 × 2.3 cm² field size.

Table 5.3
FWHM and 80-20% penumbral widths for profiles measured in solid water under 0 T and 1.2 T static transverse magnetic fields in MARDOS. Uncertainty: ±1 mm.

<table>
<thead>
<tr>
<th>Field Size (mm²)</th>
<th>DUO</th>
<th>Film</th>
<th>M512 [4]</th>
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<tbody>
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<td></td>
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<td>1.2 T</td>
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<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>8 × 8</td>
<td>3.3</td>
<td>3.0</td>
<td>2.7</td>
</tr>
<tr>
<td>Penumbral Width</td>
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<td>8.6</td>
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<td>8.1</td>
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<tr>
<td>(mm)</td>
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Table 5.4
FWHM and 80-20% penumbral widths for profiles measured in solid lung under 0 T and 1.2 T static transverse magnetic fields in MARDOS. Uncertainty: ±1 mm.

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<th>Film</th>
<th>M512 [4]</th>
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<td>1.2 T</td>
<td>0 T</td>
</tr>
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<td>3.8</td>
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<td>3.0</td>
</tr>
<tr>
<td>Penumbral Width</td>
<td>15 × 15</td>
<td>4.1</td>
<td>3.7</td>
</tr>
<tr>
<td>(mm)</td>
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<td></td>
<td>23 × 23</td>
<td>23.3</td>
<td>22.8</td>
</tr>
</tbody>
</table>

The penumbra was widened by the transverse magnetic field by a maximum of 0.5 mm in solid water and 2.3 mm in lung, as measured by the DUO. The DUO differed from film by a maximum of 1.6 mm and from the M512 by 1.9 mm. The average difference was 0.5 mm in both cases.

The FWHM in solid water widened under the transverse magnetic field by 0.3 mm, however, for the solid lung, the FWHM was wider under 0 T conditions by 0.3 mm. The DUO differed from film by a maximum of 0.8 mm, and from M512 by a maximum of 0.7 mm. The average difference was 0.4 mm and 0.3 mm, respectively.

The profiles acquired under the transverse field were found to be a maximum of 3.0% more asymmetric for solid water than those acquired under 0 T. The effect was more extreme in solid lung, with a maximum of 4.6% asymmetry measured compared to 0 T.

The maximum drop in dose from 0 T to 1.2 T in solid water without
Table 5.5
Symmetry reported as percentage in profiles measured under 0 T and 1.2 T static transverse magnetic fields in MARDOS.

<table>
<thead>
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</tr>
<tr>
<td>8 × 8</td>
<td>0.3</td>
<td>3.3</td>
</tr>
<tr>
<td>15 × 15</td>
<td>0.1</td>
<td>2.3</td>
</tr>
<tr>
<td>23 × 23</td>
<td>0.3</td>
<td>4.9</td>
</tr>
</tbody>
</table>

bolus was 10.0%. With bolus, this drop reduced to 2.6%, which was within uncertainty. The difference between DUO and film under the transverse field without bolus in the air gap of DUO was 6.5%. With bolus, this difference reduced to 1.4%.

Under 0 T conditions, for the largest field size, the difference between the smallest air gap (0.8 mm) and largest air gap (2.0 mm) was 2.2%, which was within the uncertainty of the measurements. This increased to 11.3% under the transverse magnetic field. This difference further increased to 18.4% when comparing profiles with no air gap present and those with a 2.0 mm air gap.

An energy comparison between 6 MV and 10 MV is displayed in Figure 5.17 and Figure 5.18 to note the accentuated asymmetry in the profile at the higher energy. The profile is more notably distorted in the solid lung.

5.4 Discussion

The static inline magnetic field had minimal effect on the profiles in solid water, as was demonstrated by the penumbral width, FWHM, and dose enhancement analysis. A larger effect was seen for the penumbra and FWHM in solid lung,
Figure 5.16 Comparison of profiles in solid water for transverse orientation of MAR-DOS with different air gap sizes over the sensitive volumes of the DUO detector.

Figure 5.17 Comparison of profiles in solid water for transverse orientation of MAR-DOS with different energies and using the DUO detector.
Figure 5.18 Comparison of profiles in solid lung for transverse orientation of MAR-DOS with different energies and using the DUO detector.

however, these differences were within the ±1 mm positional tolerance of the jaws [146]. The DUO agreed with both film and the M512 to within this tolerance. Dose enhancement was greater in lung, as expected, due to the lower density. The 2 mm air gap in the solid lung phantom under the inline magnetic field was more problematic for the DUO than the M512. The DUO has previously been optimised with a 0.5 mm air gap to match to film [137]. This study was limited in that only one phantom could be manufactured from the available solid lung material, and it had already been optimised for the M512 for the previous study [4]. Suitable air gap sizes can be used moving forward with implementation of novel silicon array detectors on the Australian MRI-linac.

The transverse field distorted the profiles in the direction of the magnetic force, as expected. The penumbra was widened, most notably in the solid lung phantom, by a maximum of 2.3 mm. The FWHM was not altered
significantly. The asymmetry of the profiles was also present to a greater extent in solid lung, with a difference between 0 T and 1.2 T of 4.6% asymmetry. The greater distortion in solid lung was expected due to its lower density. The contact and pressure of the bolus on the detector was a source of uncertainty that could not be quantified. The effect of the air gap above the DUO on detector profiles was non-negligible in a magnetic field and required bolus in the transverse field to create a 0 mm air gap to match the 0 T case. Bolus will be included in future designs. The distortion of the profiles was accentuated under a 10 MV beam, compared to 6 MV. There was a focus on the 6 MV analysis at this time as that was important for continuing the investigation at similar energies on the MRI-linac.

It should be noted that due to the spatial restrictions of the MARDOS system, the limited phantom size does not provide full scatter conditions, which would have an effect on the results. Calibration should also be carried out under magnetic field conditions, which could not be completed on MARDOS due to the spatial limitations. Full scatter conditions are able to be maintained on the MRI-linac (Chapter 6).

5.5 Conclusion

The effects on dose profiles under the magnetic field of a permanent magnet were measured. The extrinsic effects were measured with the DUO silicon array detector and the intrinsic effects were observed by comparing to Gafchromic film. A comparison was also made to the M512, and the DUO was found to be an appropriate alternative and improvement in terms of its higher spatial resolution. The penumbra and FWHM measured by the DUO agreed with film and M512 to within an average of ±0.5 mm. As expected, the lower density of solid lung meant that this material was more susceptible to demonstrating magnetic field effects in the dose deposited. The greatest penumbral narrowing
due to the inline field (0.7 mm), and the greatest increase of penumbral width
due to the transverse field (2.3 mm) occurred in lung. The largest asymmetry
due to the transverse magnetic field (4.6%) was also in solid lung. MARDOS
provided a suitable environment for preliminary testing of the magnetic field
effects on, and utilising monolithic high resolution silicon array detectors. These
results were a significant step towards investigating the in-house developed,
silicon array detectors on the Australian MRI-linac.
Chapter 6

Silicon Array Detector Implementation in an Inline MRI-Linac

6.1 Introduction

The aim of this study was to investigate the potential of novel silicon array detectors for use as a pre-treatment QA device on an MRI-linac. The MRI-linac used in this chapter was the second prototype of the Australian MRI-Linac Program with a 1.0 T Agilent magnet positioned in the inline orientation between magnetic field and radiation beam, which was described in Section 2.1.2 and 4.2. The effect that the magnetic and RF fields have on the detector, as well as on the deposited dose, was investigated. Full width at half maximum (FWHM) and 80-20% penumbral widths were measured from beam profiles. MR images were acquired to sequentially analyse the effect of the detector and its components on image quality for a progression towards dynamic dosimaging. EBT3 Gafchromic film was utilised as the benchmark detector to determine differences between intrinsic and extrinsic effects of the magnetic field on the silicon detector and dose distribution, respectively. Magnetic field effects on EBT3 film have been found to be minimal [107].
Use of the M512 on the 1.0 T inline MRI-linac represents the first introduction of this unique high spatial and temporal resolution device on the first MRI-linac in Australia. It is also the first use of this device in MRI-linac systems worldwide.

6.2 Methodology

6.2.1 Challenges

This investigation faced several challenges due to the non-standard nature of the set-up in comparison to a clinical linac. There was no equivalent sync pulse as is normally used to trigger the detector on a clinical linac, so another method needed to be determined. Noise reduction was addressed in Chapter 3, however this still required assessment for this study. The signal at the extended SSD was also very low.

6.2.2 Experimental Set-Up

High mode on the linatron was selected with a trigger rate of 200 Hz. The detector was sandwiched between two 5 mm PMMA plates; the front plate included the milled air gap above the sensitive volumes (2 mm air gap width for M512, 0.5 mm for DUO). The detector was housed in an aluminium shielding case with walls 2 mm thick (described in Chapter 3). This was positioned in the centre of the MRI bore. A 15 mm build-up depth of solid water (Gammex Inc., WI, USA) was positioned in front of the detector (inside the shielding case), and 10 cm of solid water backscatter material was positioned behind the detector (outside the shielding case). Rail position 1 on the linatron was used for the magnetic field measurements at 1.0 T (Figure 6.2); this was the closest position that the linatron could be to the MRI to gain maximum signal detection. The source-to-detector distance was 1.814 m. Position 8 (Figure 6.1) of the linatron was used for the fringe field measurements at ‘0 T’ (measured with a Gaussmeter...
to be 0.05 T); this was the furthest distance possible between linatron and MRI. These positions were utilised to maintain the same SSD between 0 T and 1.0 T. This was due to the requirement for a near-0 T fringe field measurement and the requirement to have the detector in the split of the bore, as it would not fit in the bore itself due to its length exceeding the dimensions of the bore width. Hence, this is why different positions were utilised to those in Chapter 4. The measured magnetic field strength of 0.05 T was as close to 0 T as achievable without ramping down the magnet, which is not recommended or easily viable with superconducting MRI machines.

6.2.3 Sync Pulse, Signal Detection and Noise Reduction

Due to the absence of a sync pulse, the method determined to trigger the M512 and the DUO (in two separate experiments) was via the signal detected first by a single diode (n-type with sensitive volume dimensions: 1.0 × 1.0 × 1.0 mm³) and then sent to M512/DUO from the diode. The diode had a
Figure 6.2 MagicPlate-512 inside aluminium shielding case positioned in the bore of the Australian MRI-linac (1.0 T).
sensitive volume large enough to detect signal for triggering. The diode was positioned in-field, upstream of the MLCs so that it would not perturb the dose distribution but would still receive an unencumbered signal from the beam.

Signal loss was anticipated not only because of the extended SSD, but also due to the detector only being triggered partway through a beam pulse. This was due to the time delay between the diode first detecting the electron gun pulse and sending it to M512/DUO before the detectors would integrate (Figure 6.3). Hence, initial measurements were completed at $d_{max}$ to ensure the maximum signal possible.

The detector was shielded identically to the method described in Chapter 3. Digital cables were made to be 1 m in length to remove the FPGA from the bore and were shielded with aluminium tape. The FPGA was also encased in its own aluminium shielding box. The shielded detector and readout electronics were introduced slowly into the MRI-bore. Another method of shielding was also tested, which was to encapsulate the FPGA in the same large aluminium case as the detector and AFE boards; this created a more compact system but did not allow the FPGA to be distanced from the bore.

Figure 6.3 MagicPlate-512 electronic readout system integrating partway through MRI-linac electron gun pulse resulting in a loss of signal.
6.2.4 Beam Profiles

Profiles across the MRI-linac beam were taken with M512 in solid water; the explanation as to why measurements were not continued with DUO is included in Section 6.3. Two small field sizes were investigated: $2.1 \times 1.9 \text{ cm}^2$ and $4.2 \times 3.8 \text{ cm}^2$. An equivalent dose of 2 Gy, as measured by the ionisation chamber in the head of the linatron, was delivered for each measurement. Equalisation was performed separately on a standard linac with a flattening filter to produce uniform irradiation of the array, which was not possible on the linatron due to the FFF beam. The method used for equalisation is described in Section 3. Profiles were compared to film in the same setup, including with aluminium casing.

Beam profiles were then measured in solid lung (CIRS Inc., Norfolk, VA, USA). A set up was created to try to accentuate any penumbral narrowing due to the magnetic field, to then measure with the M512 and compare to film. Solid lung, 30 mm thick was positioned in front of the detector, and 30 mm of solid water was placed in front of the solid lung. This was repeated behind the detector, with 30 mm of solid lung behind the detector, and 30 mm of solid water behind the solid lung. In this way, a replica lung inside a chest cavity was created, with the detector at the centre. The experimental set-up is shown in Figure 6.4.

In the same set-up, an MR imaging sequence was delivered while the detector acquired a reading during beam on. For this, a Gradient Echo (GRE) sequence was used due to its shorter relaxation times ($\text{TR} = 10 \text{ ms}, \text{TE} = 4.28 \text{ ms}$) compared to other sequences with longer relaxation times, for example, Spin Echo T1-weighted ($\text{TR} = 1000 \text{ ms}, \text{TE} = 30 \text{ ms}$), which caused interference in the FPGA and it desisted to function. Beam profile characteristics were compared to those acquired without concurrent imaging.
Figure 6.4 Experimental set-up: MagicPlate-512 positioned at the mid-depth of a replica lung cavity, corresponding to 30 mm of solid lung on either side of the detector in its PMMA casing, with 30 mm solid water on either side of the solid lung.

6.2.4.1 MR Images

MR images were acquired of a water phantom (Plastic Bottle 2000 mL, Siemens, Munich, Germany) with the M512 detector adjacent. This was not during dosimetry but was rather a progressive assessment to see the effect of the detector and its components on image quality. The following combinations were tested to observe the resulting image:

1. M512 power supply switched off, USB connected, and Faraday cage window panel removed

2. M512 power supply switched on, USB connected, and Faraday cage window panel removed

3. M512 power supply switched on, USB disconnected, and Faraday cage window panel removed
4. M512 power supply switched on, USB disconnected, and Faraday cage window panel secured in position

5. M512 acquiring a noise reading and Faraday cage window panel removed

6. Reference image: water phantom only, M512 removed

All images were taken with the Faraday cage window removed. The window was then secured in place for one of the images to note any difference. It was hypothesised that the USB cable would act as an antenna and hence, create more interference, which is why images were taken with the USB connected and then disconnected.

6.3 Results

6.3.1 Sync Pulse, Signal Detection and Noise Reduction

The M512 and the DUO were successfully triggered by the diode. As mentioned above, the diode had to first receive and send the signal; hence, the silicon array detectors could not be triggered until partway through an electron gun pulse. While this resulted in a systematic loss of signal, this was not problematic for M512, and therefore this detector was still able to acquire enough signal to measure beam profiles. Even during an imaging sequence, approximately 90% of the signal was visible above the noise. However, the DUO is approximately 8 times less sensitive than M512 due to its smaller sensitive volumes. While it was able to be triggered, the signal was unable to be detected clearly above the noise.

The FPGA did not function inside the bore, even when encased within the large shielding box. For the remaining experiments, the FPGA was distanced from the bore by the shielded digital cables.
6.3.2 Beam Profiles

Figures 6.5 to 6.8 display normalised profiles across the M512 in solid water in the x- and y-directions (horizontal and vertical directions, respectively). All profiles were normalised because, at this stage of the Australian MRI-Linac Program, the dose between 0 T and 1 T cannot be confirmed to be the same due to the effect of the magnetic field on the monitor chamber, which is undergoing investigation. Tables 6.1 to 6.2 contain the measured 80-20% penumbral widths and FWHM for 0 T and 1.0 T.

Normalised profiles across the M512 in solid lung in the x- and y-directions are displayed in Figures 6.9 to 6.14. Tables 6.3 to 6.4 contain the measured 80-20% penumbral widths and FWHM for the solid lung profiles.

In these particular set-ups, there was minimal difference between penumbral widths taken under 0 T and 1 T conditions. In the solid water phantoms,
Figure 6.6 Normalised profiles in solid water with the M512 on the MRI-linac, y-direction (vertical), $2.1 \times 1.9 \text{ cm}^2$ field size.

Figure 6.7 Normalised profiles in solid water with the M512 on the MRI-linac, x-direction (horizontal), $4.2 \times 3.8 \text{ cm}^2$ field size.
Figure 6.8 Normalised profiles in solid water with the M512 on the MRI-linac, y-direction (vertical), $4.2 \times 3.8 \text{ cm}^2$ field size.

Table 6.1
FWHM and 80-20% penumbral widths for horizontal profiles measured in solid water under 0 T and 1.0 T inline magnetic fields of the Australian MRI-linac. Uncertainty: $\pm 1$ mm.

<table>
<thead>
<tr>
<th>Field Size (mm$^2$)</th>
<th>M512</th>
<th>Film</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0 T</td>
<td>1.0 T</td>
</tr>
<tr>
<td>Left Penumbral</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Width (mm)</td>
<td>21 $\times$ 19</td>
<td>5.4</td>
</tr>
<tr>
<td></td>
<td>42 $\times$ 38</td>
<td>4.7</td>
</tr>
<tr>
<td>Right Penumbral</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Width (mm)</td>
<td>21 $\times$ 19</td>
<td>5.4</td>
</tr>
<tr>
<td></td>
<td>42 $\times$ 38</td>
<td>5.0</td>
</tr>
<tr>
<td>FWHM (mm)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>(mm)</td>
<td>21 $\times$ 19</td>
<td>22.2</td>
</tr>
<tr>
<td></td>
<td>42 $\times$ 38</td>
<td>41.4</td>
</tr>
</tbody>
</table>
Table 6.2
FWHM and 80-20% penumbral widths for vertical profiles measured in solid water under 0 T and 1.0 T inline magnetic fields of the Australian MRI-linac. Uncertainty: ±1 mm.

<table>
<thead>
<tr>
<th>Field Size (mm$^2$)</th>
<th>M512</th>
<th>Film</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0 T</td>
<td>1.0 T</td>
</tr>
<tr>
<td>Left Penumbral Width (mm)</td>
<td>21 x 19</td>
<td>2.9</td>
</tr>
<tr>
<td></td>
<td>42 x 38</td>
<td>3.6</td>
</tr>
<tr>
<td>Right Penumbral Width (mm)</td>
<td>21 x 19</td>
<td>2.9</td>
</tr>
<tr>
<td></td>
<td>42 x 38</td>
<td>4.0</td>
</tr>
<tr>
<td>FWHM (mm)</td>
<td>21 x 19</td>
<td>19.1</td>
</tr>
<tr>
<td></td>
<td>42 x 38</td>
<td>36.4</td>
</tr>
</tbody>
</table>

Figure 6.9 Normalised profiles in solid lung with the M512 on the MRI-linac, x-direction (horizontal), 2.1 x 1.9 cm$^2$ field size.
Figure 6.10 Normalised profiles in solid lung with the M512 on the MRI-linac, y-direction (vertical), $2.1 \times 1.9 \text{ cm}^2$ field size.

Figure 6.11 Normalised profiles in solid lung with the M512 on the MRI-linac, x-direction (horizontal), $4.2 \times 3.8 \text{ cm}^2$ field size.
Figure 6.12 Normalised profiles in solid lung with the M512 on the MRI-linac, y-direction (vertical), $4.2 \times 3.8 \text{ cm}^2$ field size.

Figure 6.13 Normalised profiles in solid lung with the M512 taken during image acquisition (I) on the MRI-linac, x-direction (horizontal), $2.1 \times 1.9 \text{ cm}^2$ field size.
Figure 6.14 Normalised profiles in solid lung with the M512 taken during image acquisition (I) on the MRI-linac, y-direction (vertical), $2.1 \times 1.9$ cm$^2$ field size.

Table 6.3
FWHM and 80-20% penumbral widths for horizontal profiles measured in solid lung under 0 T and 1.0 T inline magnetic fields of the Australian MRI-linac, as well as during imaging (I). Uncertainty: ±1 mm.

<table>
<thead>
<tr>
<th>Field Size (mm$^2$)</th>
<th>M512</th>
<th>Film</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0 T</td>
<td>1.0 T</td>
</tr>
<tr>
<td><strong>Left Penumbral</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Width (mm)</td>
<td>21 x 19</td>
<td>5.5</td>
</tr>
<tr>
<td></td>
<td>42 x 38</td>
<td>5.7</td>
</tr>
<tr>
<td><strong>Right Penumbral</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Width (mm)</td>
<td>21 x 19</td>
<td>5.5</td>
</tr>
<tr>
<td></td>
<td>42 x 38</td>
<td>4.5</td>
</tr>
<tr>
<td><strong>FWHM</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(mm)</td>
<td>21 x 19</td>
<td>22.3</td>
</tr>
<tr>
<td></td>
<td>42 x 38</td>
<td>41.8</td>
</tr>
</tbody>
</table>
Table 6.4
FWHM and 80-20% penumbral widths for vertical profiles measured in solid lung under 0 T and 1.0 T inline magnetic fields of the Australian MRI-linac, as well as during imaging (I). Uncertainty: ±1 mm.

<table>
<thead>
<tr>
<th>Field Size (mm^2)</th>
<th>M512</th>
<th>Film</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0 T</td>
<td>1.0 T</td>
</tr>
<tr>
<td>Left Penumbral Width (mm)</td>
<td>21 × 19</td>
<td>3.1</td>
</tr>
<tr>
<td>Right Penumbral Width (mm)</td>
<td>21 × 19</td>
<td>3.6</td>
</tr>
<tr>
<td>FWHM (mm)</td>
<td>21 × 19</td>
<td>19.8</td>
</tr>
<tr>
<td></td>
<td>42 × 38</td>
<td>36.4</td>
</tr>
</tbody>
</table>

80-20% penumbra of profiles taken under 1 T conditions agreed with those taken under 0 T by an average of 0.5 mm and maximum of 1.2 mm. The M512 agreed with film to within 0.8 mm and with an average agreement of 0.3 mm.

In solid lung, again, the average agreement between 0 T and 1 T was within 0.5 mm; however the maximum difference was slightly higher at 1.8 mm. The M512 and film agreed to within a maximum of 1.8 mm and average of 0.5 mm.

The profiles acquired during the imaging sequence widened the penum- bra in all cases due to RF noise, when compared to those taken in the magnetic field but without imaging. This widening was up to 1.2 mm and had an average of 0.9 mm. The M512 and film agreed to within 1.2 mm, and the average difference was 0.9 mm.

In solid water, the FWHM demonstrated <0.6 mm difference between profiles taken under 0 T and 1 T. The agreement of the M512 and film was
within 1.6 mm and an average of 0.8 mm.

In solid lung, the FWHM of profiles taken under 0 T agreed to within 1.1 mm of those taken in the magnetic field. The M512 agreed with film to within 1.7 mm and by an average of 0.6 mm.

The FWHM widened by up to 0.5 mm during an image acquisition when compared to profiles taken in the magnetic field but with no imaging present. A maximum difference of 1.5 mm was observed between the M512 and film, with an average of 1.0 mm.

During an imaging sequence (present in the profiles in Figures 6.13 and 6.14), the profile tails measured by the M512 demonstrate the RF noise affecting the detector, which is not visible in the film profiles.

Differences between the M512 and film profiles were smaller in the y-direction (vertical) profiles. For penumbral widths, this difference was small, with average differences of 0.4 mm and 0.3 mm for x and y, respectively in solid water, and 0.6 mm and 0.4 mm for x and y, respectively in solid lung. The difference for FWHM was larger, however, with average differences of 1.1 mm and 0.6 mm for x and y, respectively in solid water, and 1.2 mm and 0.7 mm for x and y, respectively in solid lung.

The measured uncertainties in response were taken as the 95% confidence level, and these were 1.3% for the M512 and 4.6% for film.

6.3.2.1 MR Images

The MR images of the water phantom adjacent to the M512 are shown in Figures 6.15 to 6.19. Figure 6.20 is the reference image of the water phantom with no detector present. The water phantom can be observed faintly at the
Figure 6.15 MR image of water phantom (Plastic Bottle 2000 mL, Siemens) taken on MRI-linac; MagicPlate-512 detector adjacent to water phantom with its power switched off, USB connected, and Faraday cage window panel removed.

bottom of Figures 6.15, and very faintly in amongst the noise of Figures 6.17 and 6.18. These are images taken when the power supply of the detector was either off, or the USB was disconnected. These three images are similar in that while there is background noise present, the water phantom is discernible. The phantoms have been labelled with an arrow in the images where they are difficult to see. Figure 6.16 was acquired when the power supply was switched on and the USB was connected; periodic lines of interference can be observed on this image and the phantom is not visible. Figure 6.19 not only had the power supply on and the USB connected, but the detector was also acquiring a background reading. This image is noisy, and the phantom is not discernible.

6.4 Discussion

The silicon detectors were successfully triggered utilising a diode, which first detected a signal from the radiation beam before sending the signal to
Figure 6.16 MR image of water phantom (Plastic Bottle 2000 mL, Siemens) taken on MRI-linac; MagicPlate-512 detector adjacent to water phantom with its power switched on, USB connected, and Faraday cage window panel removed.

Figure 6.17 MR image of water phantom (Plastic Bottle 2000 mL, Siemens) taken on MRI-linac; MagicPlate-512 detector adjacent to water phantom with its power switched on, USB disconnected, and Faraday cage window panel removed.
Figure 6.18 MR image of water phantom (Plastic Bottle 2000 mL, Siemens) taken on MRI-linac; MagicPlate-512 detector adjacent to water phantom with its power switched on, USB disconnected, and Faraday cage window panel secured.

Figure 6.19 MR image of water phantom (Plastic Bottle 2000 mL, Siemens) taken on MRI-linac; MagicPlate-512 detector adjacent to water phantom acquiring a noise reading, and Faraday cage window panel removed.
the silicon detectors. Aluminium housing, 2 mm thick, acted as acceptable shielding against RF interference for the detectors. While the M512 received enough signal for dosimetry data to be acquired above the noise, the DUO did not. The dimensions of each sensitive volume of the M512 are $500 \times 500 \times 100 \, \mu m^3$, compared to the smaller sensitive volumes of the DUO: $40 \times 800 \times 100 \, \mu m^3$. Hence, the M512 was chosen as the detector to use in remaining measurements. The Dose Magnifying Glass (DMG), discussed in Section 2.6.5.4, is proposed as the next detector for investigation in the continuation of this work, as it combines the high spatial resolution of the DUO with the higher radiation sensitivity of the M512.

The differences between the M512 and film when measuring 80-20% penumbral width and FWHM were minimal. Average agreement was within 1 mm, and there were a minimal number of cases where this value was exceeded. In solid water, the maximum difference between the M512 and film was 0.8 mm for
penumbral width and 1.6 mm for FWHM. The difference was slightly larger in solid lung, with maximums of 1.8 mm and 1.7 mm for penumbral widths and FWHM, respectively. As mentioned previously, these were outlying cases.

The differences in penumbral width and FWHM between 0 T and 1 T were minimal in both solid water and solid lung phantoms. Average differences were <1 mm, while maximum differences were up to 1.2 mm for solid water and up to 1.8 mm for solid lung. However, these maximums were a very few outlying cases. The lung cavity set-up did not accentuate the penumbral width differences between 0 T and 1 T; this was most likely due to the PMMA detector casing. For many profiles taken under 1 T, these had the wider penumbra; however, the differences from 0 T were so small they are attributed to statistical uncertainty. Set-up uncertainty would contribute in a small way to overall uncertainty, particularly in FWHM. The set up was more difficult than on a clinical linac due to several reasons. The room lasers cause uncertainty due to a slight bending along the length of the rails upon which they sit. There is no optical distance indicator on the linatron, hence all distance measurements were completed manually; this included determining that there was no tilt in the phantom and detector with respect to the beam and magnetic field direction. However, slight tilt was visible on at least one of the profiles. There was also no field light, which makes alignment of the detector and film especially difficult at this stage of the project.

For the M512, the profiles taken simultaneously with the delivery of an imaging sequence had wider penumbral widths and FWHM for all profiles when compared to those taken in a magnetic field but with no imaging. The penumbra widened by up 1.2 mm. This was most likely due to the effect of the RF noise visible on those profiles, which was observed in real-time during a dose acquisition. This was not the case for film, which was not affected by the RF noise; the largest increase in penumbral width was 0.4 mm and FWHM
was increased by up to 0.3 mm. The next software release for these silicon array detectors aims to distinguish which pulses are the result of noise and which are due to the beam, and they will, therefore, be able to be isolated.

While the 95% confidence level for the M512 uncertainty was small (1.3%), the 4.6% uncertainty in film attributed to the larger differences in the high dose gradient region of the penumbra. Profiles in the x-direction (horizontal) demonstrated larger differences between the M512 and film than in the y-direction (vertical). The MLCs open in a horizontal direction; hence, the vertical profiles were bordered by a leaf acting as a jaw, whereas the horizontal profiles were affected by inter-leaf transmission.

Profiles were normalised to 100% relative maximum due to the monitor chamber being affected by the magnetic field. The dose could not be confirmed to be the same between the fringe field position of the linatron (position 8) and its position proximal to the magnet (position 1). When this has been investigated, dose calibration for the detector can be completed under both 0 T and 1 T conditions.

As only noise was visible when taking a readout with the M512 with the linatron beam firing, it was determined that the effect of the detector on the image noise and the effect of the RF fields on the detector would be determined by a step-by-step method. This would assist in future attempts at noise reduction in the system. The images displayed various levels of noise, and some displayed the image of the water phantom, although faintly. These images displaying the phantom were taken when the power supply of the detector was either switched off, or the USB was disconnected. The USB cable was most likely acting as an antenna, which needs to be investigated in future work. The images are also degraded by the large amount of aluminium of the shielding box. Hence, a more efficient method of shielding will need to be determined.
for dynamic dosimaging. Currently, only the GRE imaging sequence has been investigated, as other sequences such as the Spin Echo T1-weighted sequence caused the FPGA to stop functioning. This is believed to be due to the longer relaxation times in these sequences causing electronic interference.

6.5 Conclusion

High resolution beam profiles were successfully measured on the Australian MRI-linac with the M512 and a comparison made with Gafchromic film. Average differences between the M512 and film when measuring FWHM and 80-20% penumbral widths were $<1$ mm. Further steps towards dynamic dosimaging were completed, with an imaging sequence delivered during the successful acquisition of beam profiles with the M512. More effective MR imaging during dosimetry measurements should result if further improved noise reduction can be introduced. A recommendation for future work is to use the DMG, which is not only sensitive like the M512, but has a higher spatial resolution.
Chapter 7

Discussion, Conclusion and Future Research

7.1 General Discussion

This project was undertaken to fill the gap in the current research for novel high resolution silicon array detector use on MRI-linacs. Silicon detectors were identified as advantageous in a magnetic field due to their high density in comparison with ionisation chambers and resulting shorter electron ranges and therefore magnetic field effects should be minimised. Silicon detectors can also be manufactured at a higher spatial resolution than ionisation chambers due to their greater efficiency. A high resolution detector is needed to accurately quantify the components of beam profiles. Since magnetic fields perturb the dose distribution, sometimes by a very small amount, a high resolution is required to quantify these small changes. However, as these detectors had not yet been tested in an MRI-linac environment, their performance under these conditions was at the forefront of this investigation to determine any limitations of these devices. This project also aimed to extend the use of these novel silicon array detectors to a dynamic pre-treatment verification process, named ‘dynamic dosimaging’; their high spatial resolution and real-time readout are advantageous to measure the dose to a moving target accurately.
This project primarily focused on taking steps towards dynamic dosimaging via the implementation of novel high resolution silicon detectors on the Australian MRI-linac. This involved observing the effect of the magnetic and RF fields on these detectors. The objectives outlined at the beginning of this thesis were met.

Objectives:

- To determine appropriate electronic shielding of silicon array detectors for use in the RF fields of an MRI-linac, so that the detectors are not affected by the RF and so that the MR images are not affected by noise from the detector.

- To characterise gel-water phantom material for use as part of MRI-linac dosimetry phantom and dosimeter for ‘static dosimaging’, which combines detector-tracking in real-time with MR imaging for concurrent position and dose verification. For eventual extension to ‘dynamic dosimaging’.

The first steps towards achieving the goals of this thesis were shielding, MRI safety, and the characterisation of a phantom capable of allowing the detectors to be MR visible for tracking during dose delivery. An appropriate aluminium shield was devised, and the detector made MRI-safe. A hydrogel phantom was successfully characterised for this purpose, allowing the detectors to be visible during an MR image sequence.

Objectives:

- To investigate magnetic field effects on and with the silicon array detector, DUO, in the transverse and inline, static magnetic fields (1.2 T) of a permanent magnet system on a standard linac.
• To investigate magnetic field effects on and with a novel silicon array
detector in the inline magnetic field of an MRI-linac (1.0 T), without MR
imaging and also during image acquisition.

The next stage was to assess the performance of the silicon array detector in
a permanent magnet as a preliminary testing ground prior to the MRI-linac.
This was completed successfully using the DUO detector on MARDOS
permanent magnet. The result of the project overall was the implementation of
a silicon array detector on the MRI-linac in terms of dosimetry. Profiles were
successfully acquired both with an imaging sequence and without. Steps were
taken towards visibility of the detector on an MR image so that in the future,
the detector can be tracked with MR imaging during dosimetry measurements.
Noise reduction will play a big part in subsequent progress towards dosimaging.

Objective:

• To investigate photon field penumbral narrowing due to aligned magnetic
fields from two different inline MRI-linac prototypes (1.5 T closed bore
and 1.0 T open bore) utilising EBT3 film to quantify the effect in solid
lung in comparison to solid water.

A secondary focus of this project was the quantification of the effect of
the magnetic field on dose distribution, particularly in lung. A Gafchomic
film study was successfully completed that measured the effect of the inline
magnetic field of the Australian MRI-linac on the dose distribution. These
effects were also observed with the silicon array detectors in the permanent
magnet, in both inline and transverse orientations, and the MRI-linac.

Overall, the novel silicon array detectors were successfully implemented
in the magnetic fields of an MRI-linac with significant progress made towards
Discussion, Conclusion and Future Research

dosimaging. The effect of the magnetic field on these detectors and the dose distribution was assessed.

7.2 Conclusion

Due to the stronger magnetic field effects experienced by ionisation chambers, and also the need for detector tracking for QA, it has been demonstrated that high resolution, silicon array detectors should be investigated for use in a magnetic field for use in MRI-linacs. The MagicPlate-512 (M512) silicon array detector was investigated in a preliminary dosimaging study and appropriate electronic shielding was determined. Penumbral width changes due to an inline magnetic field were measured with Gafchromic film on two different prototypes of the Australian MRI-linac: an industrial linatron paired first with a 1.5 T ex-clinical MRI scanner, and second with a 1.0 T split-bore magnet. The DUO silicon array detector was investigated on MARDOS permanent magnet system for up to 1.2 T. The M512 was then used to measured beam profiles on the second prototype of the Australian MRI-linac.

Chapter 2 presented a literature review, which discussed the MRI-linac concept and current forms of image-guidance/tracking in external beam radiation therapy and included current literature on the magnetic field impact on dosimetry and specific types of detectors. The two main detectors investigated in a magnetic field in this project were also introduced in this chapter: the M512 and DUO, which are high spatial and temporal resolution, silicon array detectors developed at CMRP.

In Chapter 3, MRI safety and shielding of detector electronics were addressed. In the gel-water study, it was found that a system with gel-water and the M512 had a discrepancy from solid water of \( \leq 1.1\% \) for percentage depth dose measurements, \( \leq 0.3 \) mm for profiles, \( \leq 1.3\% \) for output factors and
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Gamma analysis pass rate of 100% for 2%/2 mm for MLC-defined fields. It was therefore determined to be a feasible option to replace solid water for QA in an MRI-linac. These results are an important preliminary study for Chapter 6, as well as future studies of silicon array detectors as dynamic QA tools.

Chapter 4 investigated how an inline magnetic field can tighten the lateral spread of secondary electrons in air cavities, including lung tissue. This penumbral width study with Gafchromic film concluded that penumbra was reduced by up to $4.4 \pm 0.1$ mm in a 1.5 T inline magnetic field, but only up to $2.5 \pm 0.1$ mm for a 0.9 T field. Exit interface profiles between solid water and solid lung, when compared to entrance interface profiles and mid-depth profiles in lung, underwent greater narrowing by up to $2.6 \pm 0.1$ mm. These results conveyed the extent to which an inline magnetic field can alter the dose distribution, which is an important consideration for MRI-linac treatment planning.

The DUO was used in a MARDOS permanent magnet (up to 1.2 T) and compared to Gafchromic film, and utilising solid water and then solid lung phantoms in Chapter 5. In this study, penumbral widths measured with the DUO differed from film by a maximum of 1.0 mm in the inline magnetic field, and 1.6 mm in the transverse magnetic field. Average differences were within 0.5 mm. The magnetic field had the greatest effect in solid lung over solid water. Penumbral narrowing due to the inline field was up to 0.7 mm. The greatest increase of penumbral width due to the transverse field was 2.3 mm. The largest asymmetry due to the transverse magnetic field in comparison to 0 T was 4.6%. The effect of the air gap above the DUO on dosimeter profiles was non-negligible in a magnetic field and required bolus in the transverse field to create a 0 mm gap to match the 0 T case to within uncertainty. These results were a significant step towards investigating the in-house developed silicon array detectors on the MRI-linac.
When investigating the M512 on the Australian MRI-Linac in Chapter 6, this detector was used to measure beam profiles in solid water and solid lung phantoms, and a comparison was made with Gafchromic film. Average differences between the M512 and film were minimal (<1 mm) in the measurement of 80-20% penumbral widths and FWHM. The largest discrepancy between penumbral widths was 1.8 mm in solid lung. A GRE image sequence was delivered during the acquisition of beam profiles with the M512. This silicon array detector was successfully implemented on the MRI-linac, with progress also made towards dosimaging, fulfilling the aims of this thesis.

Overall, these experiments were found to be successful although there were some limitations discussed, such as the air gap size in solid lung in the MARDOS study. This work is significant for a move towards utilising silicon detectors as a QA tool for MRI-linacs. Improved QA, including for dynamic dose delivery, will add to confidence in MRI-linac treatments.

### 7.3 Recommendations for Future Work

Recommended future work to reduce noise and better enable dynamic dosimaging would include ensuring that an appropriate air gap size is utilised for each specific detector. The MARDOS study was limited in this respect, as well as in the geometry of the phantoms that could fit in the MARDOS structure. However, these issues can be resolved on the MRI-linac: the increased space in the MRI-linac over MARDOS allows the detectors to be encased in their normal packaging, and standard phantom sizes can be utilised for the provision of full scatter conditions. Another consideration for furthering the project is to investigate the DMG silicon array detector, which is sensitive and also has high spatial and temporal resolution. The main hurdle to consider in the MRI-linac is the noise for both dosimetry and imaging. Noise reduction is therefore a
major factor for the realisation of dynamic dosimaging.

Use of the M512 on the 1 T inline MRI-linac represents the first successful introduction of this unique high spatial and temporal resolution device on the first MRI-linac in Australia. The device shows promise for use on other MRI-linacs in Australia and overseas as they are introduced.
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