COPYRIGHT AND USE OF THIS THESIS

This thesis must be used in accordance with the provisions of the Copyright Act 1968.

Reproduction of material protected by copyright may be an infringement of copyright and copyright owners may be entitled to take legal action against persons who infringe their copyright.

Section 51 (2) of the Copyright Act permits an authorized officer of a university library or archives to provide a copy (by communication or otherwise) of an unpublished thesis kept in the library or archives, to a person who satisfies the authorized officer that he or she requires the reproduction for the purposes of research or study.

The Copyright Act grants the creator of a work a number of moral rights, specifically the right of attribution, the right against false attribution and the right of integrity.

You may infringe the author’s moral rights if you:

- fail to acknowledge the author of this thesis if you quote sections from the work
- attribute this thesis to another author
- subject this thesis to derogatory treatment which may prejudice the author’s reputation

For further information contact the University’s Director of Copyright Services

sydney.edu.au/copyright
On the development of a novel detector for simultaneous imaging and dosimetry in radiotherapy

Samuel Joseph Blake

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

Doctor of Philosophy

Institute of Medical Physics
School of Physics
University of Sydney
November 2014
Statement of originality

This thesis is submitted to the University of Sydney in fulfilment of the requirement for the Degree of Doctor of Philosophy.

The work presented in this thesis is, to the best of my knowledge and belief, original except as acknowledged in the text. I hereby declare that I have not submitted this material, either in full or in part, for a degree at this or any other institution.

[Signature: San-Blake]  Date: 4/11/2014
Acknowledgements

First and foremost, I want to acknowledge my primary supervisor A/Prof Zdenka Kuncic for her mentorship over the past four years. To say that this thesis would not have been possible without her support would be an immense understatement. Her enthusiasm for research and teaching is inspirational and her motivation and confidence in my ability have been invaluable assets.

I would also like to thank my supervisory team: Dr Philip Vial, Dr Lois Holloway and Dr Aimee McNamara. Phil’s efforts to acquire the prototype scintillators, his EPID dosimetry expertise and dedication to the project as a whole have been instrumental throughout. I am indebted to Lois for her commitment to this research, her clinical expertise and constant words of encouragement. And I am grateful to Aimee for her guidance as both a mentor and a friend.

I would like to express my gratitude to Prof Clive Baldock and Ms Eve Teran for their support during my transition into the PhD program as an international student. I am thankful to Prof Peter Greer, Prof Paul Keall and Prof David Thwaites for the expertise and mentorship they provided throughout my candidacy. I would also like to acknowledge The University of Sydney and the Institute of Medical Physics for scholarship support, as well as the Liverpool and Macarthur Cancer Therapy Centres for additional financial support. To my colleagues at the IMP, thank you for making room 418 such a friendly, enjoyable environment to be a part of.

I am indebted to my high school physics teacher Mr Bob Turcke for instilling in me an enthusiasm for physics and to my undergraduate honours supervisor Dr John Schreiner and supervisor at Sunnybrook Hospital Dr Melanie Davidson for introducing me to the world of medical physics.
To my mom, Dave, Ben, Ravi, Chandra and Aarani – I am forever grateful for the unwavering support that you have shown me since making the decision to study overseas. To Auntie Laurie and Uncle Kenny, thank you for the roof over my head and encouragement that made my initial journey from a small township to the big city possible. And thank you to my partner Uma for your limitless patience, whose own achievements keep pushing me to do better and for your companionship during our adventure from the Great White North to the Land Down Under.

Lastly, thank you to my dad for having left me a seemingly endless supply of luck. To him I dedicate this thesis.
Abstract

Radiotherapy uses x-ray beams to deliver prescribed radiation doses that conform to target anatomy and minimise exposure of healthy tissue. Accuracy of dose delivery is essential, thus verification of dose distributions in vivo is desirable to monitor treatments and prevent errors from compromising patient outcomes.

Electronic portal imaging devices (EPIDs) are commonly used x-ray imagers, however their non water-equivalent response complicates use for dosimetry. In this thesis, a Monte Carlo (MC) model of a standard EPID was developed and extended to novel water-equivalent configurations based on prototypes in which the high atomic number components were replaced with an array of plastic scintillator fibres. The model verified that full simulation of optical transport is not necessary to predict the standard EPID dose response, which can be accurately quantified from energy deposited in the phosphor screen. By incorporating computed tomography images into the model, its capacity to predict portal dose images of humanoid anatomy was also demonstrated.

The prototype EPIDs water-equivalent dose response was characterised experimentally and with the MC model. Despite exhibiting lower spatial resolution and contrast-to-noise ratio relative to the standard EPID, its image quality was sufficient to discern gross anatomical structures of an anthropomorphic phantom. Opportunities to improve imaging performance while maintaining a water-equivalent dose response were identified using the model. Longer fibres increased efficiency and use of an extra-mural absorber maximised spatial resolution. Optical coupling between the scintillator fibres and the imaging panel may further improve performance.

This thesis demonstrates the feasibility of developing a next-generation EPID for simultaneous imaging and dosimetry in radiotherapy. Such a detector could monitor treatment deliveries in vivo and thereby facilitate adaptations to treatment plans in order to improve patient outcomes.
## Contents

**Academic Contributions**  \( \text{xiii} \)

**List of Figures**  \( \text{xxix} \)

**List of Tables**  \( \text{xxv} \)

**List of Abbreviations and Symbols**  \( \text{xxvii} \)

### 1 Introduction

References  \( \text{4} \)

### 2 Literature Survey

2.1 Radiotherapy in cancer treatment  \( \text{8} \)

2.1.1 History of external beam radiotherapy  \( \text{8} \)

2.1.2 Modern external beam radiotherapy  \( \text{10} \)

2.1.3 Advanced radiotherapy techniques  \( \text{13} \)

2.2 EPID imaging in radiotherapy  \( \text{17} \)

2.2.1 First clinical EPIDs  \( \text{17} \)

2.2.1.1 Camera-mirror-lens based EPID  \( \text{18} \)

2.2.1.2 Scanning matrix ionization chamber EPID  \( \text{19} \)

2.2.2 Direct detection $a$-Si EPIDs  \( \text{20} \)

2.2.3 Indirect detection $a$-Si EPIDs  \( \text{21} \)

2.2.4 Novel detector configurations  \( \text{23} \)
4 Characterization of optical transport effects on EPID dosimetry using Geant4
4.1 Introduction .................................................. 111
4.2 Methods and Materials ........................................ 113
  4.2.1 Description of the Monte Carlo model .................... 113
    4.2.1.1 6 MV photon source .............................. 113
    4.2.1.2 EPID geometry and electromagnetic physics .... 114
    4.2.1.3 Optical physics .................................. 116
  4.2.2 Optical transport analysis .............................. 119
  4.2.3 EPID dose response .................................... 120
    4.2.3.1 Simulation dose response ......................... 120
    4.2.3.2 Experimental dose response and model validation 122
4.3 Results and Discussion ..................................... 123
  4.3.1 Optical transport analysis ............................. 123
  4.3.2 EPID model dose-response ............................. 127
    4.3.2.1 Field size output factors ....................... 127
    4.3.2.2 Relative dose profiles ......................... 128
4.4 Conclusion .................................................. 132
References ........................................................ 134

5 Monte Carlo simulation of the transit dosimetric response of an a-Si EPID
5.1 Introduction .................................................. 142
5.2 Monte Carlo model and detector geometry .................... 143
  5.2.1 6 MV photon source .................................... 143
  5.2.2 EPID geometry and physics processes .................. 143
  5.2.3 Phantom definitions .................................. 144
5.3 Simulated dosimetric quantities ............................ 144
  5.3.1 Transmission factors ................................. 145
Contents

5.3.2 Field size output factors and relative dose profiles . . . . . . . 145
5.3.3 Projection phantom portal dose image . . . . . . . . . . . . . 146
5.4 Experimental measurements and model validation . . . . . . . . . 146
5.5 Results and Discussion . . . . . . . . . . . . . . . . . . . . . . . 146
  5.5.1 Transmission factors . . . . . . . . . . . . . . . . . . . . . . 146
  5.5.2 Field size response . . . . . . . . . . . . . . . . . . . . . . . 147
  5.5.3 Relative dose profiles . . . . . . . . . . . . . . . . . . . . . . 148
  5.5.4 Projection phantom portal dose image . . . . . . . . . . . . 149
5.6 Conclusion . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . 150
References . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . . 151

6 Characterization of a novel EPID designed for simultaneous imag-
ing and dose verification in radiotherapy 153
  6.1 Introduction . . . . . . . . . . . . . . . . . . . . . . . . . . . . . 157
  6.2 Methods and Materials . . . . . . . . . . . . . . . . . . . . . . . 160
    6.2.1 Detector design and settings . . . . . . . . . . . . . . . . . . 160
      6.2.1.1 Standard EPID configuration . . . . . . . . . . . . . 160
      6.2.1.2 Prototype EPID configuration . . . . . . . . . . . . . 162
    6.2.2 Detector sensitivity, linearity and pixel noise . . . . . . . 163
    6.2.3 Dose response evaluation . . . . . . . . . . . . . . . . . . . 164
      6.2.3.1 Off-axis response . . . . . . . . . . . . . . . . . . . . . 165
      6.2.3.2 Field size response . . . . . . . . . . . . . . . . . . . . 166
      6.2.3.3 Transmission factors . . . . . . . . . . . . . . . . . . . 166
    6.2.4 Image quality evaluation . . . . . . . . . . . . . . . . . . . 167
      6.2.4.1 QC-3V Phantom . . . . . . . . . . . . . . . . . . . . . . 167
      6.2.4.2 Anthropomorphic phantom . . . . . . . . . . . . . . . 168
  6.3 Results and Discussion . . . . . . . . . . . . . . . . . . . . . . . 168
    6.3.1 Detector sensitivity, linearity and pixel noise . . . . . . . 168
    6.3.2 Dose response evaluation . . . . . . . . . . . . . . . . . . . 170
      6.3.2.1 Off-axis response . . . . . . . . . . . . . . . . . . . . . 170
      6.3.2.2 Field size response . . . . . . . . . . . . . . . . . . . . 172
      6.3.2.3 Transmission factors . . . . . . . . . . . . . . . . . . . 172
    6.3.3 Image quality evaluation . . . . . . . . . . . . . . . . . . . 175
## Contents

6.3.3.1 QC-3V phantom ........................................ 175  
6.3.3.2 Anthropomorphic phantom .......................... 176  
6.4 Conclusion .................................................. 177  
References ..................................................... 179  

7 Optimisation of the imaging and dosimetric characteristics of an EPID employing plastic scintillating fibres using Monte Carlo simulations ............. 187  
7.1 Introduction .................................................. 190  
7.2 Methods and Materials .................................... 192  
7.2.1 Monte Carlo source model and EPID geometry ........ 192  
7.2.2 Simulated physics processes ............................ 194  
7.2.3 Simulated imaging and dosimetric quantities .......... 196  
7.2.3.1 Imaging performance evaluation ................... 197  
7.2.3.2 Dose response evaluation ............................ 197  
7.3 Results ..................................................... 198  
7.3.1 Detection efficiency .................................... 198  
7.3.2 Detector sensitivity ..................................... 199  
7.3.3 Modulation Transfer Function .......................... 201  
7.3.4 Field size response ...................................... 202  
7.3.5 Relative dose profiles ................................... 202  
7.4 Discussion .................................................. 205  
7.5 Conclusions ................................................ 208  
References ..................................................... 210  

8 A next-generation EPID for simultaneous imaging and dosimetry in radiotherapy ........................................... 215  
8.1 Introduction .................................................. 218  
8.2 Methods and Materials .................................... 220  
8.2.1 Description of the prototype detector ................. 220  
8.2.2 Overview of the Monte Carlo model ................... 222  
8.2.3 Experimental measurements ............................ 225  
8.2.3.1 Line spread and modulation transfer functions ... 226
## Contents

8.2.3.2 Field size output factors .................................. 228  
8.2.3.3 Relative dose profiles ...................................... 229  
8.2.4 Simulated quantities ........................................... 230  
  8.2.4.1 Line spread and modulation transfer functions .......... 230  
  8.2.4.2 Field size output factors .................................. 231  
  8.2.4.3 Relative dose profiles ..................................... 231  
8.3 Results and discussion ........................................... 232  
  8.3.1 Line spread and modulation transfer functions .......... 232  
  8.3.2 Field size response .......................................... 234  
  8.3.3 Relative dose profiles ....................................... 235  
8.4 Conclusions ...................................................... 238  
References .......................................................... 240  

9 Conclusions and future work ....................................... 247  

Appendix .......................................................... 251  

xii
Chapter 4 through 8 are based on the following publications:


0. Academic Contributions


Some of the work presented in this thesis has also contributed to the following co-authored publication:


Awards

1. Most Outstanding Presentation in the “Biomarkers & diagnosis” session, Postgraduate Cancer Research Symposium, November 27 2012

2. 1st prize award for talk given at the 6th Student Research Symposium of the ACT/NSW Branch of the Australasian College of Physical Scientists and Engineering in Medicine, December 1 2011

3. Abstract titled “An investigation into optical photon transport effects on electronic portal imaging performance using GEANT4” selected for oral presentation in the Young Investigators Symposium of The 2011 Joint American Association of Physicists in Medicine (AAPM)/Canadian Organization of Medical Physicists (COMP) annual meeting, August 2011

4. Most Outstanding Presentation in the “Cancer Therapeutics” session, Postgraduate Cancer Research Symposium, December 3 2010

Published abstracts


Conference presentations


5. S.J. Blake, A.L. McNamara, P. Vial, L. Holloway, P. Greer, Z. Kuncic. “Investigation of optical transport within a novel plastic scintillator im-
0. Academic Contributions


## List of Figures

<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>2.1</td>
<td>The first x-ray radiograph showing the bones of W. Röntgen’s wife’s hand.</td>
</tr>
<tr>
<td>2.2</td>
<td>Schematic of the key components of a modern linear accelerator.</td>
</tr>
<tr>
<td>2.3</td>
<td>Photo of a multileaf collimator.</td>
</tr>
<tr>
<td>2.4</td>
<td>IMRT treatment plan for a patient with prostate cancer.</td>
</tr>
<tr>
<td>2.5</td>
<td>Schematic illustrating the design of the camera-mirror-lens based EPIDs.</td>
</tr>
<tr>
<td>2.6</td>
<td>Schematic illustrating the pixelated design and electronic components of modern $\alpha$-Si based AMFPI EPIDs.</td>
</tr>
<tr>
<td>2.7</td>
<td>Experimental setup of an $\alpha$-Si EPID in a direct-detection configuration using solid water buildup and backscatter.</td>
</tr>
<tr>
<td>2.8</td>
<td>Schematic and physical prototype of a thick, segmented phosphor scintillator.</td>
</tr>
<tr>
<td>2.9</td>
<td>Physical prototype of a thick, CsT:TI crystal scintillator.</td>
</tr>
<tr>
<td>2.10</td>
<td>Schematic of a proposed Čerenkov radiation portal imaging device.</td>
</tr>
<tr>
<td>2.11</td>
<td>Mass attenuation coefficients for $\text{H}_2\text{O}$, $\text{Gd}_2\text{O}_2\text{S}$ and BC430 plastic scintillator over the radiotherapy energy range.</td>
</tr>
<tr>
<td>2.12</td>
<td>Treatment delivery errors detectable using EPID dosimetry.</td>
</tr>
<tr>
<td>2.13</td>
<td>Assortment of plastic scintillator blocks and fibers that may be used for x-ray detection.</td>
</tr>
<tr>
<td>2.14</td>
<td>Estimating $\pi$ using pseudorandom number generation and the acceptance-rejection method.</td>
</tr>
<tr>
<td>2.15</td>
<td>Schematic illustrating the photoelectric effect.</td>
</tr>
</tbody>
</table>
## List of Figures

2.16 Schematic illustrating Compton scattering. .............................. 46
2.17 Schematic illustrating an x-ray undergoing $e^-/e^+$ pair production. 48
2.18 Schematic illustrating an electron passing near an atomic nucleus and emitting a bremsstrahlung x-ray. ................................. 49
2.19 Schematic illustrating the UNIFIED model’s treatment of optical boundaries with microfacet surface structure ......................... 53
2.20 GEANT4 visualization of the ATLAS detector at the LHC. ........... 56

3.1 Schematic of the Elekta Synergy linear accelerator photon beam model developed using BEAMnrc. ........................................ 91
3.2 Experimental and simulated PDD curves for a 6 MV photon beam in water. .......................................................... 95
3.3 Experimental and simulated relative dose profiles for a 6 MV photon beam at $d_{\text{max}}$ in water. ................................. 96
3.4 Experimental and simulated field size output factors for a 6 MV photon beam at 10 cm depth in water. ................................. 97
3.5 Comparison of measured PDD in water against simulated data for varying source electron energy. ................................. 98
3.6 Local percent differences between simulated PDD for varying source electron energy and measured data. ................................. 99
3.7 Comparison of measured dose profiles in water against simulated data for varying source electron radial intensity. ................................. 101
3.8 RMS differences between simulated profiles for varying source electron radial intensity and measured data. ................................. 102
3.9 Comparison of measured dose profiles in water against simulated data for varying source electron mean angular spread. ................................. 103
3.10 Local percent differences between simulated dose profiles for varying source electron mean angular spread and measured data. ................................. 104

4.1 Schematic of the key layers of the EPID model. .......................... 115
4.2 Illustration of particle tracks in the standard EPID model during a simulated event. ........................................ 118
4.3 Comparison of simulated standard EPID point spread functions scoring energy deposition and optical absorption. ................................. 124
4.4 Variations in calculated $\Delta x_{opt}$ for select values of $l_{phos}$, $\mu_{phos}$ and $SY$ relative to $\Delta x_{opt}^{REF}$ ................................. 125
4.5 Illustration of optical photon tracks in the standard EPID model during a simulated event for varying Rayleigh scattering length. 126
4.6 Standard EPID $PSF_{Tot}$ for variations in selected optical transport parameters. ................................................................. 127
4.7 Comparison of experimentally measured and simulated standard EPID field size output factors. ........................................ 129
4.8 Experimental and simulation standard EPID relative dose profiles. 131

5.1 [Schematic of the key layers of the standard EPID model.] Schematic of the key layers of the EPID model (not to scale). ............... 144
5.2 Measured and simulated standard EPID transmission factors. ...... 147
5.3 Measured and simulated standard EPID field size output factors. 148
5.4 Measured and simulated relative dose profiles using the standard EPID model with a 20 cm thick solid water phantom. ............... 149
5.5Measured and simulated portal images of an anthropomorphic head phantom. ................................................................. 150

6.1 Schematic illustration of the main components of the standard and prototype EPIDs. ............................................................. 161
6.2 Photograph of the segmented plastic scintillator array used in the experimental EPID ............................................................. 163
6.3 Comparison of pixel values per MU measured using the standard and prototype EPIDs ......................................................... 169
6.4 Comparison of nontransit and transit off-axis ratios measured using the MatriXX detector, standard EPID and experimental EPID. 171
6.5 Comparison of nontransit and transit field size output factors measured using a MatriXX detector, standard EPID and experimental EPID. ................................................................. 173
6.6 Comparison of transmission factors measured using a MatriXX detector, the standard EPID and the experimental EPID. ........... 174
6.7 Images of a QC-3V image quality phantom taken with the standard and experimental EPIDs. ...................................................... 175
List of Figures

6.8 Projection images of an anthropomorphic phantom taken with the standard and experimental EPIDs. ........................................ 177

7.1 Schematic of the PSA-EPID Monte Carlo model. ................. 193
7.2 Simulated prototype EPID x-ray detection efficiency as a function of energy. ................................................................. 200
7.3 Simulated prototype EPID detector sensitivity as a function of x-ray energy. ................................................................. 201
7.4 Simulated prototype EPID modulation transfer functions. .... 203
7.5 Comparison of simulated prototype EPID and dose in water field size output factors. ...................................................... 204
7.6 Comparison of simulated prototype EPID and dose in water relative dose profiles using reference and non-reference model parameters. 206

8.1 Photo and schematic of the second-generation prototype PSA. ... 221
8.2 GEANT4 model of the plastic scintillating fibre array and underlying a-Si EPID. ............................................................... 223
8.3 Experimental setup used to measure the PSA-EPID LSF via the angled slit technique. ...................................................... 227
8.4 Comparison of experimentally measured and simulated LSFs for the PSA-EPID. ............................................................. 233
8.5 Comparison of experimentally measured and simulated MTFs for the PSA-EPID. ............................................................. 235
8.6 Comparison of field size output factors simulated for the PSA-EPID and measured using a MatriXX array and the prototype PSA-EPID. ............................................................. 236
8.7 Uncorrected relative dose profiles measured using the prototype EPID and simulated using the PSA-EPID model. ............ 237
8.8 Corrected relative dose profiles measured using the prototype EPID and simulated using the PSA-EPID model. .................. 238

1 Relative dose profiles simulated for select values of μ and τ. .... 280
2 Photo of the second-generation prototype array of plastic scintillating fibres. ................................................................. 285
3 Field size output factors measured using a MatriXX array compared to those measured and simulated using the PSA-EPID prototype. .......................................................... 286
## List of Tables

2.1 Summary of the main particle transport processes. ................ 43

3.1 Summary of the optimised source electron beam parameters used in the validated linac photon beam model. .............. 91

3.2 Summary of the radiation transport and variance reduction parameters used in the validated linac photon beam model. ...... 92

4.1 Summary of the reference optical physics properties specified in the EPID model. .............................. 116

4.2 Analysis of the sensitivity of the EPID model to changes in optical transport parameters. ............................... 120

4.3 Summary of the FWHM and FWTM for $\text{PSF}_{\text{Edep}}^{\text{REF}}$ and $\text{PSF}_{\text{Tot}}$ calculated by varying corresponding optical transport parameters. 128

6.1 Quantitative comparison of the CNR and spatial resolution of the standard and experimental EPID configurations. ........ 176

7.1 Summary of the reference properties of the PSA-EPID fibres and photodiodes used in the MC model. ....................... 195

7.2 Analysis of the sensitivity of the PSA-EPID model to changes in geometric and optical transport parameters. .............. 196

8.1 Summary of the reference properties of the PSA-EPID fibres and photodiodes used in the MC model. ....................... 224
List of Abbreviations and Symbols

Chemical Formulae

- $a$-Se: Amorphous Selenium
- $a$-Si: Amorphous Silicon
- BGO: Bismuth Germanium Oxide
- Co-60: Cobalt-60
- CsI:Tl: Thallium-doped Cesium Iodide
- $\text{Gd}_2\text{O}_2\text{S}:\text{Tb}$: Terbium-doped Gadolinium Oxysulfide

Acronyms

- 2D: Two Dimensional
- 3D: Three Dimensional
- 3D-CRT: 3D Conformal Radiotherapy
- AAPM: American Association for Physicists in Medicine
- AMFPI: Active-Matrix Flat Panel Imager
- ARPANSA: Australian Radiation Protection and Nuclear Safety Agency
- BTE: Boltzmann Transport Equation
- CBCT: Cone Beam Computed Tomography
<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Full Form</th>
</tr>
</thead>
<tbody>
<tr>
<td>CPU</td>
<td>Central Processing Unit</td>
</tr>
<tr>
<td>DBS</td>
<td>Directional Bremsstrahlung Splitting</td>
</tr>
<tr>
<td>DQE</td>
<td>Detective Quantum Efficiency</td>
</tr>
<tr>
<td>EBRT</td>
<td>External Beam Radiotherapy</td>
</tr>
<tr>
<td>EMA</td>
<td>Extra-mural Absorber</td>
</tr>
<tr>
<td>EPID</td>
<td>Electronic Portal Imaging Device</td>
</tr>
<tr>
<td>GPU</td>
<td>Graphical Processing Unit</td>
</tr>
<tr>
<td>GUI</td>
<td>Graphical User Interface</td>
</tr>
<tr>
<td>HT</td>
<td>Helical Tomotherapy</td>
</tr>
<tr>
<td>ICRU</td>
<td>International Commission on Radiation Units and Measurements</td>
</tr>
<tr>
<td>IGRT</td>
<td>Image Guided Radiotherapy</td>
</tr>
<tr>
<td>IMAT</td>
<td>Intensity Modulated Arc Therapy</td>
</tr>
<tr>
<td>IMRT</td>
<td>Intensity Modulated Radiotherapy</td>
</tr>
<tr>
<td>kV</td>
<td>Kilovoltage</td>
</tr>
<tr>
<td>LET</td>
<td>Linear Energy Transfer</td>
</tr>
<tr>
<td>LHC</td>
<td>Large Hadron Collider</td>
</tr>
<tr>
<td>Linac</td>
<td>Linear Accelerator</td>
</tr>
<tr>
<td>MC</td>
<td>Monte Carlo</td>
</tr>
<tr>
<td>MEMS</td>
<td>Micro-Electro-Mechanical System</td>
</tr>
<tr>
<td>MFP</td>
<td>Mean Free Path</td>
</tr>
<tr>
<td>MLC</td>
<td>Multi-Leaf Collimator</td>
</tr>
<tr>
<td>MTF</td>
<td>Modulation Transfer Function</td>
</tr>
<tr>
<td>Abbreviation</td>
<td>Description</td>
</tr>
<tr>
<td>--------------</td>
<td>--------------------------------------------</td>
</tr>
<tr>
<td>MU</td>
<td>Monitor Units</td>
</tr>
<tr>
<td>MV</td>
<td>Megavoltage</td>
</tr>
<tr>
<td>NPS</td>
<td>Noise Power Spectrum</td>
</tr>
<tr>
<td>OAR</td>
<td>Organs At Risk</td>
</tr>
<tr>
<td>PDD</td>
<td>Percent Depth Dose</td>
</tr>
<tr>
<td>PDF</td>
<td>Probability Density Function</td>
</tr>
<tr>
<td>PSA</td>
<td>Plastic Scintillator Array</td>
</tr>
<tr>
<td>PSD</td>
<td>Plastic Scintillation Detector</td>
</tr>
<tr>
<td>PVT</td>
<td>Polyvinyltoluene</td>
</tr>
<tr>
<td>QA</td>
<td>Quality Assurance</td>
</tr>
<tr>
<td>RF</td>
<td>Radiofrequency</td>
</tr>
<tr>
<td>RMS</td>
<td>Root Mean Square</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>TPS</td>
<td>Treatment Planning System</td>
</tr>
<tr>
<td>VMAT</td>
<td>Volumetric Modulated Arc Therapy</td>
</tr>
</tbody>
</table>
Modern radiotherapy is one of the most commonly used techniques for treating cancer. It has been estimated that approximately 52% of new cancer cases are indicated for external beam radiotherapy (EBRT) at least once during their course of treatment \(^1\). EBRT involves the use of a medical linear accelerator to generate beams of high-energy x-rays that penetrate through the patient, targeting the solid tumour. Owing to the highly proliferative nature of cancer cells and their higher sensitivity to radiation damage compared to healthy cells, EBRT has proven to be an effective means of controlling tumour growth. Through the development of new technologies and improved x-ray beam delivery techniques, radiotherapy has evolved through several generations. Earlier methods including 3D conformal radiotherapy (3DCRT) are being replaced by more recently developed techniques including intensity-modulated radiotherapy (IMRT) and image-guided radiotherapy (IGRT). These modern techniques enable the delivery of beams that are both dosimetrically precise and capable of conforming spatially to complex target geometries. For instance, current ICRU guidelines for
1. Introduction

IMRT recommend that in low dose gradient regions at least 85% of the target volume receives an absorbed-dose within 5% of prescription \(^1\). Electronic portal imaging devices (EPIDs) are flat-panel x-ray detectors that are frequently used for patient imaging in modern radiotherapy owing to their high spatial resolution and real-time readout capabilities. Most linear accelerator vendors now supply EPIDs as fully integrated systems that come fixed to the accelerator’s gantry, greatly facilitating their routine clinical use. EPIDs can be used either with the megavoltage (MV) therapy beam or, for improved image contrast, with an on-board kilovoltage (kV) x-ray tube. EPIDs are most commonly used to verify patient setup prior to treatment by acquiring portal images in cases where bony landmarks serve as suitable surrogates for tumour position \(^3\). If significant changes in patient anatomy are observed, new treatment plans may be generated to adapt therapy accordingly. More recently, use of implantable fiducial markers has enabled EPIDs to image tumour motion in patients in real time using the therapy beam \(^4\).

Despite the current ability to deliver highly conformal MV x-ray beams and guide treatment using modern imaging technologies, there are still uncertainties arising in the radiotherapy process that limit the ability to predict patient outcomes. Arguably one of the largest uncertainties is the current inability to quantify the actual dose delivered to the patient. Variations in patient position and internal anatomy can have an important impact on the dose delivered to the tumour and surrounding healthy tissues \(^5\). There is also growing evidence showing that the quality of radiotherapy delivered may affect patient outcomes, including local control, toxicity rates and overall survival \(^6,7\). Owing primarily to a lack of suitable commercially available dosimeters, in vivo dosimetry is not routine clinical practice in most centres and real-time dose monitoring is limited almost exclusively to single point skin and intracavitary measurements. The ability to perform routine in vivo patient dosimetry in two or even three dimensions is highly desirable as it will verify correct treatment delivery, detect harmful treatment errors that may otherwise go undetected and identify patients that may benefit from treatment adaptations \(^8,9\).

Interest in using EPIDs for radiotherapy dosimetry has been growing since their clinical inception in the 1990s \(^8–10\). However their design – which as pre-
viously mentioned has been optimised for imaging applications – has severely limited their routine clinical use as dosimeters. Several groups have investigated the use of EPIDs for dosimetry, either by developing methods to adapt currently available detectors for this purpose or by designing novel detectors specifically to act as dosimeters rather than imagers. The former scenario typically necessitates a complex detector characterisation and calibration scheme along with custom software to convert portal images into dose images. The latter scenario has seen detector prototypes that, while capable of performing accurate patient dosimetry, suffer from decreased x-ray detection efficiency thus inhibiting their use for imaging. In both cases, proposed detectors have not been suitable for applications in both imaging and dosimetry. Furthermore, lack of a user friendly, commercially available product precludes these methods from being implemented routinely in clinical practice for most centres.

This thesis presents work on the development of a next-generation EPID designed for simultaneous imaging and dose verification applications in radiotherapy. Monte Carlo (MC) simulations of radiation transport were used to develop several different detector models. Experimental measurements were also conducted with research prototypes and clinical MV radiotherapy photon beams to validate the MC models and assess imaging and dosimetry performance. A MC model of the currently available standard clinical EPID was first developed and the different physical processes operating within the detector were characterised. This model was later extended to novel configurations based on experimental prototypes designed by our research group. The novel prototypes replaced the high atomic number materials within the standard EPID with an array of plastic scintillator fibres to make the detectors more suitable for dosimetry. By optimising specific geometrical and material properties of the plastic scintillator array, the novel EPID’s detection efficiency and spatial resolution may be comparable to that of current detectors optimised for imaging. This work therefore demonstrates the feasibility of developing a next-generation EPID for simultaneous imaging and dosimetry in radiotherapy, which would enable the measurement of dose being delivered to patients in vivo. This information could then be used as a quality assurance tool to monitor treatment accuracy and make necessary adaptations to the treatment plan in order to improve patient outcomes.
References


2

Literature Survey

This literature survey discusses the topics forming the foundation upon which the work presented in this thesis is based. The most important developments in these disciplines are summarized with relevant studies referenced throughout. This chapter is organised into five sections, each of which is further divided into relevant subsections. Section 2.1 concerns radiotherapy within the broad context of cancer therapy and includes an overview of the history and technological progression of this field. Section 2.2 provides an overview of electronic portal imaging technology, beginning with a discussion of the earliest detectors and progressing towards the latest, most novel designs. Radiation dosimetry is discussed in Section 2.3, including the range of detectors related to this work and their common clinical applications. Section 2.4 gives a technical overview of the principles behind Monte Carlo simulation techniques and specifically their applications in radiation transport physics. Finally, Section 2.5 concludes this literature survey by providing the motivation behind this work and the primary aims in the individual studies making up the subsequent chapters of this thesis.
2. Literature Survey

2.1 Radiotherapy in cancer treatment

2.1.1 History of external beam radiotherapy

Many are familiar with Wilhelm Röntgen’s famous 1895 paper in which he described, for the first time, a new kind of ray that he discovered incidentally while operating a Crooke’s tube\[^1\]. Despite having covered the Crooke’s tube in opaque, black cardboard, the “X-rays” (as Röntgen called them, for the sake of brevity) were capable of passing through an assortment of objects and inducing florescence on a nearby fluorescent screen. The first ever x-ray radiograph – an image Röntgen took of his wife’s hand (see Figure 2.1) – was published in this famous paper. As a result of his discovery, Röntgen earned the first ever Nobel Prize in Physics in 1901.

Röntgen’s findings were the first of several important discoveries made in radiation physics over an exceptionally short period of time. Also in 1895, Becquerel discovered radioactivity\[^2\] and by 1898 Pierre and Marie Curie reported the discovery of radium\[^3,4\]. Even throughout this early period of discovery, the biological effects of radiation were of major interest\[^5\]. In fact, a woman with breast cancer was reportedly treated with x-ray radiation as far back as 1896\[^4,6\].

Now, over 100 years later, these invisible rays serve several crucial roles within the medical fields of diagnostic radiology and radiotherapy.

While the earliest radium therapy trials date back to the early 1900s – not long after x-rays and radium were even discovered – the birth of modern megavoltage (MV) radiotherapy took place several decades later\[^7\]. The high cost and limited availability of radium at that time resulted in a much slower development of radium therapy relative to kilovoltage (kV) therapies using x-ray tubes. However as a direct consequence of the lower kV x-ray energies, it was eventually realised that radium offered a more practical means of treating deep-seated tumours owing to its greater depth dose. To improve depth dose further it became common practice to increase the distance between the radium source and the patient\[^8\] – a practice that led to the development of teleradium machines (the prefix tele- roughly translates to mean ‘at a distance’). The paper “The Race For Megavoltage” by Robison gives a more detailed account of these technological
advances, including a list of some of the earliest teleradium units installed across several institutions throughout the first half of the 20th century\cite{7}.

In 1946, Canadian physicist Harold E. Johns attended a lecture series given by Professor William V. Mayneord in which Mayneord suggested the possibility of using radioactive cobalt-60 (Co-60) in place of radium\cite{7,9}. The suggestion came following extensive research throughout the 1930s and 1940s at Princeton and Berkeley on the neutron capture of Co-59 to yield the Co-60 isotope. The physical properties of Co-60 are well-suited to radiotherapy, with a strong activity of 1.17 and 1.33 MeV x-rays and half-life of 5.261 years\cite{9}. Later, on October 27th, 1951 the first patient was treated with a Co-60 teletherapy machine in London, Canada that was developed by Johns’ group\cite{10}.

Particle accelerators evolved along roughly the same timeframe as the early
radium and later cobalt teletherapy machines, and eventually surpassed Co-60 machines as the most widely used form of modern radiotherapy. When clinical linear accelerators (linacs) were first developed (the earliest was an 8 MV accelerator in 1952 at the Hammersmith Hospital in London), they promised several advantages over radioactive isotopes, including the ability to control the beam energy and delivery rate. The first clinical accelerator in the U.S.A. was developed at Stanford University, installed in the Stanford Department of Radiology in 1954 and treated its first patient in January 1956. Several companies began building commercial linacs and production accelerated throughout the second half of the 20th century – one major manufacturer reportedly built over 3,200 accelerators between 1962 and 1999. In most developed countries, modern clinical accelerators have increasingly replaced older Co-60 machines. Several reviews, including those written by Farmer (1962), Karzmark (1984) and Thwaites & Tuohy (2006) detail the evolution of the modern clinical linear accelerator.

2.1.2 Modern external beam radiotherapy

Linear accelerators, of the kind used in modern radiotherapy, comprise several interconnected systems. While the following description is a drastic oversimplification of the modern MV accelerator design, such accelerators typically consist of (see Figure 2.2):

- a charged particle source (e.g. an electron gun)
- a radiofrequency (RF) system (including a RF power source, accelerating waveguide and pulsed modulator)
- a beam transport system (including focussing coils and bending magnets)
- a collimation and monitoring system (including the target, primary, secondary and multi-leaf collimators, flattening filter and monitor ionization chamber(s))

A more detailed description of the design of MV linacs may be found, for example, in Van Dyk’s *The Modern Technology of Radiation Oncology*.

Electrons ejected from the electron gun are accelerated down the accelerating waveguide to energies between 4 – 25 MeV, depending on the accelerator. The
use of a variable, non-conservative electric field originating from the RF power source with a loaded waveguide that contains periodic perturbations along its length allows the electrons to be accelerated to such high energies. Once the accelerated electrons reach the end of the waveguide, magnets bend and focus the electron beam onto a tungsten target, thereby generating a bremsstrahlung x-ray beam (or an electron beam produced without the target in place). X-ray beams generated with nominal waveguide voltages of 6, 10 and 18 MV are commonly used in modern practice, resulting in beams with mean energies typically much higher than those achievable using radioactive isotopes.

![Diagram of a modern linear accelerator](image)

Figure 2.2: Schematic of the key components of a modern linear accelerator (figure reproduced from p1073 of Podgorsak (ed. J. Van Dyk), 1999).

The overarching goal of modern radiotherapy is, of course, to maximize the dose of radiation being delivered to the target (e.g. tumour) whilst minimizing the dose delivered to surrounding healthy tissues and organs at risk (OAR). The degree to which this goal may be achieved using external beam radiotherapy (EBRT) has improved in part through the evolution of modern 3D imaging and beam delivery techniques. So-called “conventional” 2D treatment planning was
the standard approach prior even well after the advent of commercially available CT around 1972[16]. This process typically involved taking a small number of planar radiographs for diagnostic and tumour localization purposes, and treatments were largely delivered using only rectangular, coplanar fields (e.g. parallel-opposed lateral fields). As a consequence of the limitations imposed by 2D imaging and such simple beam geometries, conventional treatments typically did not conform to the target shape and the dose delivered to the tumour was restricted by the relatively large amount of normal tissue within the treated volume.

Several technological advances throughout the 1970s and 1980s paved the way for 3D Conformal Radiotherapy (3D-CRT) to supersede conventional therapies. Key amongst these included, as previously mentioned, the commercial availability of 3D imaging systems including CT, along with the necessary advances in computational efficiency to facilitate 3D dose calculations and treatment planning[16]. Furthermore, the ability to shape therapy beams into geometries that more precisely conformed to the target volume was made possible through widespread use of the multileaf collimator (MLC). The MLC consists of an opposing pair of thick, tungsten jaws mounted to the head of the linear accelerator, each comprising a set of thin interlocking leaves that move linearly in one dimension (see Figure 2.3). Being able to control the position of each individual leaf in real time during beam delivery allows the creation of highly complex beam shapes, thereby improving the ability to spatially conform therapy beams to the target. One of the earliest MLCs was described by Brahme (1987)[17].

Up to this point in time, treatment planning was typically performed in a forward manner – that is information about the patient and planned beam geometry were used to calculate the expected dose delivery to the target. An optimal plan that best met criteria specified by the radiation oncologist was then selected manually from several forward-planned dose distributions[16]. However, with the advent of the MLC – which offered many more degrees of freedom for treatment plan optimization than conventional rectangular fields – and modern computational efficiency, a new approach to treatment planning was suggested by Brahme in 1988[18]. The technique, which later came to be known as inverse planning, essentially begins with the desired patient dose distribution and uses a computer-based optimization algorithm to calculate the beam shapes and intensities to best
achieve this distribution\cite{16,18,19}.

Intensity-modulated radiotherapy (IMRT) improves upon 3D-CRT most notably by using the MLC to modulate both the shape and intensity of the therapy beam. Typically, the beam intensity may be increased in areas containing the target volume and decreased in areas containing healthy tissues and OAR\cite{18,20}. To compensate for regions that may be over- or under-dosed as a result of this intensity modulation, beams may be delivered from several different gantry angles (see Figure 2.4). The technique was first described\cite{21} by Brahme et al. in 1982 and was largely pioneered by Steve Webb\cite{22}. Several historical reviews have been written about IMRT, including those by the IMRT Current Working Group (2001)\cite{23}, Webb (2003, 2005)\cite{24,25} and Bortfeld (2006)\cite{20}. IMRT, combined with inverse treatment planning, is now one of the more commonly used methods of EBRT in current clinical practice\cite{20}. It is clear to imagine the advantages of IMRT over 3D-CRT, particularly when considering how the MLC is well suited to shielding healthy tissues. Several studies have quantified the improvements observed in dose distributions resulting from the use of IMRT over 3D-CRT\cite{26–28}.

2.1.3 Advanced radiotherapy techniques

While perhaps not used as frequently overall as either IMRT or 3D-CRT, there exist several advanced radiotherapy techniques that are commonly used to treat specific subgroups of patients.

Although EBRT has generally evolved beyond the use of radioisotopes such as radium and Co-60, certain radioisotopes are well established for use in brachytherapy – a therapy that uses small implantable radioisotope seeds to treat tumours internally (common radionuclides include Co-60, Cs-137 and Ir-192)\cite{29}. Brachytherapy offers the advantage of delivering very high doses in a conformal manner, albeit using an invasive procedure. Though not limited to these sites, it is most commonly used for certain prostate\cite{30}, gynaecologic and breast cancers. In some cases, it may be desirable to combine the use of brachytherapy with EBRT, such as in the case of a primary solid tumour with metastatic lymph node involvement, whereby seeds may be implanted within the primary tumour and distant metastases may be more effectively treated using EBRT. A historical re-
2. Literature Survey

Figure 2.3: Photo of a multileaf collimator (MLC) that is capable of forming complex beam shapes. Moving individual leaves dynamically during treatment facilitates intensity modulation of the beam (figure reproduced from p.R369 of Bortfeld, 2006[20].)

A view of brachytherapy technology and physics practice is given by Williamson[31].

Other advanced EBRT techniques, including intensity modulated arc therapy (IMAT) and helical tomotherapy (HT) have extended the basic principles of IMRT to more novel therapies. IMAT is a form of IMRT that continuously delivers the therapy beam while rotating in an arc around the patient. Whereas traditional IMRT delivers the treatment beam from several discrete gantry angles (usually from 5 – 9), IMAT may offer an added advantage by further spreading out the low dose radiation[32]. IMAT is also commonly referred to as volumetric modulated arc therapy (VMAT)[33]. Some studies have shown that IMAT can result in even more conformal treatment deliveries than conventional IMRT, owing to the increased number of control points available during the treatment plan optimization process[26]. HT was first described by Mackie et al. in 1993 and, as the name implies with the prefix *tomo*- translating to mean ‘slice’, it combines
2.1. Radiotherapy in cancer treatment

the principles of IMRT and CT by using a MV fan beam to treat patients in a series of slices\textsuperscript{[34]}. With the patient lying on the treatment table and moving slowly through the bore of the CT-like gantry, the therapy fan beam irradiates the target in a helical manner. The fan beam’s intensity may still be modulated using a binary MLC (i.e. leaves are either ‘open’ or ‘closed’). A more in depth description of HT is given by Olivera \textit{et al.}\textsuperscript{[35]} and a historical review is given by Mackie (2006)\textsuperscript{[36]}.

![Figure 2.4: IMRT treatment plan for a patient with prostate cancer. The colour overlay illustrates the spatial variation in planned absorbed dose (as a percentage of the prescribed dose) resulting from the intensity modulation of beams incident from five gantry angles (image courtesy of Varian Medical Systems, Inc.)](image)

Image-guided radiotherapy (IGRT) is an umbrella term used to describe advanced EBRT techniques that integrate various modern imaging technologies with radiotherapy planning and delivery. Within IGRT, pre-treatment imaging is typically used to monitor patient setup with the goal of improving the accuracy of
2. Literature Survey

treatment delivery. If using image guidance reduces the geometrical uncertainties associated with target localization, then it may be possible to reduce the margin of healthy tissue being treated around the clinical target volume. Several radiation-based and non radiation-based imaging systems are currently used in IGRT. Electronic portal imaging devices (EPIDs) are a type of radiation-based imaging system that may use either the MV therapy beam or a gantry-mounted kV x-ray source to acquire images. These images may then be used in real time to guide treatment delivery for a patient or may be reviewed offline at a later time. EPIDs are discussed more thoroughly in Section 2.2 of this chapter and have been studied in depth throughout this thesis. MV and kV cone-beam CT (CBCT) is a technique that utilizes gantry-mounted flat panel imagers opposite the x-ray source to acquire open-field images while rotating about a patient or phantom (an object designed to mimic the radiation absorption and scattering properties of a patient). Acquiring these projections enables 3D volumetric images of internal anatomy to be reconstructed and has set a new standard for imaging in IGRT. A recent review of these and other image guidance systems, including non radiation-based systems based on ultrasound and optical imaging, was published by De Los Santos et al. (2013).

Hadrontherapy is an advanced form of radiotherapy that uses hadrons, instead of photons or electrons, to deliver a dose of radiation to the target volume. The most common type of hadrontherapy is proton therapy (a historical review is given by Smith (2006) and it offers certain dosimetric advantages over standard MV x-ray EBRT. Charged particles are directly ionizing and therefore have a much higher linear energy transfer (LET) than photons. As a consequence, charged particles tend to deposit most of their energy in matter at a very well defined depth known as the Bragg Peak, after Sir William Henry Bragg who discovered this phenomenon in 1904. By modulating the energy of charged particles, a so-called Spread Out Bragg Peak may be generated that delivers a uniform dose to the target along the beam axis while minimizing the dose to tissues in front of and behind the target volume. One significant drawback to hadrontherapy, however, is its cost and as a consequence it is currently only available in a relatively small number of facilities worldwide (at present, there are no hadrontherapy facilities in Australia). As such, hadrontherapy is typically reserved for
a specific subset of patients requiring highly conformal treatments. As an example, paediatric patients receiving radiotherapy may have an increased likelihood for developing secondary malignancies later in life because of their young age. The delivery of highly conformal treatments to spare healthy tissues is therefore especially critical in these patients. A natural extension to hadrontherapy is ion therapy, whereby ions with higher atomic numbers may be used for therapy. The Heidelberg Ion Therapy Centre in Germany is one facility that offers ion therapy with species ranging from protons to oxygens. A recent review by Suit et al. (2010) compares proton and carbon ion beams.

2.2 EPID imaging in radiotherapy

EPIDs are flat panel x-ray detectors that have developed into essential tools for modern radiotherapy, so much so that it has become standard for linear accelerator vendors to supply retractable EPIDs directly mounted to their linac gantries. Their most common application is for on-line pre-treatment verification of patient setup. Portal images acquired using EPIDs may be registered with reference digitally reconstructed radiographs (DRRs) derived from treatment plans, which record the intended patient position. Discrepancies between the portal image and DRR may then be used to correct patient position. More contemporary EPID applications include intra-fraction imaging for patient position verification and real-time tumour tracking in 4D radiotherapy, pre-treatment and in vivo dosimetry and linac quality assurance (QA). The following subsections describe in detail the first clinical EPIDs, followed by the more recent direct- and indirect-detection active matrix flat panel imager (AMFPI) EPIDs, and finally several novel EPIDs currently under investigation for potential future applications in radiotherapy.

2.2.1 First clinical EPIDs

The ability to capture an image of a radiation field, or “port”, prior to or during therapy has long been recognized as an important means to verify correct treatment delivery. Portal images acquired prior to therapy may be used to
verify patient positioning and target alignment with the beam. Images acquired throughout treatment may be used to monitor treatment progression, and observe such processes as intra-fraction motion. Traditionally the most commonly used technology for acquiring portal images was a film cassette containing a thin radiographic film sandwiched between metal plates and/or phosphor screens. While films had a number of advantages, including high spatial resolution and sensitivity to low doses of radiation, one significant limitation was in their need to be processed before any useful information could be obtained from them. This time delay between acquiring a portal image and being able to use it motivated the development of an electronic means for acquiring portal images to be viewed, interpreted, and used to make clinical decisions in real time. EPIDs were therefore developed largely as an alternative technology to film cassettes, having similar benefits to film including a high spatial resolution and convenience, while improving upon film’s most significant limitation of the need for processing.

Prof. Larry Antonuk, who is one of the pioneers credited with the development of the modern AMFPI EPID, published an excellent historical review of EPIDs in 2002[56]. Early development of electronic portal imaging technologies dates back to the 1950s, with more widespread commercialization occurring throughout the late 1980s and early 1990s. Of the different early technologies developed, only two categories were in widespread clinical use over a significant period of time. These are the camera-mirror-lens based EPIDs and the scanning matrix ionization chamber EPIDs.

2.2.1.1 Camera-mirror-lens based EPID

A schematic depicting the typical configuration for a camera-mirror-lens based EPID is shown in Figure 2.5. The basic principle involves the use of a metal plate/phosphor screen (typically gadolinium oxysulfide, Gd$_2$O$_2$S:Tb) to convert x-rays transmitted through the patient into an optical wavelength signal. With a combination of mirrors and lenses, this optical signal is directed onto a camera, giving rise to the projected image. To protect the camera and its electronic components from the damaging effects of the primary radiation beam, it is placed outside of the primary field and a mirror angled at approximately 45° is used to
reflect the optical signal towards it.

Although this EPID's large area design enables large radiation fields to be imaged, only the emerging optical signal that falls within the small, projected area of the camera lens is actually captured. As a result, only $\approx 0.01 - 0.1\%$ of the light emerging from the converter actually reaches the camera, thereby limiting its detection efficiency $^{[56,57]}$. A second important disadvantage occurs from light that reflects off the mirror, re-scatters from the converter and eventually reaches the camera. This long range optical glare degrades spatial resolution and can make up more than 25% of the detected signal $^{[56]}$. While significant efforts have gone into improving the efficiency of this design, the maximum reported detective quantum efficiency (DQE) for camera-mirror-lens based EPIDs using a metal plate/phosphor screen are $\approx 1 - 3\%$ $^{[56,58]}$.

![Schematic of the camera-mirror-lens based EPIDs](image)

Figure 2.5: Schematic illustrating the design of the camera-mirror-lens based EPIDs (figure reproduced from pR41 of Antonuk, 2002 $^{[56]}$).

### 2.2.1.2 Scanning matrix ionization chamber EPID

The scanning matrix ionization chamber EPID offers certain advantages over the camera-mirror-lens based EPIDs, including a more compact design and avoidance
of the geometrical distortions resulting from long range optical glare. This type of EPID is constructed as a 2D area detector, with two planes of electrodes separated by a small gap filled with a liquid ionization chamber (e.g. 2,2,4-trimethylpentane). While each electrode plane contains a series of parallel wires, orienting the planes perpendicularly to each other gives rise to an effectively pixelated area detector.

The detector readout is performed in a linearly scanning fashion by applying a high voltage to each of the electrodes in succession and recording the resulting signal. Since only a single electrode is read out at a time, one disadvantage to this system is that the full number of detected x-ray quanta does not contribute to the measurable signal. The result is a reduced DQE, which is typically reported as being \( \approx 0.5\% \) in these detectors.

### 2.2.2 Direct detection a-Si EPIDs

Direct-detection AMFPI EPIDs are commercially available for diagnostic imaging and several authors have reported on their use in kV and MV portal imaging. Similar to the indirect-detection AMFPI EPIDs described in the following subsection, direct-detection AMFPI EPIDs employ a pixelated array of a-Si photodiodes on a glass substrate as a means of storing and reading out trapped charge to form a 2D digital image. Rather than using a metal plate/phosphor screen to convert incident x-rays into an optical signal, direct-detection EPIDs use a continuous layer of photoconductive material, such as amorphous selenium (a-Se). Each pixel within the matrix comprises a collection electrode and storage capacitor that lies underneath the layer of continuous photoconductive material. X-rays interacting within the build-up layers and a-Se generate secondary electrons which, in turn, cause the creation of electron-hole pairs within the a-Se. The freed electrons and holes then propagate under an applied bias electric field to opposite surfaces of the photoconductor. The holes are collected by the collection electrodes, which leads to a buildup of charge on individual pixels and thereby forms an x-ray image. Depending on the thickness of the photoconductive layer, DQE between approximately 1–3\% have been estimated in early studies investigating 1,000 \( \mu \)m of a-Se for a 6 MV photon beam.
2.2.3 Indirect detection \textit{a-Si} EPIDs

Following extensive research and development throughout the 1990s directed largely by Larry Antonuk’s group in Michigan\textsuperscript{[68–79]}, the first generation of the now standard amorphous silicon (\textit{a-Si}) AMFPI EPID became commercially available in the year 2000\textsuperscript{[56]}. Owing to the important applications for \textit{a-Si} EPIDs in patient imaging, the American Association for Physicists in Medicine (AAPM) published a set of clinical guidelines for portal imaging in radiotherapy not long after their commercial inception\textsuperscript{[80]}. As described by Antonuk, the distinguishing feature of the AMFPI EPIDs is the panel’s array of \textit{a-Si} photodiodes and electronics deposited on a 1 mm thick glass substrate using plasma enhanced chemical vapour deposition techniques\textsuperscript{[56]}. Each pixel typically consists of a thin-film transistor element coupled to a capacitor for storing detected charge. The conductivity of the transistor switches is controlled by the voltage applied along the Gate Control lines to render the pixels either conducting or non-conducting. The ability to read out the charge stored within each pixel is governed by the voltage applied to the Data Lines. For the greatest spatial resolution, one data line is read out at a time and reading out each line of pixels simultaneously reinitializes them. A schematic detailing the pixelated structure and electronic components is shown in Figure 2.6.

Standard, commercially available EPIDs employ an indirect-detection configuration. That is, a metal plate and phosphor screen are placed directly above the \textit{a-Si} array to act as a conversion layer, transforming the incident x-ray signal into an optical signal. Individual photodiodes absorb these optical photons, creating electron-hole pairs within the capacitor and leading to a build up of charge. The metal plate serves both to convert incident x-rays into an electronic signal that deposits energy within the phosphor, as well as to filter out low energy x-rays that would otherwise contribute towards system noise. Energy deposited in the phosphor induces scintillation events such that hundreds or thousands of optical photons may be created (in proportion to the amount of energy deposited) for a single electronic event.

This generation of \textit{a-Si} AMFPI EPIDs offered several advantages over the earlier camera-mirror-lens based and scanning matrix ionization chamber EPIDs.
Coupling the AMFPI array directly to the x-ray converter layers results in a flat, compact design that is more portable than the camera-mirror-lens based EPIDs. Similarly, a much greater proportion of the generated optical signal is captured and used in this configuration. The detectors have a high spatial resolution and can produce images in near real-time, in both radiographic (single frame) and fluoroscopic (frame sequence) readouts. The photodiodes and thin film transistors themselves are highly resistant to radiation-induced damage, and with proper shielding of the external electronics these detectors can withstand very high doses – in excess of $10^4$ Gy per year. Another practical advantage of these detectors is their ability to be manufactured in large sizes, with current commercially available detectors typically measuring $41 \times 41$ cm$^2$ in area.

The image quality of these detectors is frequently reported as being superior to many alternative EPID designs and is even comparable with that of film. They offer x-ray quantum-limited imaging and the DQE of early prototypes us-

Figure 2.6: Schematic illustrating the pixelated design and electronic components of modern $a$-Si based AMFPI EPIDs (figure reproduced from pR49 of Antonuk, 2002).
2.2. EPID imaging in radiotherapy

ing a 133 mg cm\(^{-2}\) phosphor screen was estimated to be slightly above 1\% for a 6 MV photon beam\(^{[66,83]}\) as well as for a 15 MV photon beam\(^{[66]}\). For thicker phosphors, however, it has been shown that direct detection EPIDs actually have a greater DQE for the same mass thickness of the active photoconductive layer as a result of the Lubbert’s effect present in indirect detection systems\(^{[64,66,84]}\).

The amount of image processing required for AMFPI EPID images is typically minimal, with the standard corrections applied consisting of an offset correction to account for dark current effects and a gain correction to account for variations in pixel sensitivity\(^{[85–87]}\). Even in their early stages, EPIDs have also been demonstrated as potentially suitable dosimeters\(^{[88–92]}\) and interest in using these detectors for dosimetry applications has grown in recent years (a recent review of EPID dosimetry has been published by van Elmpt \textit{et al.} (2008) summarizing their clinical applications\(^{[49]}\)). However, use of high-Z components including the Cu plate and Gd\(_2\)O\(_2\)S:Tb screen causes standard EPIDs to respond in a non water-equivalent manner – a characteristic that is not ideal for clinical dosimetry. While attempting to improve upon the imaging and/or dosimetric response of standard EPIDs, several groups have investigated more novel detector configurations and these are summarized in the following subsection.

2.2.4 Novel detector configurations

Much work continues to be directed towards both improving the current standard \(\alpha\)-Si EPID and developing novel technologies. Some of these developments, along with their respective advantages and disadvantages, are highlighted and summarized below.

2.2.4.1 Modified indirect-detection configuration

In an attempt to improve upon the dosimetric response of standard EPIDs, several studies by Vial (2008, 2009), Gustafsson (2009, 2011) and Sabet (2010, 2012) \textit{et al.} investigated modified forms of the standard indirect-detection EPID where materials above the photodiode array were replaced with water-equivalent buildup material (see Figure 2.7)\(^{[93–98]}\). Standard EPIDs over-respond to low energy radiation relative to water, an effect that is attributable to the high-Z components.
2. Literature Survey

within which photoelectric absorption events dominate\cite{96,91,99-102}. Their hypothesis was that by removing the high-Z components and replacing them with water-equivalent material, the EPID’s response should be closer to water-equivalent – a desirable characteristic for dosimetry applications.

In an initial study, measurements including linearity of response, relative dose profiles, field size factors, tissue-maximum ratios (TMRs), spatial resolution and image quality were taken using the standard EPID and its modified direct-detection form\cite{93}. Results were also compared to a standard ionization chamber. It was found that the sensitivity of the modified direct-detection EPID was reduced relative to the standard configuration by a factor of approximately 8, although the response in the modified configuration was still sufficient for all measurements. When measured at depth of dose maximum ($d_{\text{max}}$), profiles, field size factors and TMRs taken with the modified EPID were in excellent agreement with the water-equivalent ionization chamber, whereas those measured using the standard EPID exhibited the predicted over-response to low energy radiation. At greater depths, the modified direct-detection EPID response can differ from the ionization chamber measurements.

In a follow up study, the authors investigated applications using the modified direct-detection EPID for clinical IMRT dosimetry\cite{94}. In doing so, they first established which configuration of buildup and backscatter material resulted in a direct-detection response that best agreed with water-equivalent measured dosimetry data. This ideal configuration was found to use a thickness of $d_{\text{max}}$ solid water buildup without any additional backscatter. To investigate the suitability of this modified direct EPID configuration for IMRT dosimetry, the modified EPID was used to measure the dose distribution at $d_{\text{max}}$ for a head and neck IMRT field. Generally excellent agreement with the treatment planning system (TPS) was found with 98.4% of pixels meeting $\gamma$-index criterion\cite{103} of $3%/3\text{mm}$. When comparing the quality of images measured using the modified EPID to those measured using the standard EPID, it was found that for sufficiently high doses the direct-detection image quality was sufficient to visualize fiducial markers within a test phantom. However whereas the standard EPID could acquire high quality images with as little as 1 MU, the direct-detection EPID required significantly higher doses to achieve comparable image quality. The main con-
2.2. EPID imaging in radiotherapy

Figure 2.7: Experimental setup of an a-Si EPID modified into a direct-detection configuration using solid water buildup and backscatter (figure reproduced from p.4363 of Vial, 2008 [93]).

Conclusions drawn from these studies were that it is possible to modify the standard EPID to obtain a water-equivalent dose response, however the trade off in sensitivity results in a drastic reduction in image quality, particularly for low dose imaging [93,94,97,98].

2.2.4.2 Thick, segmented phosphors

Sawant et al. (2003, 2005a) have proposed a method to improve upon the low quantum efficiency of standard EPIDs by using micro-electro-mechanical system (MEMS) fabrication techniques to construct a 2D cell-like structure up to 2 mm tall and precisely aligned with the photodiode pixels of an underlying AMFPI [104,105]. This matrix of cells is then filled with a scintillating material to
2. Literature Survey

take advantage of the high detection efficiency obtained with a thick phosphor while maintaining good spatial resolution (see Figure 2.8). Three different configurations were investigated experimentally with varying scintillator thicknesses and packing densities. The scintillator being investigated was Gd$_2$O$_2$S:Tb, the same granular phosphor that is used in standard EPID phosphor screens. Imaging metrics including the detector sensitivity, modulation transfer function (MTF), noise power spectrum (NPS) and DQE were measured for each prototype.

![Figure 2.8](image.png)

Figure 2.8: (a) Schematic and (b) physical prototype of a MEMS-fabricated thick, segmented phosphor scintillator. The cells in (b) have a pitch of 508 µm (figures reproduced from p.554-556 of Sawant, 2005 [105]).

Results suggested that it was possible to improve the quantum efficiency by at least a factor of three over standard EPIDs using this design. Furthermore, comparable or even improved spatial resolution was measured using the segmented phosphors. The downside to this design appears to result from the highly depth-dependent light escape efficiency when using thick phosphors, resulting in high levels of Swank noise within these initial prototypes [106]. With the increased Swank noise, the overall DQE for the segmented phosphor systems was less than that of the standard EPID across all spatial frequencies. One proposed solution to the increased Swank noise was to replace the phosphor segments with more optically transparent materials, such as thallium-doped cesium iodide (CsI:Tl) or bismuth germanium oxide (BGO) crystal scintillators [104,105]. Another disadvantage to this approach is that the use of such thick, high-Z materials will not
2.2. EPID imaging in radiotherapy

improve upon the non-water-equivalent response of current EPIDs and, consequently, their suitability for portal dosimetry.

2.2.4.3 Thick, segmented crystals

Following their study of segmented phosphor scintillators, Sawant et al. (2005b) investigated the theoretical gains in DQE resulting from the use of thick, segmented crystal scintillators as the x-ray converter in a standard indirect-detection AMFPI EPID\textsuperscript{[107]}. This study investigated CsI:Tl and BGO crystal scintillators and Monte Carlo (MC) simulations were performed to estimate the signal properties, MTF, NPS and DQE for different crystal thicknesses, septal wall materials, and septal wall thicknesses. The authors’ main conclusion was that improvements in the EPID DQE up to a factor of 50 should be possible depending on the design specifications being considered.

![Physical prototype of an array of thick, CsT:Tl crystal scintillators](figure reproduced from p.1055 of Sawant, 2006\textsuperscript{[108]}).

In a follow up experimental investigation by the same group, a prototype CsI:Tl array with crystals 40 mm thick (see Figure 2.9) was incorporated into an AMFPI EPID and measurements were performed to calculate the prototype’s sensitivity, MTF, NPS and DQE\textsuperscript{[108]}. This prototype exhibited a zero-spatial
2. Literature Survey

frequency DQE of approximately 22% relative to just 1% when using a standard EPID. Despite this large increase in DQE, results were still lower than theoretical upper limits calculated using MC simulations, indicating the presence of Swank noise. The authors report that with further optimization of the detector design, it may be possible to improve the DQE up to $\approx 50\%$. Despite their immense potential for high DQE imaging, however, the proposed designs will retain the non-water equivalent response of current EPIDs owing to their high-Z components.

2.2.4.4 Segmented plastic scintillator

With the aim of improving both the quantum efficiency and water-equivalence of current EPIDs, Teymurazyan and Pang (2012) proposed a modified indirect-detection EPID that employs an array of plastic scintillator fibres in place of the metal plate and phosphor screen in standard EPIDs. Using MC simulations, they predicted that such an EPID could achieve a theoretical DQE of 37% for a 6 MV beam with fibers 30 cm in length. Other properties including the detection efficiency and MTF were also quantified. Unfortunately, however, the use of such thick (and consequently heavy) scintillators poses certain mechanical difficulties that may complicate their clinical practicality. Another important conclusion was that using plastic scintillator in place of the standard high-Z x-ray converter materials results in a water-equivalent dose response, which would potentially render this design suitable for portal dosimetry. A significant limitation of this study, however, is that the authors did not report any experimental data against which to validate their model.

2.2.4.5 Anti-scatter detector

Teymurazyan and Pang (2012, 2013) have also reported a particularly novel approach to improving EPID image quality, which involves replacing the metal plate and phosphor screen with a segmented array of optical fibers coupled to individual photodiode pixels. Rather than indirectly-detecting the incident x-ray beam by means of scintillation events, this detector builds upon a previously proposed Čerenkov detector. MV x-rays inducing Čerenkov events within the
optical fibers generate optical photons that propagate down the fibers and are subsequently detected (see Figure 2.10). Since the Čerenkov photons are generated at well-defined angles with respect to the incident x-ray trajectory and as a function of the x-ray energy, low energy scattered x-rays will tend to generate optical photons outside of the acceptance angle of the optical fibers. Higher energy primary x-rays will therefore be preferentially detected, reducing the overall contribution of low energy scatter to the image signal. Using MC simulations, the authors found that 20 cm long fibers reduces the contribution of scattered x-rays to the total signal by as much as 50% relative to standard EPIDs and the differential signal to noise ratio may be improved by up to 30%. Once again, however, the bulkiness of this geometry may complicate its practical clinical use. Furthermore, because this detector relies on the generation of Čerenkov radiation, it is inherently insensitive not only to low energy scattered x-rays but low energy primary x-rays as well\[^{111}\].

![Figure 2.10: Schematic of a proposed Čerenkov radiation portal imaging device using silica optical fibres (figure reproduced from p.1480 of Teymurazyan, 2013\[^{111}\]). Note that the authors’ proposed segmented plastic scintillator array comprised a similar geometry to that shown here, however the silica cores were replaced with plastic scintillator\[^{109}\].](image-url)
2.3 Dosimetry in radiotherapy

Radiation dosimetry involves the practices of measuring and calculating the dose delivered by a beam of ionizing radiation. Its primary clinical uses include ensuring the correct operation of all radiotherapy equipment, monitoring health workers’ incidental radiation exposure and verifying the correct dose delivery to patients. As such, performing accurate dosimetry is of critical importance within the field of radiation oncology. Current ICRU guidelines for IMRT recommend that in low dose gradient regions at least 85% of the target volume should receive an absorbed-dose within 5% of prescription and in high dose gradient regions at least 85% of the absorbed dose should be within 5 mm of the intended position\textsuperscript{113}. However, accuracy requirements are specific to each patient and clinical scenario and will, in general, vary depending on the dosimeter being used and the motivation for performing dosimetry in the first place. For example, the accuracy required to detect gross dosimetric errors may not be as high as that required to detect slight deviations from the treatment plan due to changes in patient anatomy.

The technological development of novel dosimeters combined with the publication of protocols to facilitate widespread standardization of dosimetry practices remain active areas of research. Only a brief overview of commonly used detectors, with a more specific focus on dosimetry using EPIDs and plastic scintillators, is given in the following subsections. For more in-depth information on topics pertaining to radiation dosimetry and detector physics, the reader is referred to several key resources including the classic textbooks by Johns and Cunningham (1983)\textsuperscript{114} and Attix (1986)\textsuperscript{115}, and more recent books by Williams and Thwaites (2000)\textsuperscript{116}, Metcalfe, Kron and Hoban (2007)\textsuperscript{117} and Khan (2010)\textsuperscript{118}.

2.3.1 Overview and commonly used detectors

As outlined throughout section 2.1.2, modern linacs are capable of delivering geometrically complex and spatially precise radiation beams. However, linacs must also deliver accurate and precise quantities of radiation to the target volume. Often, the absorbed dose within a patient or phantom is the quantity of interest and is defined as the amount of energy absorbed per unit mass from ionizing
2.3. Dosimetry in radiotherapy

Absorbed dose is measured in units of Gray (Gy, where 1 Gy = 1 J/kg).

There are several clinical situations that warrant routine dosimetry in radiotherapy and published, standardized guidelines exist for most common procedures. Examples include the calibration and quality assurance of photon and electron beams [40, 119–123], personal radiation monitoring for occupational radiation exposure [124], pre-treatment validation of TPS dose calculations [125, 126] and verification of the dose delivered at the patient level (in vivo) [92]. The ability to monitor dose delivery in vivo is clearly a useful means of ensuring patient safety and is especially useful in situations where it is critical to limit radiation exposure, for example when treating near a radiosensitive organ at risk or implantable device (see, for example, Studenski et al. (2012) [127]).

Dosimetry may be broadly classified into two categories: absolute dosimetry and relative dosimetry. Absolute dosimetry is commonly performed when calibrating the radiation output of a linac under reference conditions and allows one to convert between machine settings (such as monitor units (MU)) and absolute dose in Gy [40, 119, 120]. Comparing dose measurements under non-reference conditions to the absolute standard is known as relative dosimetry. Examples of relative dosimetry include the measurement of percent depth dose (PDD) curves, relative dose profiles or output factors. Measurements of this kind were frequently performed throughout this work to characterize detector performance and validate the response predicted using MC simulations.

There exist a wide range of dosimeters, each with their own advantages and disadvantages, available for use in modern clinical practice. Examples include ionization chambers [128], gafchromic film [129–131] and solid-state dosimeters [132–135] which include thermoluminescent dosimeters (TLDs) [136], EPIDs (discussed further in Section 2.3.2) and plastic scintillators (described further in Section 2.3.3). Only those dosimeters, including ion chambers and calorimeters, whose response can be converted to absolute dose from first principles may be used for absolute dosimetry [40]. These absolute dosimeters are typically calibrated against a primary standard – in Australia, primary standards are maintained at the Australian Radiation Protection and Nuclear Safety Agency (ARPANSA). Other relative dosimeters can only measure dose via cross-calibration against a calibrated
absolute dosimeter.

2.3.2 EPID dosimetry

While EPID dosimetry is still not in routine practice, there are several arguments to support ongoing development in this field. It has long been recognized that EPIDs offer several advantages over alternative 2D dosimeters, including:

- Ready accessibility, frequently provided by linac vendors
- High spatial resolution (typically 0.4 mm – greater than modern ion chamber/diode arrays)
- Real-time readout capabilities
- Linear dose response
- Dose response independent of dose rate

The primary drawback acting to complicate the use of standard EPIDs for dosimetry is their over-sensitivity to low energy radiation relative to water – the medium to which dosimeters are typically calibrated\textsuperscript{[86,91,99–102,137]}. This comes as a result of the high-Z components making up the active region of the detector including the metal plate (usually copper (Cu)) and phosphor screen (usually Gd\textsubscript{2}O\textsubscript{2}S:Tb). As a consequence of differences between the mass attenuation coefficients of x-rays in water and Gd\textsubscript{2}O\textsubscript{2}S:Tb (see Figure 2.11(a)), low energy x-rays interacting in EPID phosphor screens have a much higher probability of undergoing photoelectric absorption relative to water. EPID response is therefore highly sensitive to factors that change the incident x-ray spectrum, such as the detector’s position away from the beam central axis (beam softening) and the presence of a patient or phantom in the beam (beam hardening)\textsuperscript{[49,87,101,138–142]}. As a result, the EPID response must either be calibrated for dosimetry based on the specific procedure and geometrical configuration being used or a model-based approach must be taken to predict the detector’s response to an incident beam.

Physicists at the Netherlands Cancer Institute have been especially productive in establishing and demonstrating procedures to use current metal plate/phosphor-based EPIDs for accurate 2D dosimetry\textsuperscript{[50,143–147]}. As an example of the dosimetric accuracy that may be achieved using current EPIDs for pretreatment IMRT
verification, greater than 99% of the EPID dose image pixels in a 14 × 14 cm² region encompassing the target volume satisfied $\gamma < 1$ (2%/2 mm criteria) when compared to film, the de facto gold standard for 2D dosimetry in many institutions. The authors did, however, acknowledge that the calibration processes necessary are quite labour intensive. This may explain, at least in part, why regular EPID dosimetry continues to be performed in a relatively small number of centres worldwide.

![Graphs showing mass attenuation coefficients for different materials](image)

Figure 2.11: Comparison of the photoelectric (PE), incoherent scattering (Incoh) and total mass attenuation coefficients for (a) water and Gd₂O₂S and (b) water and BC430 plastic scintillator over the radiotherapy energy range (data obtained from NIST XCOM database).

A recent review by van Elmpt et al. (2008) gives an excellent overview of the diverse applications for EPIDs in clinical dosimetry and the range of techniques being implemented worldwide. Non-transit EPID dosimetry (i.e. dosimetry performed without a patient or phantom in the beam) is typically used as a QA tool for factors influencing beam delivery. EPIDs have been used successfully in this manner for both IMRT QA and VMAT QA. By detecting the therapy beam after having passed through a patient or phantom, EPIDs may also be used for transit dosimetry. Both pre-treatment and in vivo treatment verification may be performed by comparing the delivered dose dis-
tribution (measured with the EPID) to the planned distribution predicted using the TPS\cite{50,51,145,146,151,154,155}. Each of these dosimetry methods may be further categorized according to whether the detected fluence was reconstructed to give calculated dose in 2D\cite{145,151,154,155} or 3D\cite{50,146,151}.

With advanced radiotherapy techniques becoming increasingly common, treatment deliveries have become highly complex and are tailored specifically to individual patients. Stringent QA methods are required now more than ever to ensure the correct treatment delivery for each patient. Despite the lack of a widespread, commercially available product to facilitate clinical EPID dosimetry, some groups have developed in-house methods to use EPID dosimetry as a method for detecting several sources of treatment error\cite{50,51,145,146,149,151,154,155}. Some of these errors may be detected using pre-treatment EPID dosimetry whereas others require dosimetry to be performed \emph{in vivo}. Potential errors may be categorized into those that arise from the machine, the treatment plan or the patient. Table 4 from van Elmpt \emph{et al.} (2008) provides an excellent summary of the various sources of treatment delivery error and appropriate EPID dosimetry methods that may detect them\cite{49}(see Figure 2.12). Presently, however, there is still no general consensus regarding acceptance and rejection criteria for the errors in question.

In 2010, Mans \emph{et al.} published an excellent study that demonstrates the potential for EPID dosimetry\cite{50}. Over a period of three and a half years, their centre performed pre-treatment or \emph{in vivo} EPID dosimetry on the treatment plans of 4,337 patients. Of these plans, 17 serious errors were detected that ultimately led to plan intervention. Nine of these 17 errors would not have been detected with pre-treatment verification alone, thereby demonstrating the important clinical application of routine dosimetry performed throughout treatment delivery.

One of the primary goals for this thesis was to develop a next-generation water-equivalent EPID that may be suitable for simultaneous imaging and dose verification in radiotherapy. Using simulations and measurements to characterize the dose response of standard and novel EPIDs therefore formed a significant component of this work. The vast majority of ongoing EPID dosimetry research focuses on modelling solutions for current generation EPIDs. The investigations reported in this thesis therefore comprise several of only very few studies inves-
2.3. Dosimetry in radiotherapy

Figure 2.12: Summary of treatment delivery errors detectable using EPID dosimetry (figure reproduced from Table 4 of van Elmpt, 2008[49]).

tiating methods to improve this detector for dosimetry, without compromising imaging capability. EPID dosimetry will thus remain a constant theme throughout Chapters 4 – 8.

2.3.3 Plastic scintillation dosimetry

Plastic scintillators are one particular group of materials that have shown significant promise in the field of radiation dosimetry[156]. The reason for this is straightforward – since they are manufactured using low-Z materials, they have repeatedly been shown to respond in a very nearly water-equivalent manner to ra-
diation over the energy range typically encountered in radiotherapy\cite{102, 109, 156–161}. They exhibit additional characteristics including energy independence, dose linearity, resistance to radiation-induced damage and the ability to be manufactured in very small sizes thus enabling high spatial resolution\cite{102, 156, 161, 162}. As a result of these attributes, plastic scintillation dosimeters do not require the same correction factors as other non water-equivalent dosimeters to convert measurements into absolute dose\cite{156}. Much of the work involved in this thesis surrounds the development of a novel radiation dosimeter that employs fibres made of plastic scintillator; therefore their physical properties and related applications are summarized here.

A significant proportion of plastic scintillators are manufactured using either a polystyrene or polyvinyltoluene (PVT) base that is doped with organic fluors to give them their scintillation properties. Some of the earliest reports regarding the physical characteristics of plastic scintillating materials were published in the early 1990s, including a pair of papers by Beddar et al. (1992a, 1992b)\cite{157, 158} Properties including their physical and electron densities were found to be very similar to water. By comparing the mass-energy absorption coefficients of the plastic scintillator to those of water and noting their similarity, Beddar offered an initial demonstration of their water-equivalence (see Figure 2.11(b)). Burlin cavity theory\cite{163} was then used to give additional support to this claim\cite{157}. When compared to air, lithium fluoride, and silicon (the active components of ion chambers, TLDs and diodes, respectively), plastic scintillators were shown to have a dosimetric response much closer to that of water.

Owing to the relative ease of manufacturing plastic scintillators, they may be cut into a variety of custom shapes and sizes (see Figure 2.13). As a result, various prototype dosimeters using plastic scintillation detectors (PSDs) have been reported in the literature\cite{102, 156–159, 164–170}. Several of these prototype detectors have been applied to non-transit and transit dosimetry\cite{168, 170, 171}, and have even been demonstrated for potential in vivo IMRT and VMAT treatment verification\cite{167, 172}. Additional applications to kV dosimetry\cite{173} and brachytherapy\cite{174} have also been demonstrated.

The generation of Čerenkov light within fibre optic cables used in several detectors to transfer signal from the irradiated scintillator to a readout device
proved to be a significant source of noise in these systems. However, many techniques have since been reported to correct for this phenomenon\cite{157,160,175–177}. The generation of Čerenkov light directly within the plastic scintillator is, however, estimated to be $\approx 3$ orders of magnitude lower than that of scintillation light\cite{109}.

Plastics scintillators were studied extensively throughout this thesis while characterizing novel water-equivalent prototype EPIDs developed by our group. Saint-Gobain Crystals manufactured the BC430 plastic scintillator and BCF-99-06A plastic scintillating fibres under investigation in these works. Additional properties relevant to these specific scintillators are discussed in depth in Chapters 6 through 8, which describe our characterization of the imaging and dosimetric capabilities of these novel EPIDs.

2.4 Monte Carlo radiation transport

The Monte Carlo (MC) method is a computational method with origins dating back to the late 1940s and originally published by Metropolis and Ulam in 1949\cite{178}. It is fundamentally based upon repeatedly selecting random numbers to sample known probability distributions in order to solve problems that are difficult or impossible to solve using analytical or deterministic methods. As such, MC methods are especially well suited to problems that are stochastic in
nature. While the focus on radiation transport throughout this thesis provides an excellent example, applications for MC simulations are widespread across several disciplines in science, engineering, statistics and finance\cite{179}. This section attempts to provide sufficient background knowledge in MC simulations as they apply to radiation transport physics by first describing the founding principles of the MC method, then explaining how these principles may be used to model physical phenomena and finally extensions to specific applications in medical radiation physics research.

2.4.1 Foundations and principles

The MC method takes advantage of modern computational capabilities by generating on the order of millions of random numbers each second. In its most basic form, MC simulations rely on two fundamental principles: the generation of random numbers and the sampling of known probability density functions (PDFs) relevant to the problem being solved. These two aspects of the MC method are explored more thoroughly in the following subsections.

2.4.1.1 Random number generators

The ability to perform MC simulations ultimately depends on the ability to generate random numbers. In principle, there are several ways to generate an infinitely long sequence of random numbers. Examples range from the repeated roll of a die to the measurement of radioactive decay, which is a stochastic physical process. The downfall to these approaches is their practicality for use in modern MC techniques that typically require one to sample millions of random numbers within a time frame of a few seconds. While these examples are straightforward and accurate in their ability to generate a sequence of numbers truly random in nature, they are simply too slow to be useful for modern applications.

As a means around this issue, much effort has gone into using modern computers to generate sequences of random numbers that can then be applied to numerical MC applications. While this typically resolves the time constraint for generating large sequences of numbers, an obvious downside is that writing a computer program to generate a sequence of random numbers will result in num-
bers that are, by their very nature, not random. Nevertheless, there are several algorithms in common practice today that are capable of generating sequences of numbers that appear random, with periods (the sequence length before repetition occurs) on the order of $10^{18}$ or greater\textsuperscript{180}. Such algorithms are known as pseudorandom number generators to reflect the fact that they approximate the generation of truly random numbers.

While there are several different algorithms that may be used to create a robust pseudorandom number generator, one class of algorithms most commonly implemented is the linear congruential generator\textsuperscript{179}. This class of generators uses the following recursive algorithm to generate pseudorandom numbers $X_t$ based on integers $a$ and $c$ with a period up to the modulus, $m$:

$$X_t = (aX_{t-1} + c) \mod m \quad t = 1, 2, ...$$

P. L’Ecuyer originally described a multiplicative form of the linear congruential algorithm (taking $c = 0$) known as the RANECU engine in his 1988 paper\textsuperscript{180}. This pseudorandom number generator has a period of $\approx 10^{18}$ and is one of several engines pre-programed into the MC toolkit used throughout this thesis. The RANECU engine was chosen because it has a large period and has been frequently used in the literature. A recent textbook by Kroese, Taimre and Botev (2011) provides an overview of several alternate generators as well as tests that may be performed to evaluate their quality\textsuperscript{179}.

### 2.4.1.2 Probability density functions and sampling

A probability density function (PDF) is a term used to describe a function that gives the relative probabilities for a random variable to assume a specific value. While PDFs typically concern continuous variables, a simple example of a discrete PDF is one that gives the relative probabilities of rolling a given number on a standard die. Having six possible outcomes, each with an equal chance of occurring, the discrete PDF for such a scenario would be a constant with value $1/6$.

When applying the MC method to simulate a physical process, the PDFs un-
2. Literature Survey

der consideration reflect the probabilities for these physical processes to occur. As an example, consider an x-ray with a given energy, momentum and polarization in a known medium. This x-ray may undergo one of several physical processes, although certain processes may be more likely to occur than others. If one knows that PDF describing the relative probabilities for these processes to occur, then a random number sampling that PDF may be used to determine which process the x-ray undergoes. To accurately and realistically simulate such a physical situation, one must know all the relevant PDFs for the processes under consideration.

Most pseudorandom number generators supply numbers with uniform probability – that is, if a particular generator is constrained to returning numbers on the interval (0, 1) then there is no preference within this range to generate any particular subset of numbers. To generate numbers that follow known PDFs, there are several algorithms in place to sample numbers from an inherently uniform pseudorandom number generator. The acceptance-rejection method is a particularly simple algorithm which involves the uniform sampling of a pseudorandom number and accepting it for further use only if it falls within the known PDF\textsuperscript{181}. If it falls outside of the PDF, this number is rejected and another one is sampled, with the process repeating itself until a number is eventually accepted. Using pseudorandom numbers to estimate the value of \( \pi \) provides a simple demonstration of the acceptance-rejection method (see Figure 2.14). Since the ratio of the area of a circle to that of a square with equal diameter is \( \pi/4 \), two pseudorandom numbers \( x \) and \( y \) may be uniformly generated in \((-1, 1)\) and accepted only if \( x^2 + y^2 \leq 1 \). The ratio of accepted \((x, y)\) to the total number \( n \) generated will therefore converge to \( \pi \) for sufficiently large \( n \).

While the acceptance-rejection is simple to implement, it can be inefficient if several numbers must be sampled before one is eventually accepted. The textbook by Lemieux (2009) describes the acceptance-rejection and several alternate methods, including the inversion, composition and convolution methods, that may be better suited to handling different PDFs\textsuperscript{181}.
2.4. Monte Carlo radiation transport

Figure 2.14: Estimating $\pi$ using pseudorandom number generation and the acceptance-rejection method. The ratio of green “accepted” points to the total number of points generated, $n$, converges to $\pi/4$ for increasing $n$.

2.4.1.3 Ongoing and future developments

The MC method is very well suited for solving problems that are stochastic and multidimensional in nature, and that may not be tackled with more conventional means. As an inherent consequence of sampling random numbers to solve a problem, one must ensure that apparent results arising from this method are not attributable to random noise within the simulations. Therefore, depending on the problem being solved and desired level of statistical uncertainty, exceptionally large quantities of random numbers may be generated in a single simulation, which can be very computationally intensive.

Recent and ongoing developments are attempting to mediate this problem by optimizing the way MC simulations are performed on modern computers. One key example uses parallel processing on multi-core and multi-threaded machines including personal laptops, desktop computers and large-scale clusters. With each core of a computer’s central processing unit (CPU) able to independently perform calculations, one can take advantage of the repetitive nature of MC simulations to break a single job into several smaller jobs that each may be executed on a single CPU. Doing so effectively shares the computational burden across several CPUs, thereby reducing the overall time required for a simulation. Results from these statistically independent runs may then be combined after the fact. Depending on the specific conditions of the simulations, the total processing time may be
linearly reduced by a factor that is approximately equal to the number of cores being used.

More recent developments that have been shown to greatly increase the efficiency of performing MC simulations involve the use of graphical processing units (GPUs) in conjunction with conventional CPUs. While modern CPUs contain several cores to process serial computations, by delegating a small but highly computationally intensive fraction of a MC simulation to the GPU (e.g. random number generation), efficiency can be increased by several orders of magnitude\[^{182}\]. Unlike CPUs, GPUs possess hundreds to thousands of cores that enable the simultaneous execution of simple, repetitive tasks. This particular field of research shows incredible promise with regards to future developments of MC simulations\[^{183-185}\].

### 2.4.2 Radiation transport physics

Radiation transport is an inherently stochastic, multidimensional and non-linear process and, in that capacity, is perfectly suited to MC simulations. The physics of radiation transport is governed by the Boltzmann Transport Equation (BTE, equation 2.2) which, when written in terms of the particle flux \( \Phi(\mathbf{r}, \Omega, t) \) (units of \( \text{m}^{-2} \, \text{s}^{-1} \, \text{sr}^{-1} \)) at a position \( \mathbf{r} \), direction \( \Omega \) and time \( t \) in a medium of interest, may be written as\[^{186}\]:

\[
\left( \frac{1}{c} \frac{\partial}{\partial t} + \mathbf{\Omega} \cdot \nabla + l_{\text{ext}}^{-1} \right) \Phi(\mathbf{r}, \Omega, t) = l_{\text{sct}}^{-1} \int_{4\pi} d^2 \Omega' \Phi(\mathbf{r}, \Omega', t) P(\Omega \cdot \Omega') + S(\mathbf{r}, \Omega, t)
\]

(2.2)

\( P(\Omega \cdot \Omega') \) is the probability density for a photon to scatter from an initial direction \( \Omega' \) to a final direction \( \Omega \), \( c \) is the speed of light, and \( l_{\text{ext}} \) and \( l_{\text{sct}} \) are, respectively, the extinction and scattering lengths and are properties of the medium. Equation 2.2 comprises five terms\[^{186}\], the first two of which describe the leakage of photons due to photon propagation out from a point \( \mathbf{r} \) in direction \( \Omega \). The third term describes the loss of photons at \( \mathbf{r} \) and in direction \( \Omega \) due to interactions occurring at that point and the fourth term represents the gain in photons at \( \mathbf{r} \), travelling in direction \( \Omega \) due to scattering at that point from all possible incoming directions.
2.4. Monte Carlo radiation transport

\( \Omega' \) (hence the integral over \( 4\pi \) steradians). Finally, the fifth term describes the gain in photons due to a source \( S(r, \Omega, t) \) at that point.

The BTE may be solved with MC methods by defining a particle source and simulating each particle history by sampling the PDFs associated with that particle’s possible interactions. A particle’s cross section is the term used to describe the probability for a given interaction to occur and is calculated using random sampling of the cross section’s associated PDF. When considering radiation transport, each physical process that a particle may undergo has its own cross section that can depend on several parameters, often including at least the particle’s energy \( E \) and the atomic number \( Z \) of the medium. The total cross section, \( \sigma(E, Z) \), is equal to the sum of the individual cross sections for each interaction process and represents the total probability for any interaction to occur. A summary of the main particle transport processes simulated throughout this work is provided in Table 2.1.

<table>
<thead>
<tr>
<th>Particle</th>
<th>Process</th>
<th>Cross section</th>
</tr>
</thead>
<tbody>
<tr>
<td>X-ray</td>
<td>Photoelectric absorption</td>
<td>( \sigma_{\text{pe}} )</td>
</tr>
<tr>
<td></td>
<td>Compton scattering</td>
<td>( \sigma_{\text{kn}} )</td>
</tr>
<tr>
<td></td>
<td>Pair production</td>
<td>( \sigma_{\text{pp}} )</td>
</tr>
<tr>
<td>Electron</td>
<td>Bremsstrahlung</td>
<td>( \sigma_{\text{Brem}} )</td>
</tr>
<tr>
<td></td>
<td>Mass collisional stopping power</td>
<td>( \frac{dT}{dE} )</td>
</tr>
<tr>
<td></td>
<td>Mass radiative stopping power</td>
<td>( \frac{dT}{dE} )</td>
</tr>
<tr>
<td>Optical photon</td>
<td>Rayleigh scattering</td>
<td>( \sigma_{\text{ray}} )</td>
</tr>
<tr>
<td></td>
<td>Bulk absorption</td>
<td>–</td>
</tr>
<tr>
<td></td>
<td>Boundary processes</td>
<td>–</td>
</tr>
</tbody>
</table>

\( \sigma(E, Z) \) is intimately related to the particle’s mean free path (MFP), \( \lambda(E) \), by the relationship:

\[
\lambda(E) = \frac{1}{\sum_i [n_i \cdot \sigma(E, Z_i)]} \tag{2.3}
\]

where \( n_i \) represents the number of atoms per unit volume of element \( i \) in a compound material. While Equation 2.3 gives the MFP for a known particle, the
actual step length simulated is sampled from a PDF of the form \(1 - e^{-n_\lambda}\), where \(n_\lambda\) is the total number of MFPs travelled by the particle and is a random variable calculated from:

\[
n_\lambda = -\log \eta \tag{2.4}\n\]

Here, \(\eta\) is a uniformly distributed random number on the interval \((0, 1)\). \[^{[187]}\]

The following subsections describe the most relevant x-ray (2.4.2.1 – 2.4.2.3) and charged-particle (2.4.2.4 and 2.4.2.5) interactions in matter over the energy range typically encountered in radiotherapy. Optical photon transport processes are also detailed in Section 2.4.2.6 due to their relevance when modelling a-Si EPIDs. Each gives the necessary theoretical background governing the calculation of each interaction’s cross section and explains the nature of applying these processes within the context of MC radiation transport simulations.

### 2.4.2.1 Photoelectric absorption

Photoelectric absorption was first described by Einstein in 1905 and occurs when a photon, incident upon a material, is absorbed and its energy transferred to a bound electron (see Figure 2.15). If the energy transferred to the electron is greater than its binding energy, the electron will be ejected from the atom with kinetic energy equal to the difference between the incident photon energy and the binding energy.

![Figure 2.15: Schematic illustrating the photoelectric effect. An x-ray incident from the left causes a K-shell electron to be ejected from the atom.](image)
2.4. Monte Carlo radiation transport

The photoelectric absorption cross-section $\sigma_{\text{pe}}$ for a photon with energy $E_\gamma$ and an atom with atomic number $Z$ is given by equation 2.5:

$$\sigma_{\text{pe}} \propto \frac{Z^5}{E_\gamma^{7/2}} \quad \text{(N.R.)}$$

$$\sigma_{\text{pe}} \propto \frac{Z^5}{E_\gamma} \quad \text{(E.R.)} \quad (2.5)$$

for incident photons in the non-relativistic range and away from absorption edges (N.R.) and in the extreme relativistic range (E.R.)\cite{188}.

MC radiation transport codes simulate photoelectric absorption by sampling the MFP that photons travel before undergoing photoelectric absorption, as well as the energy and direction of the ejected electron\cite{187}. The MC software toolkit GEANT4, used extensively throughout this thesis and discussed in depth in section 2.4.3.2, calculates the photon’s MFP using the parameterized photoabsorption cross section proposed by Biggs et al.\cite{187,189}:

$$\sigma_{\text{pe}}(Z, E_\gamma) = \frac{a(Z, E_\gamma)}{E_\gamma} + \frac{b(Z, E_\gamma)}{E_\gamma^2} + \frac{c(Z, E_\gamma)}{E_\gamma^3} + \frac{d(Z, E_\gamma)}{E_\gamma^4} \quad (2.6)$$

As described in the GEANT4 Physics Reference Manual, the coefficients $a$, $b$, $c$ and $d$ were determined using separate fits to experimental data (Grichine et al., 1994) in several energy intervals\cite{187,190}. GEANT4 stores atomic shell binding energies in tables and these are used to compute the ejected electron’s kinetic energy. The ejected electron’s polar angle $\theta$ is sampled from the Sauter-Gavrila distribution\cite{191}:

$$\frac{d\sigma_{\text{pe}}}{d(\cos \theta)} \approx \frac{\sin^2 \theta}{(1 - \beta \cos \theta)^2} \left\{ 1 + \frac{1}{2} \gamma(\gamma - 1)(\gamma - 2)(1 - \beta \cos \theta) \right\} \quad (2.7)$$

where $\beta = v/c$ and $\gamma = (\sqrt{1 - \beta^2})^{-1}$ are the Lorentz factors of the ejected photoelectron with velocity $v$.

2.4.2.2 Compton scattering

Compton scattering occurs when a photon scatters inelastically off an atomic electron, imparting some of its energy to the electron in the process (see Figure
2. Literature Survey

2.16).

![Diagram of Compton scattering](image)

Figure 2.16: Schematic illustrating Compton scattering. An incident photon with energy $E_0$ inelastically scatters off a free electron at an angle $\theta$ and with a reduced energy $E_\gamma$.

It is generally assumed that the electron is at rest and free. This is typically a valid assumption since in situations where binding effects would be significant (such as high $Z$ and low $E_\gamma$), photoelectric absorption events dominate (owing to the $Z^2$ dependence from equation 2.5). Simple kinematics therefore suffice in the determination of the well known Compton formula:\(^{188}\):

$$\epsilon = \frac{E_\gamma}{E_0} = \frac{m_e c^2}{m_e c^2 + E_0(1 - \cos \theta)}$$  \hspace{1cm} (2.8)

where $E_0$ and $E_\gamma$ are the incident and scattered photon energies respectively, $r_e$ is the classical electron radius, $m_e c^2$ is the rest electron mass and $\theta$ is the scattering angle.

The differential cross section per electron is given by the Klein-Nishina formula:\(^{188,192}\):

$$\frac{d\epsilon \sigma_{KN}}{d\Omega} = \frac{r_e^2}{4} \frac{E_\gamma^2}{E_0^2} \left[ \frac{E_0}{E_\gamma} + \frac{E_\gamma}{E_0} - 2 + 4 \cos^2 \theta \right]$$  \hspace{1cm} (2.9)

The total electronic cross section, which gives the probability for a Compton scattering event between the incident x-ray and an electron, may be obtained by integrating equation 2.9 over all $\Omega$.

A modified form of the differential cross section per atom (note the additional factor $Z$ accounting for the number of electrons within the atom) is used by
To sample the total atomic cross section for Compton scattering events, GEANT4 uses a form for \( a_{\text{kn}}(Z, E) \) with empirically-derived coefficients that reproduce experimental data down to 10 keV:

\[
\frac{d a_{\text{kn}}}{d\epsilon} = \pi r^2 m_e c^2 E_0 Z \left[ \frac{1}{\epsilon} + \epsilon \right] \left[ 1 - \frac{\epsilon \sin^2 \theta}{1 + \epsilon^2} \right]
\]

(2.10)

This empirical expression for \( a_{\text{kn}} \) may be used to calculate the probability for a Compton scattering event between the incident x-ray and an atom with atomic number \( Z \). The value of \( \epsilon \) is sampled by applying combined acceptance-rejection and composition MC methods to equation 2.9. After the successful sampling of \( \epsilon \), the polar angles of the scattered photon with respect to the direction of the parent photon are generated. The azimuthal angle, \( \phi \), is generated isotropically in 2\( \pi \) and \( \theta \) is calculated using equation 2.8.

### 2.4.2.3 Pair production

Pair production is the conversion of a photon into an electron/positron pair, occurring most often within a Coulomb field near an atomic nucleus, and hence is the opposite process to electron/positron annihilation (see Figure 2.17). The laws of conservation of energy and momentum result in a lower threshold energy for photons undergoing pair production, which is equal to two times the rest mass energy of an electron (1.022 MeV).

The differential cross section for pair production is given by the Bethe-Heitler formula which, when corrected for the screening effect and Born approximation, takes the form:

\[
a_{\text{kn}}(Z, E_\gamma) = \left[ P_1(Z) \frac{\log(1 + 2X)}{X} + \frac{P_2(Z) + P_3(Z)X + P_4(Z)X^2}{1 + aX + bX^2 + cX^3} \right]
\]

(2.11)
2. Literature Survey

![Schematic illustrating an x-ray with energy $E_\gamma$ undergoing pair production and converting into an $e^-/e^+$ pair.]

Figure 2.17: Schematic illustrating an x-ray with energy $E_\gamma$ undergoing pair production and converting into an $e^-/e^+$ pair.

\[
\frac{d\sigma_{\gamma\gamma}(Z, \epsilon)}{d\epsilon} = \sigma_e^2 Z [Z + \xi(Z)] \left\{ \left[ \epsilon^2 + (1 - \epsilon^2) \right] \left[ \Psi_1(\delta(\epsilon)) - \frac{F(Z)}{2} \right] + \frac{2}{3} \epsilon(1 - \epsilon) \left[ \Psi_2(\delta(\epsilon)) - \frac{F(Z)}{2} \right] \right\}
\]

(2.12)

Here, $\delta(\epsilon)$ is a screening variable and $\Psi_1(\delta)$ and $\Psi_2(\delta)$ are screening functions that are introduced into equation 2.12 to correct for screening of the Coulomb field by outer-shell electrons. A Coulomb correction factor $F(Z) = 8/3 \ln Z$ arises from application of the Born approximation for $E_\gamma < 50$ MeV.

The total cross section per atom for the conversion of an x-ray into an $e^-/e^+$ pair may be parameterized as:

\[
\sigma_{\gamma\gamma}(Z, E_\gamma) = Z(Z + 1) \left[ F_1(X) + F_2(X)Z + \frac{F_3(X)}{Z} \right]
\]

(2.13)

where $E_\gamma$ is the incident x-ray energy and $X = \ln(E_\gamma/m_ee^2)$. The functions $F_n(X)$ for $n = 1, 2, 3$ are a set of 5 degree polynomials in $X$, with coefficients obtained from fits to data provided in Hubbell et al. (1980).\(^{193}\) GEANT4 uses equation 2.13 to calculate the probability for a pair production interaction in materials with $1 \leq Z \leq 100$ and $1.5$ MeV $\leq E_\gamma \leq 100$ GeV. Below the energy $E_{low} = 1.5$ MeV, GEANT4 uses the following extrapolation:\(^{187}\):
2.4. Monte Carlo radiation transport

\[
\sigma_{pp}(E) = \sigma_{pp}(E_{low}) \cdot \left( \frac{E - 2m_ec^2}{E_{low} - 2m_ec^2} \right)^2
\]  

(2.14)

The energies for the \(e^-/e^+\) pair may be sampled by using combined acceptance-rejection and composition methods on equation 2.12. The polar angles \(\theta_-\) and \(\theta_+\) with respect to the direction of the parent photon may be generated for the \(e^-/e^+\) pair by sampling a PDF that approximates the energy-angular distribution given by Tsai (1977)\[187,194,195]\]. The azimuthal angles \(\phi_-\) and \(\phi_+\) are generated isotropically in \(2\pi\).

2.4.2.4 Bremsstrahlung

Bremsstrahlung radiation occurs when a charged particle decelerates while passing near a region of charge. A common example is that of an electron traveling close to an atomic nucleus, such that the nucleus’ electric field alters the electron’s trajectory (see Figure 2.18). In approximately 2-3\% of cases, an inelastic interaction occurs between the charged particle and the nuclear field such that a bremsstrahlung x-ray is emitted\[115\]. These x-rays may obtain up to 100\% of the charged particle’s kinetic energy, causing the charged particle to be drastically slowed down (bremsstrahlung is the German word for “braking radiation”).

Figure 2.18: Schematic illustrating an electron passing near an atomic nucleus and emitting a bremsstrahlung x-ray.

An expression for the differential cross section for bremsstrahlung, which gives
2. Literature Survey

the probability that an x-ray quanta with energy $E$ will be emitted in direction $\theta_0$ relative to the direction of the incident electron and that the electron will be scattered in a direction with polar angles $\theta$ and $\phi$ relative to the x-ray quanta is given by Bethe and Heitler (1934)\cite{188}. By integrating over all angles, the differential cross section describing the probability for generating a bremsstrahlung photon with $E$ may be written in a more concise form\cite{115,188}:

$$\frac{d\sigma_{\text{brems}}}{dE} = 5.80 \times 10^{-28} \frac{B_r Z^2}{E} \left( \frac{T + m_e c^2}{T} \right)$$

(2.15)

where $T$ is the kinetic energy of the incident electron.

Extensive tables of differential cross section data that are typically used by MC codes to sample bremsstrahlung photon energies have been published by Seltzer and Berger (1985, 1986)\cite{187,196,197}. To sample the polar angles governing the post-interaction particle directions, an approximation to the complex cross sections provided by Tsai (1974, 1977)\cite{194,195} may be given by\cite{187}:

$$f(u) = \frac{9a^2}{9+d} \left( u e^{-au} + d u e^{-3au} \right)$$

(2.16)

with $a = 0.625$ and $d = 27$ for the variable $u = E\theta/m$. The polar angle $\theta$ may then be sampled from $f(u)$ using combined acceptance-rejection and composite methods\cite{187}.

2.4.2.5 Collisional and radiative stopping power

Charged particles typically experience a very large number of interactions when passing through matter, though they may only lose a minute portion of their kinetic energy with each interaction. For example, a 1 MeV charged particle may undergo $\approx 10^5$ interactions before losing all of its kinetic energy\cite{115}. For this reason, it is convenient to consider the rate of energy loss by a charged particle with kinetic energy $T$ per unit path length $x$ in a medium with density $\rho$, a quantity known as the mass stopping power, $dT/\rho dx$.

When passing through matter, charged particles may lose energy through collisions and through radiative loses (the most common being bremsstrahlung x-ray emission as described previously in subsection 2.4.2.4). The total mass stopping
power may be calculated as a sum of mass collision and radiative stopping powers:

\[
\frac{dT}{\rho dx} = \left( \frac{dT}{\rho dx} \right)_c + \left( \frac{dT}{\rho dx} \right)_r
\] (2.17)

For electrons and positrons, the mass collision stopping power is given by:

\[
\left( \frac{dT}{\rho dx} \right)_c = k \ln \left( \frac{\tau^2(\tau + 2)}{2(I/m_e c^2)^2} \right) + F^\pm(\tau) - \frac{2C}{Z}
\] (2.18)

where \( k \equiv 0.1535Z/A^2, \tau \equiv T/m_e c^2, \delta \) is a density-effect correction term and \( C/Z \) is a shell correction term. The expressions for \( F^\pm(\tau) \) are polynomials in \( \tau \) with the Lorentz factor \( \beta \) to account for relativistic effects\[115]\.

The mass radiative stopping power for electrons and positrons is given by:

\[
\left( \frac{dT}{\rho dx} \right)_r = \frac{N_A Z^2}{A} (T + m_e c^2) B_r
\] (2.19)

where \( \sigma_0 = \frac{1}{137} (e^2/m_e c^2)^2 \) is a constant, \( N_A \) is Avogadro’s number and \( B_r \) is a slowly varying function in \( Z \) and \( T \)[115].

Values for mass collision and mass radiative stopping powers as function of the charged particle energy may be found in published tables for a wide range of materials (see, for example, the NIST ESTAR database\[198\]). MC codes may then either call upon stored look-up tables for stopping powers relevant to the situation being simulated, otherwise they may be calculated prior to runtime for specified media over an energy range of interest\[187\].

### 2.4.2.6 Optical processes

In the context of this thesis, optical photons are generated via Čerenkov radiation and scintillation events. Once generated, they may undergo three types of processes: elastic (Rayleigh) scattering, bulk absorption and boundary interactions (reflection and refraction).

GEANT4’s treatment of Rayleigh scattering requires users to specify the Rayleigh scattering attenuation length for each material being considered. This scattering length is the mean distance travelled by an optical photon before it undergoes a Rayleigh scattering event. As an elastic scattering process, the only quantity
sampled by GEANT4 is the polar angle $\theta$ of the photon’s new polarization vector with respect to its original. The differential cross section for Rayleigh scattering, $d\sigma_{\text{ray}}/d\Omega$, is proportional to $1 + \cos^2 \theta$.

Simulating bulk absorption requires only that users input the bulk absorption length (the mean distance travelled before a photon is absorbed) for each material being considered. Modelling these events is trivial, as optical photon tracks undergoing bulk absorption are simply terminated.

The manner in which boundary processes are handled naturally depends upon the nature of the two materials making up the boundary. Broadly, there are two such user-specifiable categories in GEANT4:

- dielectric-metal boundary
- dielectric-dielectric boundary

In the simple case of a dielectric-metal boundary, the optical photon cannot be transmitted and the user specifies the relative probabilities for the photon to either be absorbed or reflected back into the dielectric.

For a dielectric-dielectric boundary, the optical photon can either be transmitted or reflected. The probability for the photon to be reflected or refracted and its respective angle of reflection or refraction depend on the indices of refraction for the adjoining media and are governed by Fresnel’s equations$^{[199]}$. More details on the precise calculation of these quantities are given in the GEANT4 Physics Reference Manual$^{[187]}$.

Following GEANT4’s implementation of the UNIFIED model for the treatment of boundaries and surfaces published by Levin and Moisan (1996), users may further customize the nature of boundaries between adjacent materials$^{[200]}$. The simple case of a perfectly smooth boundary is trivial, however there are alternative boundary descriptions that account for the existence of microfacets in the boundary surface (see Figure 2.19). The amount of “roughness” is governed by a user-specified parameter $\sigma_\alpha$ such that an increased $\sigma_\alpha$ correlates to a rougher surface. When an optical photon reaches such a boundary, the true surface normal vector is sampled from a Gaussian distribution with standard deviation $\sigma_\alpha$ and the angle of incidence is calculated with respect to this vector.
2.4. Monte Carlo radiation transport

Figure 2.19: Schematic illustrating the unified model’s treatment of optical boundaries with microfacet surface structure and resulting specular spike (SS), specular lobe (SL), diffuse lobe (DL) and back-scatter spike (BS) optical reflection and transmission distributions (figure reproduced from Levin and Moisan, 1996\textsuperscript{[200]}).

As a consequence of a rough optical surface, different types of optical reflection may occur. These include specular spike, specular lobe, diffuse lobe and backscatter spike reflections, each of which may be assigned a relative probability of occurrence ($C_{ss}$, $C_{sl}$, $C_{dl}$ and $C_{bs}$ respectively, as in Figure 2.19) with their sum equalling unity. For more details, the reader is referred to the original paper by Levin and Moisan\textsuperscript{[200]}.

2.4.3 Specific medical physics applications

Several MC codes are suitable for radiation transport modelling within the context of medical physics. Many are freely available and open-source so that users are made aware of the algorithms used to simulate different physical processes. In these cases, users may also modify existing code to suit custom needs.

Two MC codes that are particularly popular within the field of medical
2. Literature Survey

physics, and which were used throughout this work, are EGSnrc\textsuperscript{[201]} (National Research Council of Canada) and GEANT4\textsuperscript{[202,203]} (CERN). While each has its own advantages and disadvantages, both are frequently used within medical physics research. Within the context of this work, two major uses for these tools were to comprehensively model clinical radiotherapy photon beams and a-Si EPIDs. Details concerning each of these broad tasks are given below, along with a brief summary of other areas application for MC radiation transport codes.

2.4.3.1 Linear accelerator modeling with EGSnrc and BEAMnrc

Radiotherapy source models were developed throughout this work using the MC codes EGSnrc\textsuperscript{[201]} and BEAMnrc\textsuperscript{[204]}. EGSnrc was written specifically for medical physics applications and simulates x-ray, electron and positron transport within the radiotherapy energy range. There also exist a number of freely-available user codes that run on top of the EGSnrc system\textsuperscript{[205]}. BEAMnrc is one example and it provides a user-friendly graphical user interface (GUI) to build models of clinical linear accelerators. BEAMnrc provides a template for each component of a linac and users simply input the relevant geometrical measurements that match the accelerator they wish to model. It is capable of modelling both clinical photon and electron beams and its ability to export the phase-space data of simulation particles was frequently used throughout this work to score particles comprising treatment beams. The output phase-space files may then be used as input into a different simulation to, for example, model the transport of the clinical beam within a phantom or detector. Details concerning the development and validation of source models used in this work are given in Chapter 3.

EGSnrc and BEAMnrc are widely used by medical physics researchers and are frequently reported in the literature. They have become the standard tools to generate and characterize radiotherapy source models, including both photon and electron beams over a range of energy spectra\textsuperscript{[206–212]}. An early publication by Mohan et al. simulated clinical photon beams using EGS Version 3 (a predecessor to EGSnrc). This frequently-cited paper simulated the energy spectra and angular distribution of photon beams produced by a number of Varian linear accelerators\textsuperscript{[206]}. After the development of BEAM (a predecessor to BEAMnrc),
2.4. Monte Carlo radiation transport

Shiekh-Bagheri and Rogers (2002) evaluated the sensitivity of calculated depth-dose curves and off-axis ratios to several incident electron beam and accelerator mechanical parameters\textsuperscript{207}. They further used BEAM to generate photon beam energy spectra for nine MV accelerators from three different manufacturers and then compared these to the results originally published by Mohan \textit{et al.}\textsuperscript{206,208}. A more recent study by Faddegon \textit{et al.} compared the accuracy of EGSnrc against two other MC codes for modelling the electron scatter from 13 and 20 MeV monoenergetic electron beams\textsuperscript{211}. EGSnrc has been used with alternative user codes for detector modelling studies, including characterizing the response of $\alpha$-Si EPIDs\textsuperscript{101,213–218}. The main drawback to using EGSnrc for this application is its inability to simulate the optical wavelength transport relevant for such scintillation detectors. Alternate MC codes such as GEANT4 self-consistently model both x-ray and optical transport, and as such are better suited to these specific applications.

2.4.3.2 Detector modeling with Geant4

GEANT4 is a MC software toolkit that offers a radiation transport physics platform upon which users build custom applications across a range of disciplines\textsuperscript{202,203}. By offering several physics packages, GEANT4 is equally applicable to the very high energy physics studied at the Large Hadron Collider (LHC) and the lower energy physics relevant for medical applications. Within this work, GEANT4 was used primarily to study the physics operating within $\alpha$-Si EPID, however it is often used in other areas of medical physics. Examples include kV and MV radiotherapy\textsuperscript{219–223}, hadrontherapy\textsuperscript{224–228}, gel dosimetry\textsuperscript{229}, brachytherapy\textsuperscript{230}, dose calculations\textsuperscript{231}, emission tomography\textsuperscript{232}, microbeam radiation\textsuperscript{233} and radiobiology\textsuperscript{234} simulations. Each GEANT application must contain at least a user-defined geometry, radiation source, and physics packages. GEANT4 is particularly adept at handling even highly complex geometries, such as the ATLAS detector within the LHC (see Figure 2.20).

This work was predominantly concerned with simulating the response of an $\alpha$-Si EPID for portal imaging and dosimetry applications. A large proportion of this involved validating models of commercial and prototype EPIDs to direct
2. Literature Survey

![Figure 2.20: GEANT4 visualization of the ATLAS detector at the LHC (image courtesy of the GEANT4 collaboration).]

the future development of a next generation detector. Generally, one may use MC simulations to model a detector’s response by scoring quantities such as the particle fluence or energy deposited in a specific region. By comparing simulated quantities to measured data, the model may then be validated. GEANT4 was particularly well-suited to this work because, amongst other freely available MC packages, it has the unique ability to simulate self-consistently both x-ray and optical photon transport, relevant for indirect-detection EPIDs\cite{202,203}. Although the explicit simulation of optical transport has generally been regarded as unimportant insofar as dosimetry simulations using standard EPIDs are concerned\cite{217,218,235}, the self-consistent simulation of x-ray and optical transport had not been modelled within standard EPIDs prior to the studies reported in this thesis. Therefore, part of the motivation for this work (including most specifically that reported in Chapter 4) was to use a novel model that more realistically simulated the physical processes operating within standard EPIDs to independently validate these previous authors’ claims.
2.5. Motivation and Project Aims

Recently, several other groups have used GEANT4 to investigate optical photon transport within phosphor screens\[^{236}\], crystal scintillators\[^{237-239}\] and novel EPID designs\[^{109-111}\]. Pistrui-Maximean \textit{et al.} used GEANT4 to model kV x-ray and optical transport within a Gd\(_2\)O\(_2\)S:Tb phosphor screen and investigated their relative contributions to the imager’s MTF\[^{236}\]. Others have measured\[^{240}\] and simulated\[^{237}\] light collection from BGO crystal scintillators while studying the impact of optical reflectance on the scintillator surfaces. This study took advantage of GEANT4’s implementation of the \textsc{unified} model for optical transport\[^{200}\] and its ability to model transport based on user-specified look-up tables of measured optical reflectance angular distributions. Novel EPID designs have been modelled by Teymurazyan and Pang using GEANT4. One design incorporated an array of plastic scintillating fibres\[^{109}\] while another used an array of standard optical fibers\[^{111}\] and was based on an earlier Čerenkov portal imaging device\[^{112}\].

2.5 Motivation and Project Aims

The overarching goal of this work is to develop a next-generation \(a\)-Si EPID that is capable of simultaneous imaging and dosimetry in modern radiotherapy. Despite the rapid evolution in our ability to deliver highly conformal x-ray beams, limitations with current detectors hinder our ability to measure dose delivery to the patient for QA and treatment plan optimization. The availability of such a detector would be extremely beneficial both to verify that the actual dose delivered matches the planned dose distribution and for identifying those patients that would benefit from treatment adaptations, thus making radiotherapy safer and more effective. While current metal plate/phosphor-based EPIDs offer quantum-limited imaging and may be used for dosimetry, calibration processes are labour intensive. Consequently, their dosimetric abilities have not translated into widespread clinical implementation. A next-generation EPID that improves upon this limitation by, for example, offering water-equivalent dosimetry would simplify \textit{in vivo} dosimetry procedures by reducing the calibration workload and, as a result, may prove to be a more reliable measurement system. One important hypothesis that persists throughout all investigations comprising this thesis is that a next-generation EPID may be developed that:
2. Literature Survey

1. Offers imaging performance equivalent to or better than current EPIDs, and

2. Exhibits a dose response that differs by less than 2% from standard measurements of radiotherapy photon beams using reference dosimeters

The first task towards achieving this goal was to develop and validate a comprehensive model of the standard commercially available a-Si EPID. In doing so, a more specific aim was to characterize the relative importance of different physical processes on the detectors response. This study self-consistently simulates, for the first time, x-ray and optical photon transport within the standard a-Si EPID. The development and validation of the standard EPID model as applied to non-transit dosimetry are discussed in Chapter 4. An extension of the standard EPID model to transit dosimetry is presented in Chapter 5.

The next major task, presented in Chapter 6, was to evaluate the imaging and dosimetry capabilities of a novel EPID designed and purchased by our group. This EPID employed an array of plastic scintillating fibres and was designed for simultaneous imaging and dosimetry capabilities. The study involved the first experimental characterization of this type of detector to be reported in the literature and demonstrated its water-equivalent response.

To aid in the characterization of our physical prototype and the future optimization of its design, a model of this novel EPID was developed. The previously validated standard EPID model was reconfigured into the novel EPID geometry based on an updated, second-generation prototype purchased by our group. The different fibre geometry used in this second-generation prototype was primarily influenced by the measurements performed using the first-generation prototype described in Chapter 6. With limited information available to describe radiation transport within this geometry, it was necessary to first investigate how different model parameters affected the detector response. A characterization of this new model and preliminary optimization of the novel EPID geometry is presented in Chapter 7.

Finally, the novel EPID model was validated against experimental measurements taken with the second-generation prototype. The goal was to quantify and validate both the imaging and dosimetric response of the novel EPID model, and
2.5. Motivation and Project Aims

this work is presented in Chapter 8. The final, validated model of the novel EPID represents a very useful tool that will be used to investigate aspects of the detector design that may be optimized in future generations. The main conclusions derived throughout this work, along with considerations for future studies, are presented in Chapter 9.
References


References


References


References


References


98. Sabet, M., Rowshanfarzad, P., Vial, P., Menk, F. W. & Greer, P. B. Transit dosimetry in IMRT with an a-Si EPID in direct detection configuration.


References


72

118. Khan, F. M. *The Physics of Radiation Therapy* (Lippincott Williams and Wilkins, Philadelphia, 2010), 4 edn. 30


References


References


References


References


References


221. Constantin, M. *et al.* Modeling the TrueBeam linac using a CAD to Geant4 geometry implementation: Dose and IAEA-compliant phase space calculations.


Source modelling and validation

The work presented in this chapter constitutes the development, optimisation and experimental validation of a Monte Carlo model of a clinical 6 MV x-ray photon beam source.
3. Source modelling and validation

3.1 Overview of the BEAMnrc source model

The Monte Carlo radiation transport user code BEAMnrc is one of the most frequently used MC packages within the field of medical physics[1]. It was designed to facilitate the development of clinical linear accelerator (linac) photon and electron beam source models and uses the EGSnrc code for simulating radiation transport within the accelerator geometry[2].

To create a linac source model, users simply build their desired accelerator by combining the necessary BEAMnrc pre-defined component modules. The user then specifies geometric and material parameters, most commonly according to information provided by the linac manufacturer. Rather than relying on manufacturer-provided values, certain parameters may instead be measured directly and these values inputted into the BEAMnrc model for a more precise detector specification[3].

The component modules specified in the linac source models used throughout this work included a tungsten target, primary collimator, flattening filter, monitor ionization chamber, mirror, multi-leaf collimator and tungsten jaws. More details describing the various functions of each component may be found in the original paper describing the BEAM system[1] as well as the user manual available with the current version of BEAMnrc. A schematic illustrating the geometrical configuration of these components in the modelled linac is illustrated below in Figure 3.1.

As described in Section 2.1.2 of Chapter 2, to generate a clinical photon beam of high energy x-rays, electrons are generated from an electron gun and accelerated along a waveguide before colliding with a high atomic number target[4]. BEAMnrc replaces the electron gun and waveguide with a simplified beam of electrons directly incident upon the target with a user-specified energy and geometry. When validating the source model against experimental measurements, it is common practice to adjust sensitive parameters such as the electron energy spectrum and beam geometry, which are difficult to measure directly, when optimising the agreement between simulated and measured data. Almberg et al. described a multivariate source electron optimisation procedure for users desiring a simple and not time-consuming method to achieve percent-level agreement between
3.1. Overview of the BEAMnrc source model

Figure 3.1: Schematic of the Elekta Synergy linear accelerator photon beam model developed using BEAMnrc.

Simulated and measured data\(^5\). A summary of the optimised source electron parameters is given below in Table 3.1.

Table 3.1: Summary of the optimised source electron beam parameters used in the validated linac photon beam model.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Electron energy (MeV)</td>
<td>6.55</td>
</tr>
<tr>
<td>BEAMnrc Source Number</td>
<td>19 (ellipse)</td>
</tr>
<tr>
<td>(\sigma_x)</td>
<td>1 mm</td>
</tr>
<tr>
<td>(\sigma_y)</td>
<td>1 mm</td>
</tr>
<tr>
<td>Mean angular spread (degrees)</td>
<td>1.35</td>
</tr>
</tbody>
</table>

To improve the simulation efficiency of a BEAMnrc linac model, users may specify global electron and photon cut-off energies (variables named ECUT and PCUT respectively). When an electron’s energy falls below ECUT, its history is terminated and its remaining energy is deposited locally (the same occurs
when a photon’s energy drops below $pcut$). Furthermore, several optional variance reduction techniques are offered that users may choose to take advantage of$[1]$. One such technique, known as directional bremsstrahlung splitting (DBS), was used for all linac simulations in this study and involves the “splitting” of bremsstrahlung photons that are aimed into a region of interest (typically a cone described by the linac target position and a DBS radius defined at the isocentre). When a photon is “split”, a significantly greater number of identical photons (determined by the DBS splitting number) are generated instead, each with a reduced weight equal to the inverse of the DBS splitting number. This technique effectively reduces the total number of primary histories that need to be simulated in order to achieve a given level of statistical uncertainty in the simulation. The cut-off energies and DBS parameters used in this work are summarised in Table 3.2. More details concerning this and other variance reduction techniques may be found in the BEAMnrc manual.

Table 3.2: Summary of the radiation transport and variance reduction parameters used in the validated linac photon beam model.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>ECUT</td>
<td>0.7 MeV</td>
</tr>
<tr>
<td>PCUT</td>
<td>0.01 MeV</td>
</tr>
<tr>
<td>DBS splitting number</td>
<td>1,000</td>
</tr>
<tr>
<td>DBS splitting radius</td>
<td>40 cm</td>
</tr>
</tbody>
</table>

Phase-space files of the 6 MV photon beam were generated using the BEAMnrc source model by scoring the charge, energy, position and momenta of all particles passing through a pre-defined plane $z = 89.5$ cm from the linac target. These phase-space files were created for open field sizes ranging from $5 \times 5$ to $40 \times 40$ cm$^2$ and were used as input to a GEANT4 model of a water phantom. To validate the linac source model, central axis percent depth dose (PDD) curves, relative dose profiles, and field size output factors were calculated from simulations of dose deposited in water and compared with experimentally measured values.

Validation of the original BEAMnrc source model, as reported in the appendix of Blake et al. (2013)$[6]$, is discussed thoroughly in Section 3.2. Following that study’s publication, the source model was optimised further according to the
3.2. Non-transit dosimetry source model validation

The following source validation procedure and results were originally reported in the appendix of the publication characterising optical transport within a model of a standard $\alpha$-Si EPID which is included in Chapter 4 of this thesis. In order to validate the MC source model, dose in water measurements were performed including PDD measurements, relative dose in-plane and cross-plane profile measurements at depth of dose maximum ($d_{\text{max}}$), and field size output factor measurements. Static open fields measuring $5 \times 5$, $10 \times 10$, $20 \times 20$ and $40 \times 40$ cm$^2$ were used. PDD and relative dose profile measurements were performed using a linearly scanning CC13 ionization chamber (IBA Dosimetry GmbH, Germany) in a water tank with a source to surface distance (SSD) of 100 cm. Relative dose profiles were also verified using film for the $5 \times 5$ and $10 \times 10$ cm$^2$ fields to account for volume averaging effects of measurements made using the ion chamber. Output factors were measured with the ionization chamber positioned on the beam central axis at a depth of 10 cm in the water phantom and with a SSD of 90 cm. All dose in water measurements were performed with irradiations of 50 MU and a nominal dose rate of 500 MU/min, where 1 MU is defined as a delivered dose of 1 cGy at $d_{\text{max}}$ in water under reference conditions (at the center of a $10 \times 10$ cm$^2$ field with SSD = 100 cm).

3.2.1 Percent depth dose validation

A comparison of simulated and experimentally measured PDD curves in water for a 6 MV photon source is shown in Figure 3.2. The energy deposited within 2 mm of the central axis was scored in 1 mm bins along the central axis to calculate the PDD data. The curves for the $5 \times 5$, $10 \times 10$, $20 \times 20$ and $40 \times 40$ cm$^2$ open field sizes have been scaled by factors of 0.9, 1.0, 1.1, and 1.2, respectively,
3. Source modelling and validation

for improved clarity. Error bars indicating statistical uncertainties for simulation results are smaller than the symbols representing the data points. Between $d_{\text{max}}$ and 300 mm, 99%, 99%, 100% and 100% of the MC simulated data points have $\gamma \leq 1$ (2%/2 mm) for the $5 \times 5$, $10 \times 10$, $20 \times 20$ and $40 \times 40$ cm$^2$ open field sizes, respectively.

3.2.2 Relative dose profile validation

Relative dose profiles in the cross-plane direction at $d_{\text{max}}$ for simulated and experimentally measured data are shown in Figure 3.3. Relative dose profiles were calculated from slices through the central axis at a depth of 15 mm in the water phantom in 1 mm bins. Seventy-four percent, 81%, 91%, and 95% of MC simulated data points had $\gamma \leq 1$ with 2%/2 mm criteria for the $5 \times 5$, $10 \times 10$, $20 \times 20$ and $40 \times 40$ cm$^2$ profiles, respectively. Similar agreement was obtained in the in-plane direction (data not shown). Subsequent to this work, an improved agreement in penumbra was determined by optimising the source electron beam parameters following the procedure described by Almberg et al$^9$. However, since this study’s primary focus was on the effects of optical transport within the EPID model, and given the impractical time burden of repeating the modeling study, the accuracy achieved using our original source electron beam parameters was sufficient for our purposes.

3.2.3 Field size response validation

A comparison of simulated and experimental field size output factors is shown in Figure 3.4 with the values agreeing to within 1 SD (experimental error bars are smaller than the corresponding points on the figure). Field size output factors were calculated from the mean energy deposited in the central $1 \times 1$ cm$^2$ at 10 cm depth in the water phantom. The largest difference between simulated and measured output factors was 1.1% and occurred for the $40 \times 40$ cm$^2$ field size. This discrepancy is most likely due to the systematic error in the source electron beam parameters described previously for the relative dose profiles.
3.2. Non-transit dosimetry source model validation

Figure 3.2: Experimental (Exp) and simulation (Sim) PDD curves for a 6 MV photon beam with $5 \times 5$, $10 \times 10$, $20 \times 20$, and $40 \times 40$ cm$^2$ field sizes in water. For improved clarity, the $5 \times 5$, $10 \times 10$, $20 \times 20$, and $40 \times 40$ cm$^2$ curves have been scaled by factors of 0.9, 1.0, 1.1, and 1.2, respectively. A subplot showing the $\gamma$-index (2%/2 mm) comparing the experimental and simulation curves is also shown.
3. Source modelling and validation

Figure 3.3: Experimental and simulation relative dose profiles for a 6 MV photon source at dmax in water for (a) $5 \times 5$, (b) $10 \times 10$, (c) $20 \times 20$, and (d) $40 \times 40$ cm$^2$ open field sizes. Profiles have been normalised to the central axis dose. Subplots indicating the gamma-index (2%/2 mm) of the experimental and simulation profiles are also shown.
Figure 3.4: Experimental (Exp) and simulation (Sim) field size output factors at depth = 10 cm in water with SSD = 90 cm. Error bars represent a statistical uncertainty of one SD of the MC calculated dose in water within the central 1 × 1 cm² region.

3.3 Source electron parameter optimisation

Following publication of the initial source model described in Section 3.2, the linac model was further optimised according to the method described by Almberg et al. for optimising the source electron parameters in BEAMnrc [5]. Briefly, this procedure follows a three step process whereby users tune individual source electron parameters to determine first the optimal electron energy, second the optimal radial electron intensity and finally the electron beam’s mean angular spread.

The source electron energy was first optimised by comparing simulated central axis PDD curves to experimental measurements for a 5 × 5 cm² open field size. The PDD curves were measured and simulated for a photon beam normally incident on a water phantom, with the relative dose deposited calculated as a function of depth within the phantom. Varying the source electron energy in the linac model caused a slight change in the position of \( d_{\text{max}} \), with higher electron energies resulting in greater \( d_{\text{max}} \). As such, the optimal source electron energy was determined by observing which energy resulted in a depth dose curve that
3. Source modelling and validation

best matched the experimental measurements. Figure 3.5 compares several depth dose curves simulated for monoenergetic electron energies ranging from 6.00 MeV to 6.80 MeV against experimentally measured data for a 5 × 5 cm² field. Given the very slight changes observed for the series of depth dose curves, the local percent difference between the simulated and measured data was also calculated. These differences, along with lines of best fit, are illustrated in Figure 3.6. The electron energy that resulted in a local percent difference plot with minimal slope is that which provided the best agreement with the measured data. Hence, a monoenergetic electron energy of 6.55 MeV was the optimal value for our linac source model and this value was used in all subsequent work.

Figure 3.5: Percent depth dose curves calculated for a series of BEAMnrc simulations with varying source electron energy and experimentally measured data. A subplot showing the γ-index (2%/2 mm) comparing the experimental and simulation curves is also shown.

Once the source electron energy was optimised, the radial intensity of the electron beam incident upon the target was adjusted and relative dose profiles were calculated in directions perpendicular to the beam central axis (parallel and perpendicular to the MLC leaves). As a consequence of the non-zero radial electron beam intensity, dose profiles measured for static, open fields comprise three regions: an in-field region, an out-of-field region and a penumbra region surrounding the field edges. Altering the radial intensity of the source electron beam noticeably modified the shape of the profile penumbra, therefore by matching the
3.3. Source electron parameter optimisation

Figure 3.6: Plots of the local percent differences between each simulated depth dose curve and the experimentally measured data. Lines of best fit are also shown for each dataset.
3. Source modelling and validation

penumbra shape to experimental measurements the optimal radial intensity was
determined. Experimental measurements were performed using gafchromic film
because this dosimeter offers the greatest spatial resolution, critical for accurately
defining the penumbra shape. 5 × 5 and 10 × 10 cm² dose profiles were measured
by irradiating the film with 50 MU at 6 MV in a solid water phantom setup with
9 cm of backscatter, 1.5 cm of buildup and a source to surface distance (SSD) of
100 cm.

BEAMnrc offers users several pre-defined source electron beam shapes ranging
from the simplest point source to very complex geometries. Source 19, which is
defined as an elliptical beam with Gaussian distributions in x and y, parallel or
with radial divergence, was used for all linac source modelling in this work. When
using this source geometry, the Gaussian full width at half-maximum (FWHM,
σx and σy) defined the width of the elliptical source in the x and y dimensions,
respectively.

Figure 3.7 illustrates experimental profiles measured in directions parallel and
perpendicular to the MLC leaves for a 5 × 5 cm² static, open field along with a
series of simulated profiles calculated with varying radial distributions, specified
using values of σ ranging from 0.0 to 0.25 cm. To determine the optimised
values of σx and σy that gave the greatest agreement with the experimental data,
the root mean square (RMS) difference between each simulated profile and the
experimental profile was calculated for the small regions within ±2 mm of the
field edges and plotted as a function of σ (see Figure 3.8). The values for σx
and σy that minimized the RMS difference were therefore determined to be the
optimised values for the source electron radial distribution. Despite not giving the
minimal RMS difference for profiles in the direction perpendicular to the MLC
leaves, values of 0.1 cm were used for both σx and σy in all subsequent work to
maintain radial symmetry of the source electron beam.

The final source electron beam parameter to be optimised using the method
described by Almberg et al. was the mean angular spread of the source electrons
about the beam central axis. For a parallel beam, the mean angular spread
would have assumed a value of zero. However by increasing the value of this
parameter, changes in the shape of the dose horns at distances away from the
central axis were observed in the relative dose profiles for large fields. The mean
3.3. Source electron parameter optimisation

Figure 3.7: Relative dose profiles calculated parallel (above) and perpendicular (below) to the MLC leaves for a range of $\sigma$ compared to experimentally measured profiles. Subplots showing the $\gamma$-index (2%/2 mm) comparing the experimental and simulation profiles are also shown.

Angular spread may therefore be adjusted to improve agreement in large field profiles without affecting the response for small fields. Figure 3.9 illustrates several relative dose profiles calculated for a $40 \times 40$ cm$^2$ field size and a range of mean angular spread values, as well as an experimentally measured profile, in directions both parallel and perpendicular to the MLC leaves. The local percent difference between simulated and measured profiles was used to determine the mean angular spread that minimized the difference between the datasets. These local percent differences are shown in Figure 3.10. A mean angular spread of 1.35$^\circ$ was found to be sufficient and was used throughout the remainder of this
3. Source modelling and validation

work. A summary of the final optimised source electron beam parameters is given in Table 3.1.

Figure 3.8: The RMS difference between each of the above relative dose profiles and the experimental data. RMS differences are plotted as a function of $\sigma$ for profiles calculated in both the directions parallel and perpendicular to the MLC leaves.
3.3. Source electron parameter optimisation

Figure 3.9: Relative dose profiles calculated parallel (above) and perpendicular (below) to the MLC leaves for a range of mean angular spread values compared to experimentally measured large field (40 x 40 cm$^2$) profiles. Subplots showing the $\gamma$-index (2%/2 mm) comparing the experimental and simulation profiles are also shown.
3. Source modelling and validation

Figure 3.10: The local percent difference between simulated relative dose profiles and the experimental data. The differences are shown for profiles calculated in both the directions parallel (‘Par’) and perpendicular (‘Perp’) to the MLC leaves.
References


Characterization of optical transport effects on EPID dosimetry using GEANT4

The work presented in this chapter constitutes the development, characterisation and experimental validation of a Monte Carlo model of a standard amorphous silicon electronic portal imaging device and clinical 6 MV x-ray photon beam source.
4. Simulating optical transport in an $\alpha$-Si EPID

Statement of joint authorship

This work has been published in *Medical Physics* as:


<table>
<thead>
<tr>
<th>Author</th>
<th>Specific involvements</th>
</tr>
</thead>
<tbody>
<tr>
<td>S.J. Blake</td>
<td>Developed, built and ran simulation code, aided experimental design, performed experiments, analysed results, wrote manuscript (80%)</td>
</tr>
<tr>
<td>P. Vial</td>
<td>Experimental design, performed experiments, edited manuscript (5%)</td>
</tr>
<tr>
<td>L. Holloway</td>
<td>Aided experimental design, performed experiments, edited manuscript (5%)</td>
</tr>
<tr>
<td>P.B. Greer</td>
<td>Aided experimental design, edited manuscript (2.5%)</td>
</tr>
<tr>
<td>A.L. McNamara</td>
<td>Performed experiments, edited manuscript (2.5%)</td>
</tr>
<tr>
<td>Z. Kuncic</td>
<td>Intellectual contribution toward Monte Carlo modelling, aided simulations, aided experimental design, edited manuscript (5%)</td>
</tr>
</tbody>
</table>
Abstract

Purpose: Current amorphous silicon electronic portal imaging devices (a-Si EPIDs) that are frequently used in radiotherapy applications employ a metal plate/phosphor screen configuration to optimize x-ray detection efficiency. The phosphor acts to convert x-rays into an optical signal that is detected by an underlying photodiode array. The dosimetric response of EPIDs has been well characterised, in part through the development of computational models. Such models, however, have generally made simplifying assumptions with regards to the transport of optical photons within these detectors. The goal of this work was to develop and experimentally validate a new Monte Carlo (MC) model of an a-Si EPID that simulates both x-ray and optical photon transport in a self-contained manner. Using this model the authors establish a definitive characterisation of the effects of optical transport on the dosimetric response of a-Si EPIDs employing gadolinium oxysulfide phosphor screens.

Methods: The GeANT4 MC toolkit was used to develop a model of an a-Si EPID that employs standard electromagnetic and optical physics classes. The sensitivity of EPID response to uncertainties in optical transport parameters was evaluated by investigating their effects on the EPID point spread function (PSF). An optical blur kernel was also calculated to isolate the component of the PSF resulting purely from optical transport. A 6 MV photon source model was developed and integrated into the MC model to investigate EPID dosimetric response. Field size output factors and relative dose profiles were calculated for a set of open fields by separately scoring energy deposited in the phosphor and optical absorption events in the photodiode. These were then compared to quantify effects resulting from optical photon transport. The EPID model was validated against experimental measurements taken using a research EPID.

Results: Optical photon scatter within the phosphor screen noticeably broadened the PSF. Variations in optical transport parameters reported in the literature caused fluctuations in the PSF full width at half maximum (FWHM) and full width at tenth maximum (FWTM) of less than 3% and 5%, respectively, confirming model robustness. Greater deviations (up to 9.5% and 36% for FWHM and FWTM, respectively) were observed when optical parameters were largely...
different from reference values. When scoring energy deposition in the phosphor, measured and calculated output factors agreed within statistical uncertainties and at least 94% of the MC simulated profile data points passed 3%/3mm $\gamma$-index criterion for all field sizes considered. Despite statistical uncertainties in optical simulations arising from computational limitations, no differences were observed between optical and energy deposition profiles.

**Conclusions:** Simulations demonstrated noticeable blurring of the EPID PSF when scoring optical absorption events in the photodiode relative to energy deposition in the phosphor. However, modelling the standard electromagnetic transport alone should suffice when using MC methods to predict EPID dose response to static, open 6 MV fields with a standard $a$-Si photodiode array. Therefore, using energy deposition in the phosphor as a surrogate for EPID dose response is a valid approach that should not require additional corrections for optical transport effects in current $a$-Si EPIDs employing phosphor screens.
4.1 Introduction

Patient-specific verification of dose delivery is ever more desirable to ensure the correct delivery of complex treatment fields in radiotherapy. By monitoring the dose delivered to the patient throughout the treatment course, necessary adaptations to the treatment plan may be made (for example, in cases of changing patient anatomy). Electronic portal imaging devices (EPIDs), which are based on active matrix flat-panel imager technology[1], have been demonstrated to be suitable for radiotherapy dosimetry applications due in part to their high spatial resolution, real-time data acquisition capabilities, and resilience to radiation-induced damage[2–4]. Furthermore, they have been shown to exhibit a response that is linear with dose and independent of dose rate[2].

Despite their demonstrated dosimetric capabilities, EPIDs are primarily designed for megavoltage imaging and used clinically for verifying patient positioning. Present commercially available EPIDs indirectly detect incident radiation by means of a metal plate and phosphor screen (typically terbium-doped gadolinium oxysulfide, Gd$_2$O$_2$S:Tb) that converts x-rays into optical photons which are detected by an underlying amorphous silicon (a-Si) photodiode array. These indirect-detection EPIDs have been shown to be input-quantum-limited at low doses with a reported detective quantum efficiency of approximately 1%[5]. A high atomic number x-ray converter provides efficient detection of photons suitable for low dose imaging in commercial EPIDs. However, such x-ray converters present problems when using these detectors for dosimetry applications. It is well reported[6–8] that the Gd$_2$O$_2$S:Tb phosphor has greater sensitivity to low energy photons as photoelectric absorption events dominate for energies lower than $\approx 300\text{ keV}$. This results in a nonwater equivalent dosimetric response that complicates EPID calibration to reference dosimeters such as an ion chamber in water. Previous studies have demonstrated the relative importance of EPID layers to dose–response[9].

One method of performing dose verification using EPID images is to compare portal dose images with predicted dose distributions calculated using a predictive EPID model. A number of studies have demonstrated the ability to accurately predict EPID dosimetric images by employing Monte Carlo (MC) simulations of
radiation transport in EPIDs\textsuperscript{[6,10–13]}. The majority of these studies have, however, made simplifying assumptions regarding the significance and/or effects of optical photon transport in these detectors. The blurring effects of optical transport have been estimated to be less than 1 mm in a notional detector modeled by Wittenau \textit{et al.}\textsuperscript{[10]} That study, however, did not explicitly model optical transport and incorporated a relatively thin (35.7 mg cm\textsuperscript{-2} Gd\textsubscript{2}O\textsubscript{2}S:Tb) phosphor screen.

A detailed MC model of a Varian aS500 EPID was developed by Siebers \textit{et al.} to compute static and IMRT portal dose images using a 6 MV photon beam\textsuperscript{[11]}. Calibrated MC-computed dose images agreed very well with measured images, however, an empirically derived layer of backscatter material was required to match computed and measured dose profiles. While optical transport was not modeled in this study, the authors acknowledged that the effects of optical scatter would be similar to the scattering effects resulting from the applied backscatter layer, and that perhaps both backscatter and optical scatter contributed toward signal blurring. Other studies have also reported agreement between MC-computed and measured EPID response. Parent \textit{et al.} developed a MC model of an Elekta iViewGT EPID to investigate changes in response for open and IMRT fields due to spectral variations occurring away from the beam central axis\textsuperscript{[12]}. Generally good agreement between simulated and measured IMRT fields was found, despite response variations of up to 29\% observed at off-axis field positions compared to central axis fields. Wang \textit{et al.} used MC methods to produce a series of monoenergetic EPID dose kernels for varying thicknesses of backscatter material\textsuperscript{[13]}. The effective backscatter thickness required to match predicted and measured field size response was determined for five different Varian aS500/aS1000 imagers. By convolving the dose kernels with the energy-differential particle fluence at the imager plane, EPID field size response for the five imagers was modeled to within 0.34\% of measured values.

While the aforementioned studies scored the energy deposited in the phosphor layer to calculate portal dose images, others have used empirical\textsuperscript{[14]} and MC methods\textsuperscript{[15]} to include the effects of optical transport in their prediction models. Warkentin \textit{et al.} reported dosimetric IMRT verification by deconvolving EPID images with dose deposition and optical glare kernels into primary fluence distributions that were compared with measurements\textsuperscript{[14]}. The optical glare kernel was
described as a double-exponential function with parameters determined by fitting the resulting fluence profiles to measurements. A similar deconvolution approach was reported by Kirkby and Sloboda, however this study used MC methods to calculate dose deposition and optical point spread functions (PSFs)\textsuperscript{[15]}. These PSFs were then combined to yield an overall kernel that was used to deconvolve EPID images into fluence distributions. In that study, optical transport noticeably broadened the overall kernel.

The treatment of optical blurring in modeling EPID dose response is inconsistent and its importance remains unclear. The goal of this work was to develop and experimentally validate a new MC model of an EPID that is self-contained in its treatment of x-ray and optical photon transport. While this method incurs a significant computational burden associated with the individual tracking of large numbers of optical photons, modeling EPID response in this manner offers a more physically realistic scenario of the interactions and transport effects leading to image formation in current \textit{a-Si} EPIDs than previously reported models employing cascaded approaches or neglecting optical transport altogether. With this model we may therefore provide a definitive characterisation of the effects of optical transport on the dosimetric response of \textit{a-Si} EPIDs.

4.2 Methods and Materials

4.2.1 Description of the Monte Carlo model

All MC simulations were performed using a computer cluster employing 252 × 2.67 GHz CPUs. The open source message passing interface OpenMPI\textsuperscript{*} was employed to automate management of batch simulations run in parallel.

4.2.1.1 6 MV photon source

The radiation transport MC code \textsc{egsnrc}\textsuperscript{[16]} (V4 2.3.1) and user code \textsc{beamnrc}\textsuperscript{[17]} (V4 2.3.1) were used to model an Elekta Synergy 6 MV photon source that was used for all EPID simulations. A detailed description of the linear accelerator

\textsuperscript{*}http://www.open-mpi.org/
4. Simulating optical transport in an $a$-Si EPID

components was provided by the manufacturer (Elekta, Crawley, UK) for input into BEAMnrc. Source 19, which defines the spatial distribution of the source electron beam by an ellipse, was used with a FWHM of 0.11 cm about the central axis in the $x$ and $y$ directions and zero angular spread. A Gaussian distribution with a mean energy of 6.0 MeV and FWHM of 0.21 MeV defined the source electron energy spectrum. Global $\text{ecut}$ and $\text{pcut}$ of 0.7 and 0.01 MeV, respectively, were used. Finally, directional bremsstrahlung splitting with a splitting number of 1,000 and a splitting radius of 40 cm was used when modeling all open fields. After passing through all field-defining components of the linear accelerator, particles traversing a predefined plane ($z = 89.5$ cm from the upper surface of the tungsten target) perpendicular to the beam central axis were scored and saved in an output phase-space file. It was found that $5 \times 10^8$ primary histories were sufficient to achieve statistical uncertainties in the EPID model of less than 1%. The history-by-history uncertainty method described by Walters $et$ $al.$ was used to calculate all statistical uncertainties quoted in this study, unless otherwise stated$^{[18]}$. This method accounts for correlations that may exist between particles in a phase-space source originating from the same primary history. Validation of the source model is provided in the Chapter 3.

4.2.1.2 EPID geometry and electromagnetic physics

The GEANT4 MC simulation toolkit$^{[19,20]}$ (version 9.4) was used to develop the $a$-Si EPID model and was chosen for this study because of its demonstrated ability to simulate x-ray and optical photon transport$^{[21,22]}$. ROOT$^{[23]}$ (version 5.28.00) was used for the postprocessing data analysis. The model consists of a series of uniform layers representing individual detector components. Geometrical and material compositions were obtained from specifications supplied by the manufacturer of the EPID used in the validation stage of this study (PerkinElmer, Santa Clara, CA). The EPID model (XRD 1640 AN CS) has a cross-sectional area measuring $41 \times 41$ cm$^2$ centered on, and perpendicularly oriented to the central axis of the incident beam. A schematic of the key layers of the EPID model is given in Figure 4.1. The modeled phosphor screen includes front and back cellulose acetate protective overcoats, a reflective support layer, and a Gd$_2$O$_2$S:Tb
active layer with a reduced density of 5.64 g cm\(^{-3}\) (calculated from manufacturer specifications) to account for the effect of the low density polyurethane binder used to suspend the phosphor grains. The a-Si photodiode layer was modeled as a thin (0.1 mm), uniform layer of a-Si supported by a 1 mm SiO\(_2\) substrate. The pixelated structure and nonunity fill factor of the individual pixels were not explicitly modeled. Materials outside of the 41 \(\times\) 41 cm\(^2\) active region of the research EPID, such as the external electronics, were not included in the model.

![Figure 4.1: Schematic of the key layers of the EPID model (not to scale).](image)

The standard electromagnetic physics and optical physics GEANT4 classes were used to simulate radiation and optical photon transport within the EPID model. Simulated electromagnetic processes included Compton scattering, pair production, photoelectric absorption, impact ionization, bremsstrahlung radiation, electron/positron annihilation, scintillation, Čerenkov radiation, and multiple scattering.

A default range cut value of 1 mm was specified for all GEANT4 simulations. Briefly, GEANT4 tracks all particles down to zero range. However, for computational performance purposes, users may specify a range cut value that is internally converted to an equivalent energy for each of the materials included in the simulation. When a particle’s energy drops below the range cut value for a given material, secondary particles are no longer produced and the remaining energy is instead deposited locally. Range cut values as low as 10\(\mu\)m were investigated however there were no significant variations in quantities calculated from the simulations.
4. Simulating optical transport in an a-Si EPID

4.2.1.3 Optical physics

Optical physics processes that were simulated include bulk absorption, Rayleigh scattering, and boundary processes (reflection and refraction). Optical properties such as refractive indices $n$, bulk absorption lengths $l$, Rayleigh scattering lengths $\mu$ and scintillation parameters (scintillation yield, $SY$; resolution scale $RS$; optical emission spectra $\mu(E)$ with peak wavelength, $\mu_{\text{peak}}$; and time decay constant, $t$) were specified for the relevant detector components and are summarized in Table 4.1. These physical properties were based on specifications provided by the manufacturer (Carestream Health, Inc. Rochester, NY) where possible, or otherwise found in the literature. Although Mie scattering would more accurately model optical transport within granular phosphor screens, Rayleigh scattering was used instead as it requires fewer user specified parameters and is thus more straightforward to implement in GEANT4. A comparison of simulations performed using isotropic, Mie and Rayleigh scattering confirmed that there were no significant differences in calculated quantities between these methods.

Table 4.1: Summary of the reference optical physics properties specified in the EPID model.

<table>
<thead>
<tr>
<th>Layer</th>
<th>Property</th>
<th>Value</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Phosphor</td>
<td>$n_{\text{phos}}$</td>
<td>Refractive index, phosphor</td>
<td>2.4</td>
</tr>
<tr>
<td></td>
<td>$l_{\text{phos}}$</td>
<td>Absorption length, phosphor</td>
<td>4.0 cm</td>
</tr>
<tr>
<td></td>
<td>$\mu_{\text{phos}}$</td>
<td>Scattering length, phosphor</td>
<td>17 $\mu$m</td>
</tr>
<tr>
<td></td>
<td>$SY$</td>
<td>Scintillation Yield (nominal)</td>
<td>60,000/MeV</td>
</tr>
<tr>
<td></td>
<td>$SY$</td>
<td>Scintillation Yield (actual)</td>
<td>1,000/MeV</td>
</tr>
<tr>
<td></td>
<td>$\lambda(E)$</td>
<td>Emission spectrum</td>
<td>380 – 620 nm</td>
</tr>
<tr>
<td></td>
<td>$\lambda_{\text{peak}}$</td>
<td>Peak emission wavelength</td>
<td>545 nm</td>
</tr>
<tr>
<td></td>
<td>$t$</td>
<td>Scintillation time decay constant</td>
<td>1 ms</td>
</tr>
<tr>
<td>Overcoat</td>
<td>$n_{\text{coat}}$</td>
<td>Refractive index, overcoat</td>
<td>1.48</td>
</tr>
<tr>
<td>Photodiode</td>
<td>$n_{\text{diode}}(\lambda)$</td>
<td>Refractive index, photodiode</td>
<td>0.46 – 5.187</td>
</tr>
<tr>
<td></td>
<td>$l_{\text{diode}}(\lambda)$</td>
<td>Absorption length, photodiode</td>
<td>5.29 – 13 300 nm</td>
</tr>
<tr>
<td>Boundary A</td>
<td>$\sigma_{a,A}$</td>
<td>Surface roughness</td>
<td>1.0</td>
</tr>
<tr>
<td>Boundary B</td>
<td>$\sigma_{a,B}$</td>
<td>Surface roughness</td>
<td>1.0</td>
</tr>
</tbody>
</table>

The parameter $SY$ defines the mean number of optical photons generated per unit of energy deposited during a scintillation event. The actual number of optical photons generated is sampled from a Gaussian distribution with mean
SY and standard deviation $RS \cdot \sqrt{SY}$, where $RS$ is an intrinsic property of the scintillating material related to impurities in doped scintillators. Different values of $RS$ would simply affect the width of the distribution from which the number of optical photons created for a given event is sampled. Altering this parameter does not affect the physical transport of the optical photons, hence an arbitrary value of unity was chosen for this study. While the Gd$_2$O$_2$S:Tb scintillator has a nominal $SY$ of 60,000 optical photons per MeV of deposited energy$^{[25]}$, it was found that using a decreased value of $SY$ significantly reduced the simulation computation time without compromising the optical response, above a minimum threshold value. This technique of lowering $SY$ to improve simulation efficiency was first reported by Star-Lack et al.$^{[26]}$ The effects of varying $SY$ within the context of our specific model were investigated and are detailed below in Section 4.2.2.

To allow more control over the treatment of boundaries within the layers of the phosphor screen, a $G4OpticalSurface$ was specified at the interface between the reflective support and phosphor layer (hereafter referred to as boundary A) and at the interface between the phosphor and adjacent overcoat layer (hereafter referred to as boundary B). These $G4OpticalSurfaces$ were characterised with a ground finish to properly account for the existence of microfacets at these boundaries. GEANT4 provides users two options, named GLISUR$^{[30]}$ and UNIFIED$^{[31]}$, to model optical boundary processes. The UNIFIED model was used in this study as it allows the user more control over parameters governing boundary processes. Briefly, this model allows the user to specify a parameter $\sigma_\alpha$ which governs the degree of boundary roughness. A given microfacet will have a normal vector forming an angle $\alpha$ with the average surface normal. For each optical boundary process, the angle $\alpha$ is sampled from a Gaussian distribution with standard deviation $\sigma_\alpha$ which was arbitrarily taken to be 1.0 for both boundaries A and B. The effect of varying $\sigma_\alpha$ for boundaries A and B on optical transport within the phosphor screen was studied and is detailed in Section 4.2.2 below.

The refractive index of the reflective support was not provided by the manufacturer, therefore boundary A was specified with an assumed optical reflectivity of unity. Optical properties for the photodiode are sensitive to the energy of incident photons, therefore these parameters were specified as a function of $\lambda$. The
4. Simulating optical transport in an α-Si EPID

bulk absorption and Rayleigh scattering lengths of the overcoat layer were found not to significantly affect optical transport in the model and as such the same values of 4 cm and 17 μm, respectively, were used as for the phosphor layer.

Figure 4.2 illustrates a simulated event in which electronic energy deposition occurs within the phosphor and subsequent optical photons are absorbed by the photodiode.

Figure 4.2: (a) An x-ray incident from the left undergoes a Compton scattering event in the reflective support layer [magnified (b)]. The Compton electron proceeds to deposit energy within the phosphor layer near boundary A, thereby generating optical photons. Some optical photons are then absorbed by the photodiode, after scattering throughout the phosphor layer. Locations of energy deposition and optical absorption events are indicated by circular markers. Note that for image clarity a reduced scintillation yield was used.
4.2. Methods and Materials

4.2.2 Optical transport analysis

Many of the optical transport parameters specified for the relevant EPID components are difficult to measure directly and may not be precisely known for a specific detector due, for example, to variations occurring during the manufacturing process. A sensitivity analysis was therefore conducted on the optical transport parameters from Table 4.1 to evaluate the robustness of the EPID model to the uncertainties in these values. The chosen parameters along with the tested values are listed in Table 4.2 (note that the parameters RS and t were not included in this analysis because variations in these parameters would not spatially alter the physical transport of optical photons). The analysis was performed by using a point source to emit x-rays along a line from the point source through the center of the EPID and normally incident on its surface. X-ray energies were sampled from a 6 MV spectrum that was created by scoring the energies of all particles in the source phase-space file. The simulation of $1 \times 10^7$ primary x-rays was sufficient to achieve statistical uncertainties less than 0.5% in both the dose deposited at the center of the phosphor plane and the number of absorbed optical photons at the center of the photodiode plane. To achieve this level of statistical uncertainty, approximately 3,000 CPU-hours were required. For all simulations involving this point source, the parameter SY was assigned its nominal value of 60,000/MeV unless otherwise specified.

The EPID PSF was evaluated by calculating three different quantities: the PSF of the detected optical photons at the photodiode ($PSF_{Tot}$), the PSF of the energy deposited in the phosphor ($PSF_{Edep}$), and the displacement of the optical photons relative to their point of origin ($\Delta x_{opt}$). In this model $PSF_{Tot}$ represents the total image signal, $PSF_{Edep}$ represents the commonly used surrogate for image signal (ignoring optical effects), and $\Delta x_{opt}$ represents the optical component of $PSF_{Tot}$, quantifying the spatial distribution of EPID signal resulting purely from the transport of optical photons. The calculation of $\Delta x_{opt}$ was determined from the frequency of optical photon absorption events in the photodiode, scored as a function of displacement from their point of origin. The $\Delta x_{opt}$ determined in this way is analogous to previously reported optical glare or optical blur kernels because it acts to reduce the spatial resolution of the EPID[6,14,15].
4. Simulating optical transport in an $a$-Si EPID

Table 4.2: Analysis of the sensitivity of the EPID model to changes in optical transport parameters.

<table>
<thead>
<tr>
<th>Property</th>
<th>Values tested</th>
</tr>
</thead>
<tbody>
<tr>
<td>$n_{\text{phos}}$</td>
<td>Refractive index, phosphor 2.2, 2.6$^{[15]}$</td>
</tr>
<tr>
<td>$l_{\text{phos}}$</td>
<td>Absorption length, phosphor 0.1 and 10 cm$^{[15]}$</td>
</tr>
<tr>
<td>$\mu_{\text{phos}}$</td>
<td>Scattering length, phosphor 10, 25$^{[24]}$ and 50 $\mu$m</td>
</tr>
<tr>
<td>$\lambda(E), \lambda_{\text{peak}}$</td>
<td>Emission spectrum and peak wavelength Monoenergetic 545 nm</td>
</tr>
<tr>
<td>$n_{\text{coat}}$</td>
<td>Refractive index, overcoat 1.3, 1.6</td>
</tr>
<tr>
<td>Boundary process model</td>
<td>GLISUR</td>
</tr>
<tr>
<td>Boundary finish</td>
<td>Polished</td>
</tr>
<tr>
<td>$\sigma^a_{\alpha}$</td>
<td>Surface roughness 0.1, 10, 100</td>
</tr>
</tbody>
</table>

$^a$Applies for unified model with ground finish only.

When extrapolation is necessary, a superscript $REF$ will be used to indicate PSFs calculated using the reference optical transport parameters from Table 4.1.

All PSFs were calculated in the 2D plane containing the phosphor and photodiode layers. PSFs were calculated both on a 2D Cartesian grid and as a function of radial distance from the source ($PSF_{\text{Tot}}$ and $PSF_{\text{Edep}}$) or from the point of optical photon creation ($\Delta x_{\text{opt}}$). Cartesian scoring was performed using bins measuring $0.4 \times 0.4 \text{mm}^2$ and radial scoring was performed using equally spaced radial bins measuring 0.4 mm to calculate PSFs at the same spatial resolution as current commercial EPIDs. To account for the increasing area of consecutive radial bins at greater distances from the origin, the energy deposition and number of absorption events were normalized to the corresponding bin area.

4.2.3 EPID dose response

4.2.3.1 Simulation dose response

The dose response characteristics of EPIDs investigated in this study were field size output factors and relative beam profiles. These were calculated for various open field sizes ($5 \times 5$, $10 \times 10$, and $20 \times 20 \text{cm}^2$). The energy deposited in the $\text{Gd}_2\text{O}_2\text{S}:\text{Tb}$ phosphor layer and the number of optical photons absorbed in the $a$-Si photodiode (using the reference optical transport parameters) were tallied
4.2. Methods and Materials

independently in 2D histograms to quantify their relative contributions to the output factors and profiles. Each histogram contained $1024 \times 1024$ bins, giving an effective square pixel size of $0.4 \times 0.4 \text{mm}^2$, equal to the pixel pitch of the research EPID. The dependence of energy deposition and absorption events on depth within the layers is beyond the scope of this study, although others have reported on such dependencies\(^{[15]}\).

Field size output factors were calculated from the mean response within the central $\approx 1 \times 1 \text{cm}^2$ region for each 2D histogram, normalized to the response from the $10 \times 10 \text{cm}^2$ open field. Output factors calculated from energy deposition events in the phosphor were compared to those calculated from optical photon absorption events in the photodiode and experimentally measured values. Uncertainties in all output factor calculations are quoted as the standard deviation of the response within the central $\approx 1 \times 1 \text{cm}^2$ region.

Dose profiles were first normalized to a central-axis response of 100%. One-dimensional relative profiles were then obtained by extracting the response along a 1D slice through the center of the 2D histograms in the in-plane and cross-plane directions. Relative dose profiles obtained from histograms scoring energy deposition in the phosphor were compared to those obtained from histograms scoring optical photon absorption in the photodiode and experimentally measured values. Agreement between simulated and measured profiles was evaluated by calculating the percentage of data points with a $\gamma$-index $\leq 1$ based on 3%/3 mm criteria (with dose differences calculated globally relative to the dose at the central-axis and considering only those points above a minimum threshold relative dose of 10%)\(^{[32]}\).

Due to the high level of computational time required to simulate optical transport in an event-by-event manner, the parameter $SY$ was assigned a value of $1,000/\text{MeV}$ in all dose response simulations involving optical transport, unless otherwise specified. Furthermore, it was necessary to limit the range of field sizes investigated in this study as it would have required an impractical duration of time to achieve the desired levels of statistical uncertainty in calculated optical output factors and profiles for large field sizes. Therefore, statistical uncertainties in profiles scoring the number of optical absorption events in the EPID photodiode for open field simulations were only achieved to levels of approximately
4. Simulating optical transport in an α-Si EPID

2.6%, 2.5% and 2.4% for the 5×5, 10×10 and 20×20 cm² fields, respectively. Approximately 200, 400, and 1,500 CPU-hours were required for these respective field sizes. To achieve lower levels of statistical uncertainty (≤ 1%) in profiles scoring energy deposition in the phosphor, simulations were repeated without optical transport. In this case, only approximately 36, 180, and 760 CPU-hours were required for the 5×5, 10×10 and 20×20 cm² fields, respectively.

4.2.3.2 Experimental dose response and model validation

Experimental measurements to validate the model were made using the research α-Si EPID described in Section 4.2.1.2. An Elekta (Elekta, Crawley, UK) Synergy 6 MV linear accelerator with the MLCi multileaf collimator was used for all experimental measurements. The research α-Si EPID used in this study was manufactured by Perkin Elmer (PerkinElmer, Santa Clara, CA) and incorporates a 1 mm copper buildup layer, a 133 mg cm² Gd₂O₂S:Tb phosphor x-ray converter screen (Lanex Fast Back, Carestream Health, Inc. Rochester, NY) and an α-Si photodiode array. The copper acts as buildup for the primary beam and filters low energy scattered photons and electrons. Energy deposited in the phosphor is converted into scintillation optical wavelength photons which are then absorbed by the photodiode. Thin-film transistors on the photodiode array allow one to integrate and read out charge stored in the individual pixels. The photodiode array comprises 1024×1024 pixels with a pixel pitch of 0.4 mm, giving a total active surface area of ≈ 41×41 cm². All images were acquired using irradiations of 50 monitor units (MU) with the photodiode array positioned at isocenter distance. To minimize backscatter from the treatment couch, the research EPID was positioned vertically (i.e., on its side) on the couch and centered on the collimator axis of rotation with the gantry rotated to 90 degrees. No external buildup or backscatter materials were used in this study.

The XIS software package (PerkinElmer, Santa Clara, CA) was interfaced with the research EPID to collect all images. A gain setting of 4 pF with a frame integration time of 133 ms was found to provide the highest signal without saturating any pixels. The EPID was left to warm up for 20 minutes in order to reduce the effect of fluctuating dark currents in the photodiode prior to irradiation.
Dark field frames were obtained by integrating the EPID signal for 30 frames with the beam off, prior to collecting images. This dark field was subtracted from subsequent open field images and was updated prior to each new open field size series of irradiations.

A pixel sensitivity correction based on the method described by Greer\textsuperscript{[33]} was applied to all measured images using in-house code written in MATLAB (version R2011a, The MathWorks, Inc.). Whereas the traditional method of dividing raw images by a flood-field image removes both the nonuniform pixel sensitivities and off-axis response from EPID images, this method removes only the nonuniform pixel sensitivities. The effects of image lag and ghosting have been previously reported for EPIDs\textsuperscript{[34,35]}. While not explicitly investigated in this study, dark fields were frequently updated to minimize such effects on all experimental measurements.

4.3 Results and Discussion

4.3.1 Optical transport analysis

Figure 4.3 shows a comparison between the 2D $PSF_{Edep}^{REF}$ and $PSF_{Tot}^{REF}$. It is clear that $PSF_{Tot}^{REF}$ is noticeably broader than $PSF_{Edep}^{REF}$ with differences attributable to the scattering of detected optical photons within the phosphor screen. The PSF broadening effect shown here is discernible at the 0.4 mm resolution of commercial EPIDs and qualitatively agrees with results reported by Kirkby and Sloboda, despite differences in the EPID model and spatial resolution considered\textsuperscript{[15]}.

Differences in calculated $\Delta x_{opt}$ were observed when performing simulations with variations in the optical transport parameters, as listed in Table 4.2. When specifically investigating the use of values reported from previous studies (notably $n_{phos} = 2.6$; $l_{phos} = 10$ cm; and $\mu_{phos} = 25$ $\mu$m), it was found that $\Delta x_{opt}$ did not vary significantly from $\Delta x_{opt}^{REF}$. However, parameters that differed more significantly from the reference values resulted in noticeable changes to $\Delta x_{opt}$. Figure 4.4(a) illustrates the changes in $\Delta x_{opt}$ for selected values of $l_{phos}$ and $\mu_{phos}$ relative to $\Delta x_{opt}^{REF}$. Overall, variations in $n_{phos}$ did not greatly affect $\Delta x_{opt}$ whereas decreasing $l_{phos}$ to 1 mm narrowed $\Delta x_{opt}$ as optical photons were more readily

123
4. Simulating optical transport in an \( \alpha \)-Si EPID

Figure 4.3: EPID model point spread functions (PSFs) calculated by scoring (a) the energy deposited in the phosphor layer and (b) detected optical photons at the photodiode. Both PSFs were calculated using the reference optical transport parameters listed in Table 4.1.

absorbed in the phosphor. Correspondingly, decreasing \( \mu_{\text{phos}} \) to 10 \( \mu \text{m} \) narrowed \( \Delta x_{\text{opt}} \) as optical photons traveled shorter distances on average between scattering events. Increasing either of \( l_{\text{phos}} \) or \( \mu_{\text{phos}} \) acted to broaden \( \Delta x_{\text{opt}} \), with greater broadening occurring for larger values of these parameters. Figure 4.4(b) illustrates the changes in \( \Delta x_{\text{opt}} \) for the values of SY listed in Table 4.2. Overall, decreasing SY had little effect on \( \Delta x_{\text{opt}} \) although for values of \( SY = 1,000/\text{MeV} \) differences relative to the case with \( SY = 60,000/\text{MeV} \) were distinguishable. It was therefore concluded that reducing \( SY \) to 1,000/\text{MeV} should have little effect on the physical optical transport within the EPID model, while significantly decreasing simulation time. As stated in Table 4.1, SY was taken to be 1,000/\text{MeV} for all open field simulations (described below in Section 4.3.2).

Screenshots taken using the visualization functionality of the MC model were used to discern qualitatively the effects of \( \mu_{\text{phos}} \) on \( \Delta x_{\text{opt}} \). An example of this is shown in Figure 4.5, where \( \mu_{\text{phos}} \) was varied between 1, 17, and 50 \( \mu \text{m} \) for the same simulation event. It is clear that by increasing \( \mu_{\text{phos}} \) the spatial distributions of optical scattering in the phosphor and absorption in the photodiode increased accordingly.
4.3. Results and Discussion

Figure 4.4: Variations in calculated $\Delta x_{\text{opt}}$ for (a) select values of $l_{\text{phos}}$ and $\mu_{\text{phos}}$ relative to $\Delta x_{\text{opt}}^{REF}$ and (b) variations in the phosphor $SY$. Subplots show the same data plotted on a linear ordinate scale for comparison.

The observed differences in $\Delta x_{\text{opt}}$ arising from variations in the optical transport parameters generally translated to proportional differences between the corresponding $PSF_{\text{Tot}}$. This was to be expected as $PSF_{\text{Tot}}$ may be considered to be formed from a convolution of $PSF_{\text{Edep}}$, which is unaffected by the optical transport parameters, and $\Delta x_{\text{opt}}^{REF}$ [15,36]. A series of $PSF_{\text{Tot}}$ calculated for variations in optical transport parameters selected from Table 4.2 are illustrated in Figure 4.6, with $PSF_{\text{Edep}}^{REF}$ shown for comparison. For each optical parameter that was investigated, the full-width at half-maximum (FWHM) and full-width at tenth-maximum (FWTM) of the resulting $PSF_{\text{Tot}}$ were calculated. Selected values are summarized below in Table 4.3 along with their percent differences relative to the FWHM and FWTM calculated from $PSF_{\text{Tot}}^{REF}$. Using the values for $n_{\text{phos}}$, $l_{\text{phos}}$ and $\mu_{\text{phos}}$ that were reported from previous studies resulted in $PSF_{\text{Tot}}$ that differed only slightly from $PSF_{\text{Tot}}^{REF}$, with the greatest percent deviations in FWHM and FWTM (occurring for $\mu_{\text{phos}} = 25 \mu\text{m}$) being less than 3% and 5%, respectively. Percent differences in FWHM and FWTM up to 9.5% and 36%, respectively, were observed for optical parameters that were significantly greater or less than the reference values. Optical parameters that were investigated but are not listed in Table 4.3 resulted in differences in the FWHM and FWTM less
4. Simulating optical transport in an α-Si EPID

Figure 4.5: GEANT4 visualization of the EPID phosphor screen demonstrating the effects of increasing the scattering length in phosphor, $\mu_{\text{phos}}$, from (a) $1 \mu$m to (b) $17 \mu$m and (c) $50 \mu$m while keeping all other parameters constant. The primary x-ray track incident from the left undergoes a Compton scattering event in the support layer. The secondary electron track scatters downward and deposits energy in the phosphor layer. Optical photons are then generated and scatter throughout the phosphor. Energy deposition events in the phosphor and optical absorption events at the photodiode surface are illustrated by circular markers. Note that a reduced scintillation yield was used to improve image clarity.

Figure 4.5: GEANT4 visualization of the EPID phosphor screen demonstrating the effects of increasing the scattering length in phosphor, $\mu_{\text{phos}}$, from (a) $1 \mu$m to (b) $17 \mu$m and (c) $50 \mu$m while keeping all other parameters constant. The primary x-ray track incident from the left undergoes a Compton scattering event in the support layer. The secondary electron track scatters downward and deposits energy in the phosphor layer. Optical photons are then generated and scatter throughout the phosphor. Energy deposition events in the phosphor and optical absorption events at the photodiode surface are illustrated by circular markers. Note that a reduced scintillation yield was used to improve image clarity.

This analysis demonstrates that the dosimetric response of the EPID model remains relatively unaffected over a range of optical parameters reported in the literature for Gd$_2$O$_2$S:Tb and at the spatial resolution of current commercial EPIDs (0.4 mm). Large deviations from $\Delta x_{\text{opt}}^{\text{REF}}$ and $\text{PSF}^{\text{REF}}_{\text{Tot}}$ were found only when the optical transport parameters differed significantly from values reported in the literature, confirming that optical transport would be quantitatively different in alternate materials. A similar analysis would be necessary in order to validate optical transport simulations within alternative configurations employing higher resolution detectors and/or thicker x-ray converters. Under these conditions, PSF broadening effects may be more sensitive to variations in optical
4.3. Results and Discussion

Figure 4.6: (a) $PSF_{Tot}$ for variations in selected optical transport parameters [magnified, with a subplot showing the same data on a linear ordinate scale for comparison (b)].

parameters. Experimental validation of $PSF_{Tot}^{REF}$, such as with the measurement of the EPID line spread function, is beyond the scope of this work and is the subject of ongoing investigations. As an aside, it is worth commenting that although Čerenkov radiation was included in the model for completeness, less than 1% of optical photons originated from Čerenkov processes. The remainder originated from scintillation events in the Gd$_2$O$_2$S:Tb phosphor layer.

4.3.2 EPID model dose-response

4.3.2.1 Field size output factors

Experimental and simulated EPID field size output factors are shown in Figure 4.7. Simulated output factors calculated by scoring energy deposition within the phosphor and optical photons absorbed by the photodiode both agreed with the experimental measurements to within 1 SD. The greatest difference between the simulated and experimental EPID output factors was 2% and was observed for the $5 \times 5$ cm$^2$ field size. This difference may be attributed to slight deviations in the beam phase-space used for the simulations when compared to the experimental
Table 4.3: Summary of the FWHM and FWTM for $PSF_{\text{REF}}^{E_{\text{dep}}}$ and $PSF_{\text{Tot}}$ calculated by varying the corresponding optical transport parameter. Percent differences relative to $PSF_{\text{Tot}}^{\text{REF}}$ are given in parentheses.

<table>
<thead>
<tr>
<th>Property</th>
<th>Value tested</th>
<th>FWHM (mm)</th>
<th>FWTM (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$PSF_{\text{REF}}^{E_{\text{dep}}}$</td>
<td>...</td>
<td>0.807 ($-12%$)</td>
<td>1.16 ($-36%$)</td>
</tr>
<tr>
<td>$PSF_{\text{REF}}$</td>
<td>...</td>
<td>0.922 (0.82%)</td>
<td>1.85 (1.6%)</td>
</tr>
<tr>
<td>$PSF_{\text{Tot}}$</td>
<td>$\gamma_{\text{phos}}$</td>
<td>0.828 ($-9.5%$)</td>
<td>1.17 ($-36%$)</td>
</tr>
<tr>
<td></td>
<td>$l_{\text{phos}}$</td>
<td>0.932 (1.8%)</td>
<td>1.89 (3.6%)</td>
</tr>
<tr>
<td></td>
<td>2.6$[15]$, 10 cm$[15]$</td>
<td>0.922 (0.82%)</td>
<td>1.85 (1.6%)</td>
</tr>
<tr>
<td></td>
<td>$\mu_{\text{phos}}$</td>
<td>0.889 ($-2.9%$)</td>
<td>1.69 ($-7.2%$)</td>
</tr>
<tr>
<td></td>
<td>10 $\mu$m$[24]$</td>
<td>0.939 (2.7%)</td>
<td>1.91 (5.0%)</td>
</tr>
<tr>
<td></td>
<td>25 $\mu$m$[24]$</td>
<td>0.995 (8.6%)</td>
<td>2.29 (23%)</td>
</tr>
<tr>
<td></td>
<td>50 $\mu$m</td>
<td>0.995 (8.6%)</td>
<td>2.29 (23%)</td>
</tr>
<tr>
<td></td>
<td>$\sigma_{\alpha}$</td>
<td>0.924 (1.0%)</td>
<td>1.86 (2.1%)</td>
</tr>
</tbody>
</table>

beam (see Chapter 3 and Figure 3.4). Although the largest discrepancy in output factors calculated when validating the source was found for the $40 \times 40 \text{cm}^2$ field size, EPID output factors were not calculated for this large field due to the computational demand required to achieve the desired statistical uncertainty when modeling optical transport. The EPID model accurately reproduced the detector response to variations in open field size. Simulated output factors calculated by scoring energy deposition within the phosphor and optical photons absorbed by the photodiode agreed within statistical uncertainties in the simulations.

4.3.2.2 Relative dose profiles

Experimentally measured and simulated EPID relative dose profiles are shown in Figure 4.8 in the cross-plane direction. For the profiles calculated by scoring energy deposition in the phosphor, 94.5%, 97.6% and 98.5% of the MC simulated data points had $\gamma \leq 1$ with 3%/3 mm criteria for the $5 \times 5$, $10 \times 10$ and $20 \times 20 \text{cm}^2$ profiles, respectively, when compared to the experimental profiles. It should be noted that the increased percentage of passing gamma values reported here relative to those stated in Chapter 3 for the source validation is a result of the finer scoring grid used in the EPID simulations compared to the dose in water simulations. The larger variations in the optical profiles are attributed to statis-
4.3. Results and Discussion

Figure 4.7: Experimental (Exp) and simulation EPID field size output factors. Output factors calculated from energy deposited in the phosphor (Sim–Edep) and optical photons absorbed by the photodiode (Sim–Opt) are shown. Error bars represent a statistical uncertainty of one standard deviation of the MC calculated EPID response within the central \( \approx 1 \times 1 \text{cm}^2 \) region. Experimental error bars are smaller than the corresponding points on the figure.

Statistical noise arising from the impractical length of time required to achieve lower uncertainties, particularly with the larger fields. Despite the larger statistical uncertainties present in the profiles scoring optical absorption events, 92.4%, 97.6% and 98.2% of the MC simulated data points had \( \gamma \leq 1 \) with 3%/3 mm criteria for the 5 \( \times \) 5, 10 \( \times \) 10 and 20 \( \times \) 20 cm\(^2\) profiles, respectively, when compared to the experimental profiles. These results indicate excellent agreement between simulated and experimental cross-plane profiles. In particular, the increased EPID response off-axis due to the detector’s sensitivity to low energy radiation, clearest for the 20 \( \times \) 20 cm\(^2\) field size, was accurately reproduced with the model. Profiles taken in the in-plane direction agreed equally well with experimental profiles. The small differences observed in penumbra shape between the simulation and experimental profiles (most notable for the 5 \( \times \) 5 cm\(^2\) field size) may be attributed to a systematic uncertainty in the source phase-space that was similarly present in the source validation results (see Chapter 3 and Figure 3.3). This uncertainty, discussed in more detail in Chapter 3, resulted from a small deviation in the
source electron beam parameters relative to more optimal values.

A comparison of the simulation profiles scoring energy deposition in the phosphor with those scoring optical photons absorbed by the photodiode shows that for all field sizes investigated there are no statistically significant differences between the profile shapes. This agrees with the results presented for the EPID PSFs as differences between the optical and energy deposition PSFs were only visible below a relative response of about $10^{-1}$ on a semilog plot. At this level, the PSFs differed spatially by only a few hundred microns. These differences would be difficult to detect with a pixel size of 0.4 mm, particularly when considering the low dose gradient out-of-field EPID response present in EPID profiles at the relative response level of $10^{-1}$.

The close agreement between the energy deposition and optical absorption profiles within the penumbral region is consistent with the results of the optical analysis presented in Section 4.3.1. Close agreement across all field sizes suggests that detailed modeling of optical transport does not result in any significant dosimetric effects for commercial $a$-Si EPIDs with a pixel pitch of 0.4 mm. It is worth commenting that this EPID MC model was not developed for widespread practical applications, but rather as an in-house research tool that can be used to gain deeper insight into the relevant physical processes. This model is only practical for those with dedicated computational resources. However, the results of this study give confidence that for clinical dosimetry purposes involving static open fields, modeling the standard electromagnetic transport alone should suffice for the prediction of EPID dose response using MC methods, and this approach does not impose nearly as great a computational burden. The extension of this finding to modulated delivery techniques such as IMRT and VMAT has not been tested.
4.3. Results and Discussion

Figure 4.8: Experimental (Exp) and simulation EPID relative dose profiles for (a) 5×5, (b) 10×10 and (c) 20×20 cm² field sizes. Simulated profiles calculated from energy deposited in the phosphor (SimEdep) and optical photons absorbed by the photodiode (SimOpt) are shown. Subplots show the γ-index (3%/3 mm criteria) for the experimental and SimEdep profiles.
4. Simulating optical transport in an $a$-Si EPID

Optical transport effects are expected to be more significant when using the EPID for imaging studies (such as for modeling DQE) and for EPIDs employing novel designs (such as different scintillator materials and thicknesses). Work is currently underway to use this comprehensive MC model to assess the importance of optical transport on EPID imaging performance, where the optical blurring effects demonstrated with simulations of the EPID PSF can impact on the detector MTF and DQE. The assessment and optimization of thick segmented scintillators, where the optical effects may be more complex and may not be neglected, are another important future application for this model.

4.4 Conclusion

A comprehensive MC model of an $a$-Si EPID has been developed using GEANT4. The model presented here is the first to simulate both x-ray and optical photon transport in a self-contained manner within the EPID layers. Simulations demonstrated minor but noticeable blurring of the EPID PSF when scoring optical absorption events relative to energy deposition in the phosphor. The optical PSF was insensitive to uncertainties in optical transport parameters reported in the literature. EPID field size output factors and relative dose profiles calculated by scoring energy deposition in the phosphor layer of the MC model agreed with experimental measurements within statistical uncertainties. Furthermore, despite the relatively larger statistical uncertainties in optical simulations, optical output factors and relative dose profiles also agreed with experimental measurements. Specifically, no differences were observed in optical profile penumbrae when compared to energy deposition profiles. Therefore, using energy deposition in the phosphor as a surrogate for EPID dose response to static, open 6 MV photon beams is a valid approach that does not require additional corrections for optical transport effects in current $a$-Si EPIDs employing phosphor screens.

Acknowledgments

The authors would like to acknowledge funding support from the Cancer Institute NSW (Research Equipment Grant Nos. 10/REG/1-20 and 10/REG/1-10), Can-
4.4. Conclusion

cer Council NSW (Grant No. ID RG 11-06), and the National Health and Medical Research Council of Australia (Project Grant No. ID569211). The data for the input files used to model the Elekta Synergy x-ray source were provided under a nondisclosure agreement between Elekta Oncology, the University of Leeds/St. James’s Institute of Oncology, and the University of Sydney. The authors would therefore like to acknowledge Elekta Oncology for providing this information. Professor David Thwaites from the University of Sydney and Dr. Steven Weston, David Paynter, and Dan Johnson from the University of Leeds and St. James’s Institute of Oncology are gratefully acknowledged for their aid in generating the x-ray source input files. The authors would also like to thank Shrikant Deshpande and Jared Begg from the Liverpool and Macarthur Cancer Therapy Centres for their help collecting EPID images and film profiles, respectively. The authors also thank Dr. Josh Star-Lack for providing slides\(^{26}\) and Dr. Rebecca Fahrig for advice on more efficient modeling of optical transport. Finally, the authors would like to thank the reviewers of this manuscript for their insight and suggestions which helped to improve the scientific quality of this paper. S.B. would also like to thank The University of Sydney and the Institute of Medical Physics for scholarship support, as well as the Liverpool and Macarthur Cancer Therapy Centres for additional financial support. The authors report no conflicts of interest in conducting the research.
References


References


References


Monte Carlo simulation of the transit dosimetric response of an $a$-Si EPID

The work presented in this chapter constitutes the extension of the Monte Carlo model described in Chapter 4 to transit dosimetry and the simulation of portal images of an anthropomorphic phantom.
Statement of joint authorship

This work has been published in *Journal of Physics: Conference Series* as:


<table>
<thead>
<tr>
<th>Author</th>
<th>Specific involvements</th>
</tr>
</thead>
<tbody>
<tr>
<td>S.J. Blake</td>
<td>Developed, built and ran simulation code, aided experimental design, performed experiments, analysed results, wrote manuscript (80%)</td>
</tr>
<tr>
<td>A.L. McNamara</td>
<td>Aided simulations, aided experimental design, performed experiments, edited manuscript (5%)</td>
</tr>
<tr>
<td>P. Vial</td>
<td>Experimental design, provided CT image set, performed experiments, edited manuscript (5%)</td>
</tr>
<tr>
<td>L. Holloway</td>
<td>Aided experimental design, performed experiments, provided CT image set, edited manuscript (3%)</td>
</tr>
<tr>
<td>P.B. Greer</td>
<td>Aided experimental design, edited manuscript (2%)</td>
</tr>
<tr>
<td>Z. Kuncic</td>
<td>Intellectual contribution toward Monte Carlo modelling, aided simulations, aided experimental design, edited manuscript (5%)</td>
</tr>
</tbody>
</table>
Abstract

Amorphous silicon (a-Si) electronic portal imaging devices (EPIDs) are x-ray detectors frequently used in radiotherapy imaging and dosimetry applications. EPIDs employ a copper plate and gadolinium oxysulfide phosphor screen with an array of a-Si photodiodes to indirectly detect incident radiation. In this study, a previously developed Monte Carlo (MC) model of an a-Si EPID has been extended for transit dosimetry. The GEANT4 MC toolkit was used to integrate an a-Si EPID model with two phantoms and a 6 MV x-ray source. A solid water phantom was used to simulate EPID transmission factors, field size output factors and relative dose profiles and results were compared to experimental measurements. An anthropomorphic head phantom was used to qualitatively compare simulated and measured portal images of humanoid anatomy. Calculated transmission factors and field size output factors agreed to within 2.0% and 1.9% of experimental measurements, respectively. A comparison of calculated and measured relative dose profiles yielded > 98% of points passing a gamma analysis with 3%/3 mm criterion for all field sizes. The simulated anthropomorphic head phantom image shows macroscopic anatomical features and qualitatively agrees with the measured image. Results validate the suitability of the MC model for predicting EPID response in transit dosimetry.
5. Transit dosimetry simulation of an $a$-Si EPID

5.1 Introduction

Amorphous silicon ($a$-Si) electronic portal imaging devices (EPIDs) serve a number of important clinical applications in modern radiotherapy. EPIDs are routinely used to image patient anatomy and verify patient setup prior to treatment. EPIDs are also suitable dosimeters since the pixel values of acquired images correlate to the absorbed dose in the detector. One method of performing dose verification using EPIDs is therefore by comparing portal dose images to dose distributions predicted using an EPID model. EPID dosimetric characteristics and their various clinical uses for dosimetry have been reviewed by van Elmpt et al.\[^1\]

A number of arguments support the integration of EPID dosimetry into routine clinical practice. Linear accelerator (linac) vendors typically supply $a$-Si EPIDs with the necessary hardware mounted directly to the gantry, in line with the megavoltage (MV) treatment x-ray source. This configuration provides a readily available mechanism to detect the MV beam and enables direct monitoring of both patient position and dose delivery from the beam’s-eye view. When compared to alternative 2D dosimeters such as arrays of diodes or ion chambers, $a$-Si EPIDs offer increased spatial resolution (typically $0.4 \times 0.4 \text{mm}^2$) and real-time data acquisition capabilities. Additionally, EPIDs are resilient to radiation-induced damage and respond both linearly with dose and independently of dose rate\[^2,3\]. One centre has reported on the routine use of EPIDs for pre-treatment and \textit{in vivo} dosimetry, including the EPID’s ability to detect errors in treatment delivery\[^4\].

The goal of this study is to extend the functionality of an EPID model that we previously developed for non-transit dosimetry\[^5\] by integrating phantom geometries into the model. In doing so, we may investigate the EPID response in a transit dosimetry configuration that is more representative of clinical treatment situations. Furthermore, we aim to validate the transit dosimetric response of this model against experimental measurements.
5.2 Monte Carlo model and detector geometry

5.2.1 6 MV photon source
The MC radiation transport code EGSnrc\textsuperscript{[6]} (V4 2.3.1) with user code BEAMnrc\textsuperscript{[7]} (V4 2.3.1) were used to create a 6 MV photon source model of an Elekta Synergy linac (Elekta, Crawley, UK). A description and validation of the source model has been previously reported\textsuperscript{[5]}. The simulation of 10\textsuperscript{9} primary histories was performed to generate phase space files for square fields ranging in size from 2 \times 2 to 9 \times 9 cm\textsuperscript{2} (defined at the isocentre, 100 cm from the target).

5.2.2 EPID geometry and physics processes
The GEANT4 MC simulation toolkit\textsuperscript{[8]} (version 9.4) was previously used to develop a model of an a-Si EPID and validate its dosimetric response in a non-transit configuration\textsuperscript{[5]}. A complete description of the EPID geometry may be found in the previous study as only a brief overview is given here.

The EPID model (Figure 5.1) consists of a series of uniform slab layers with geometries and material compositions based on specifications provided by the manufacturer of a research EPID (XRD 1640 AN CS) used in the validation stage of this study (PerkinElmer, Santa Clara, CA). The EPID model has a cross-sectional area of 41 \times 41 cm\textsuperscript{2} and was positioned at a source to detector distance (SDD) of 160 cm for all simulations. It incorporates a 1 mm Cu buildup layer, a 133 mg cm\textsuperscript{-2} Gd\textsubscript{2}O\textsubscript{2}S:Tb phosphor screen (Lanex Fast Back, Carestream Health, Inc. Rochester, USA) and a 0.1 mm thick layer of a-Si supported by a 1 mm SiO\textsubscript{2} substrate.

The standard GEANT4 electromagnetic physics models were used to simulate radiation transport within the MC model. The transport of optical photons originating in the phosphor screen was not explicitly simulated. We previously found that optical transport does not significantly change calculated dosimetric quantities relative to those calculated using only standard electromagnetic physics\textsuperscript{[5]}. Simulated processes included Compton scattering, pair production, photoelectric absorption, impact ionization, bremsstrahlung radiation, electron/positron annihilation and multiple scattering.
5. Transit dosimetry simulation of an \(a\)-Si EPID

5.2.3 Phantom definitions

This study incorporated two distinct phantom geometries into the MC model to investigate separate EPID dosimetric characteristics. The first phantom was a simple homogeneous box of solid water with a cross-sectional area of \(40 \times 40 \text{ cm}^2\) and a thickness along the central axis that varied depending on the quantity being simulated. The second phantom was an anthropomorphic head that was defined by integrating a set of CT images into the MC model using functions distributed with the GEANT4 source code. Both phantoms were centred about the isocentre and the head phantom was oriented with its anterior-posterior (AP) axis aligned with the beam central axis.

![Schematic of the key layers of the standard EPID model](image)

Figure 5.1: Schematic of the key layers of the EPID model (not to scale).

5.3 Simulated dosimetric quantities

The EPID dose response characteristics investigated in this study include transmission factors, field size output factors and relative beam profiles. An image of
an anthropomorphic head phantom was also simulated for qualitative evaluation. All quantities were calculated by tracking particles from the source phase space files and scoring the energy deposited in the phosphor layer of the EPID in a 2D histogram. Each histogram contained $1024 \times 1024$ bins ($0.4 \times 0.4$ mm$^2$ pixels), equal in number and size to the pixels of the research EPID. All MC simulations were performed using a computer cluster of $252 \times 2.67$ GHz CPUs and the open source message passing interface OpenMPI* was used to facilitate parallel processing. ROOT\textsuperscript{[9]} (version 5.28.00) was used for all post-processing analysis.

### 5.3.1 Transmission factors

Transmission factors were calculated by varying the solid water phantom thickness from 0 to 40 cm in 10 cm increments with a fixed beam field size of $9 \times 9$ cm$^2$. The mean response within the central $1 \times 1$ cm$^2$ region of each 2D histogram was calculated, normalized to the response for the phantom thickness of 0 cm. Uncertainties in all output factor calculations are quoted as the standard deviation of the response within the central region.

### 5.3.2 Field size output factors and relative dose profiles

Field size output factors and relative dose profiles were calculated by varying the beam field size from $2 \times 2$ to $9 \times 9$ cm$^2$ with a fixed solid water phantom thickness of 20 cm. Output factors were calculated as the mean response within the central $1 \times 1$ cm$^2$ region of each 2D histogram, normalized to the $9 \times 9$ cm$^2$ field response. Uncertainties in all output factor calculations are quoted as the standard deviation of the response within the central region.

Dose profiles were first normalized to a central axis response of 100%. 1D relative profiles were then obtained by extracting the response along a slice through the centre of the 2D histograms in the cross-plane direction. Agreement between simulated and measured profiles was evaluated by calculating the percentage of data points with a $\gamma$-index $\leq 1$ based on 3%/3 mm criteria (with dose differences calculated globally relative to the dose at the central-axis and considering only those points above a minimum threshold relative dose of 10%)\textsuperscript{[10]}.

\*http://www.open-mpi.org
5. Transit dosimetry simulation of an a-Si EPID

5.3.3 Projection phantom portal dose image

A static 9 × 9 cm² beam field size was used to generate an AP projection portal image of an anthropomorphic head phantom using the EPID model.

5.4 Experimental measurements and model validation

Experimental measurements to validate the MC model were made using the research a-Si EPID described in Section 5.2.2. An Elekta (Elekta, Crawley, UK) Synergy 6 MV linac with the MLCi multi-leaf collimator was used for all measurements. Images were acquired by averaging 50 frames when delivering a nominal dose rate of 500 MU/min. To minimize backscatter from the treatment couch, the EPID was positioned vertically (i.e. on its side) on the couch and centered on the collimator axis of rotation at a SSD of 160 cm, with the gantry rotated to 90 degrees. Phantoms (as described in Section 5.2.3) positioned on the couch were centred at the isocentre.

The XIS software package (PerkinElmer, Santa Clara, CA) was interfaced with the research EPID to acquire all images. A gain setting of 4 pF was used with a frame integration time of 499 ms. Images acquired for validation of the transmission factors and field size output factors were both dark-field and flood-field corrected. Flood-field corrections were not applied for validation of EPID relative dose profiles or the anthropomorphic head phantom image as this correction would remove the well-known off-axis detector response[11].

5.5 Results and Discussion

5.5.1 Transmission factors

Transmission factors calculated using the MC model and measured using the research EPID are shown in Figure 5.2. Calculated and measured transmission factors are in excellent agreement, with a maximum percent difference of only 2.0% (occurring for the 40 cm phantom thickness).
5.5. Results and Discussion

5.5.2 Field size response

Figure 5.3 shows the calculated and measured variation in EPID response with beam field size when a 20 cm thick solid water phantom is used. The calculated and measured field size responses are in close agreement with the greatest difference of 1.8% occurring for the $3 \times 3 \text{cm}^2$ field size.
5. Transit dosimetry simulation of an α-Si EPID

Figure 5.3: Measured (Exp) and calculated (Sim) EPID field size output factors.

5.5.3 Relative dose profiles

Relative dose profiles calculated using the MC model and measured using the research EPID are presented in Figure 5.4 for selected beam field sizes between $2 \times 2$ and $9 \times 9$ cm$^2$ when a 20 cm thick solid water phantom is used. The subplot shows the results of a $\gamma$ comparison between the calculated and measured profiles for each field size using 3%/3 mm criterion. 98% and 99% of profile data points had $\gamma \leq 1$ for the $2 \times 2$ and $3 \times 3$ cm$^2$ field sizes respectively, whereas 100% of points had $\gamma \leq 1$ for the remaining field sizes. These results demonstrate excellent agreement between the calculated and measured EPID off-axis response.
5.5. Results and Discussion

Figure 5.4: Calculated relative dose profiles from the standard EPID model using a 20 cm thick phantom of solid water (top panel) and corresponding γ-values for 3%/3 mm agreement with measured profiles (bottom panel).

5.5.4 Projection phantom portal dose image

Measured and calculated portal images of an anthropomorphic head phantom are presented in Figure 5.5. A qualitative comparison of these images demonstrates that the MC model is able to simulate spatial variations in detector response resulting from the use of an inhomogeneous phantom representative of human anatomy. The statistical noise present in the calculated image made it difficult to resolve the fine anatomical structures and slight changes in relative density within the phantom. However, macroscopic features such as the orbits and nasal cavity are discernable in the calculated image.
5. Transit dosimetry simulation of an $a$-Si EPID

5.6 Conclusion

A Monte Carlo model of a standard $a$-Si EPID that was previously developed for non-transit dosimetry has been extended to transit dosimetry applications. Transmission factors, field size output factors and relative dose profiles were calculated using the model and validated against experimental measurements with excellent agreement. The simulation of an anthropomorphic head phantom portal dose image provides a demonstration for applying this model to predicting EPID images of humanoid anatomy.

Acknowledgments

Financial support was provided by the Cancer Institute NSW (Research Equipment Grants 10/REG/1-20 and 10/REG/1-10), Cancer Council NSW (Grant ID RD 11-06) and the National Health and Medical Research Council of Australia (Project Grant ID569211). SJB acknowledges The University of Sydney and the Institute of Medical Physics for scholarship support, as well as the Liverpool and Macarthur Cancer Therapy Centres for additional financial support.
References


References


Characterization of a novel EPID designed for simultaneous imaging and dose verification in radiotherapy

The work presented in this chapter includes the first experimental demonstration of a novel, first-generation prototype electronic portal imaging device employing an array of plastic scintillating fibres.
6. A novel EPID for imaging & dosimetry

Statement of joint authorship

This work has been published in Medical Physics as:


<table>
<thead>
<tr>
<th>Author</th>
<th>Specific involvements</th>
</tr>
</thead>
<tbody>
<tr>
<td>S.J. Blake</td>
<td>Aided experimental design, performed experiments, analysed results, wrote manuscript (70%)</td>
</tr>
<tr>
<td>A.L. McNamara</td>
<td>Aided experimental design, performed experiments, edited manuscript (7%)</td>
</tr>
<tr>
<td>S. Deshpande</td>
<td>Performed experiments, edited manuscript (3%)</td>
</tr>
<tr>
<td>L. Holloway</td>
<td>Aided experimental design, performed experiments, edited manuscript (3%)</td>
</tr>
<tr>
<td>P.B. Greer</td>
<td>Aided experimental design, edited manuscript (3%)</td>
</tr>
<tr>
<td>Z. Kuncic</td>
<td>Aided experimental design, edited manuscript (4%)</td>
</tr>
<tr>
<td>P. Vial</td>
<td>Experimental design, performed experiments, edited manuscript (10%)</td>
</tr>
</tbody>
</table>
Abstract

Purpose: Standard amorphous silicon electronic portal imaging devices (a-Si EPIDs) are x-ray imagers used frequently in radiotherapy that indirectly detect incident x-rays using a metal plate and phosphor screen. These detectors may also be used as two-dimensional dosimeters; however, they have a well-characterized nonwater-equivalent dosimetric response. Plastic scintillating (PS) fibers, on the other hand, have been shown to respond in a water-equivalent manner to x-rays in the energy range typically encountered during radiotherapy. In this study, the authors report on the first experimental measurements taken with a novel prototype PS a-Si EPID developed for the purpose of performing simultaneous imaging and dosimetry in radiotherapy. This prototype employs an array of PS fibers in place of the standard metal plate and phosphor screen. The imaging performance and dosimetric response of the prototype EPID were evaluated experimentally and compared to that of the standard EPID.

Methods: Clinical 6 MV photon beams were used to first measure the detector sensitivity, linearity of dose response, and pixel noise characteristics of the prototype and standard EPIDs. Second, the dosimetric response of each EPID was evaluated relative to a reference water-equivalent dosimeter by measuring the off-axis and field size response in a nontransit configuration, along with the off-axis, field size, and transmission response in a transit configuration using solid water blocks. Finally, the imaging performance of the prototype and standard EPIDs was evaluated quantitatively by using an image quality phantom to measure the contrast to noise ratio (CNR) and spatial resolution of images acquired with each detector, and qualitatively by using an anthropomorphic phantom to acquire images representative of human anatomy.

Results: The prototype EPID’s sensitivity was 0.37 times that of the standard EPID. Both EPIDs exhibited responses that were linear with delivered dose over a range of 1–100 monitor units. Over this range, the prototype and standard EPID central axis responses agreed to within 1.6%. Images taken with the prototype EPID were noisier than those taken with the standard EPID, with fractional uncertainties of 0.2% and 0.05% within the central 1 cm², respectively. For all dosimetry measurements, the prototype EPID exhibited a near water-equivalent response.
response whereas the standard EPID did not. The CNR and spatial resolution of images taken with the standard EPID were greater than those taken with the prototype EPID.

**Conclusions:** A prototype EPID employing an array of PS fibers has been developed and the first experimental measurements are reported. The prototype EPID demonstrated a much more water-equivalent dose response than the standard EPID. While the imaging performance of the standard EPID was superior to that of the prototype, the prototype EPID has many design characteristics that may be optimized to improve imaging performance. This investigation demonstrates the feasibility of a new detector design for simultaneous imaging and dosimetry treatment verification in radiotherapy.
6.1 Introduction

Amorphous silicon (a-Si) electronic portal imaging devices (EPIDs), based on active-matrix flat panel imager (AMFPI) technology\(^\text{[1]}\), serve a number of important clinical applications in modern radiotherapy. EPIDs are routinely used to image patient anatomy and verify patient setup prior to, and during treatment. While primarily used for such imaging applications, EPIDs are also suitable dosimeters since the pixel values of acquired images relate to the dose absorbed in the detector. One center has reported on the use of in-house developed software employing back-projection methods to enable routine \textit{in vivo} and pretreatment EPID dosimetry for IMRT deliveries\(^\text{[2-4]}\). Other methods for pretreatment EPID dosimetry have been developed which include absolute dose prediction using a modified algorithm of a treatment planning system\(^\text{[5]}\) and conversion of EPID portal dose images to dose planes in water using mathematical methods\(^\text{[6-9]}\). A literature review summarizing EPID dosimetric characteristics and procedures for the calibration and clinical use of EPIDs for dosimetry has been provided by van Elmpt \textit{et al}\(^\text{[10]}\).

A number of arguments support the integration of EPID dosimetry into routine clinical practice. Since the early 2000s, linear accelerator (Linac) vendors have supplied a-Si EPIDs with the necessary hardware mounted directly to the gantry, in line with the megavoltage (MV) treatment x-ray source. This configuration provides a readily available mechanism to detect the MV beam with minimal setup required. Furthermore, this arrangement provides the only means to directly monitor patient position and dose delivery from the beam’s-eye view. When compared to alternative 2D dosimeters such as arrays of diodes or ion chambers, a-Si EPIDs offer increased spatial resolution (typically 0.4 × 0.4 mm\(^2\)) and real-time data acquisition capabilities\(^\text{[11]}\). Additionally EPIDs are resilient to radiation-induced damage with a response that is both linear with dose and independent of dose rate\(^\text{[12-14]}\).

One of the main factors acting to complicate EPID uses for dosimetry is the nonwater-equivalent response of commercially available detectors\(^\text{[15-18]}\). Such EPIDs indirectly detect incident radiation by means of a metal plate and phosphor screen (typically copper, Cu, and terbium-doped gadolinium oxysulfide,
Gd$_2$O$_2$S:Tb, respectively). Photon and electron interactions deposit energy within the phosphor screen, causing the emission of scintillation optical wavelength photons that may subsequently be detected by the underlying array of a-Si photodiodes. The high atomic number (Z) materials result in increased detector sensitivity (relative to water) to lower energy x-rays where photoelectric absorption events dominate. EPIDs designed in this manner have been shown to be input quantum limited with a detective quantum efficiency (DQE) of $\approx 1\%$\textsuperscript{[19]}. By removing the metal plate and phosphor layers, the EPID may be used to directly detect incident x-rays with a near water-equivalent response\textsuperscript{[20,21]}, albeit with a detection efficiency reduced by approximately 90%\textsuperscript{[12,17]}. Other studies have shown that the EPID DQE at radiotherapy energies may be significantly increased to greater than 20% (zero spatial frequency) by replacing the phosphor screen with thick, segmented scintillators\textsuperscript{[22,23]}. Examples include a linearly scanning array of ZnWO$_4$ crystals that were individually coupled to photodiodes and read-out electronics\textsuperscript{[24]}, arrays of CsI:Tl crystals incorporated into a CCD camera-based EPID\textsuperscript{[25,26]} and directly coupled to a flat panel imager\textsuperscript{[27]}, a thick segmented 2D array of Gd$_2$O$_2$S:Tb phosphor with an underlying AMPFI\textsuperscript{[28]}, and more recently even thicker (up to 40 mm) segmented arrays of CsI:Tl and BGO coupled to an AMPFI\textsuperscript{[22,23]}. The use of a thick, segmented scintillator can greatly increase the DQE while maintaining a high spatial resolution; however such high-Z materials will still result in a nonwater-equivalent detector response. By replacing the high-Z scintillators in these designs with a low-Z material such as plastic scintillator, the detector response may become almost water-equivalent and therefore more useful for dosimetry.

The physical characteristics of plastic scintillating materials and their applications in radiotherapy have been extensively studied and reported in the literature\textsuperscript{[29–36]}. These scintillators are manufactured using low-Z materials and have been shown to respond in a nearly water-equivalent manner to both x-ray and electron beams in the energy range relevant for radiotherapy\textsuperscript{[29,30,33,34]}. In particular, plastic scintillators have a dosimetric response closer to that of water than air, lithium fluoride, and silicon (the active components of ionization chambers, thermoluminescent detectors, and diodes, respectively)\textsuperscript{[29,33]}. These detectors exhibit minimal temperature dependence\textsuperscript{[29]}, excellent resistance to radiation-
induced damage\cite{29} and respond in a stable and reproducible manner that is linear with dose and independent of dose rate\cite{30}.

Various prototype dosimeters that use plastic scintillation detectors (PSDs) have been reported in the literature\cite{29,30,35,37,38}. Early work involving the development of a planar detector to characterize brachytherapy dose distributions used sheets ($< 1$ mm thick) of plastic scintillator in alignment with an optical image intensifier and CCD camera\cite{29}. While the authors acknowledged that the spatial resolution of such a system was limited by light transport throughout the detector, it was found that plastic scintillator was potentially free from energy-response artifacts unlike thermoluminescent and diode detectors. Other dosimeter designs reported in the literature include small PSDs coupled to fiber optic cables that guide scintillation light to a photo-sensitive detector, such as a photomultiplier tube\cite{29,30} or CCD camera\cite{35,37,38}. While a significant source of noise in these systems was the generation of Čerenkov light within the fiber optic cables, many techniques have been reported to correct for this phenomenon\cite{32} (including background subtraction\cite{29}, temporal separation\cite{40}, and chromatic filtration\cite{41}). The generation of Čerenkov light directly within the plastic scintillator is estimated to be about 3 orders of magnitude lower than that of scintillation light\cite{42}. Therefore, by directly coupling the plastic scintillator fibers to a photo-sensitive detector such as the photodiode array in commercial $a$-Si EPIDs, the contribution of Čerenkov light as a source of noise may be drastically reduced.

The overall goal of this study is to experimentally evaluate the imaging performance and dosimetric response of a novel prototype $a$-Si EPID relative to the standard, commercially available detectors. An EPID that responds in a water-equivalent manner would enable portal images to be used clinically not only for monitoring patient positioning and motion but also for real time monitoring of the dose being delivered to the patient. Such a system would enable clinicians to acquire more information about the treatment delivery, without the use of any additional equipment or setup time. The prototype EPID under investigation in this study employs an array of plastic scintillator fibers coupled directly to the photodiode array, in place of the standard metal plate and phosphor screen. This first prototype provides a proof of principle for simultaneous imaging and dosimetry and is hypothesized to exhibit a water-equivalent dosimetric response.
6. Methods and Materials

6.2 Detector design and settings

All EPID measurements reported in this study were performed using 6 MV photon beams delivered by an Elekta Synergy Linac (Elekta, Crawley, UK) with a nominal dose rate of 540 monitor units (MU) per minute. Unless otherwise stated, all beam field sizes and positions are defined with respect to the isocenter, located at a source-to-detector distance (SDD) of 100 cm. Both the experimental prototype and standard configurations employ the same XRD 1640 AN CS flat panel imaging device (PerkinElmer, Santa Clara, CA). This detector, which consists primarily of a $40 \times 40\,\text{cm}^2$ a-Si photodiode array, is used routinely as the photosensitive component of EPIDs on medical Linacs. The array comprises 1024 $\times$ 1024 pixels, giving a pixel pitch of approximately 0.4 mm. Thin-film transistors on the photodiodes allow integration and read out of the charge stored in individual pixels. The XIS software package (PerkinElmer, Santa Clara, CA) was interfaced with this research EPID to acquire all images. The EPID was left to warm up for 20 min prior to each set of measurements in order to reduce the effect of fluctuating dark currents. Dark field corrections were obtained by integrating the EPID signal for 30 frames with the beam off, prior to collecting images. This dark field was subtracted from subsequent open field images and was updated regularly during experiments. Flood field corrections were obtained as a frame-averaged exposure of a region-of-interest (ROI) on the EPID to a beam sufficiently large enough to cover the ROI. For the standard EPID configuration this ROI was the entire detector; however, for the prototype EPID configuration this ROI was of an area slightly smaller than, and centered on, the plastic scintillator array. Dark field corrected images were then divided by the flood fields to correct for nonuniformities in individual pixel sensitivities across the photodiode. Dead pixel corrections were applied in all measurements.

6.2.1.1 Standard EPID configuration

The standard EPID configuration consists of the research EPID setup as it is used routinely in radiotherapy clinical practice. This design incorporates front
and rear aluminum covers, as well as a 1 mm thick copper sheet and 133 mg cm$^2$ Gd$_2$O$_2$S:Tb phosphor scintillator screen (Lanex Fast Back, Carestream Health, Inc. Rochester, USA) that is coupled to the $a$-Si photodiode array (see Figure 6.1(a)). The copper acts as buildup for the primary beam and filters low energy scattered photons and electrons. Energy deposited in the phosphor is converted into scintillation optical wavelength photons, which are then absorbed by the photodiode leading to charge integration in individual pixels. The research EPID, when configured in this manner, shall be referred to as the “standard EPID” throughout the remainder of this paper.

Figure 6.1: Schematic illustration of the main components of (a) the standard clinical EPID incorporating a metal plate/phosphor screen and (b) the prototype experimental EPID incorporating a segmented plastic scintillator array. The same $a$-Si photodiode array was employed in both configurations. Schematics are not drawn to scale and the gaps located between neighboring layers are for illustrative purposes only.
6. A novel EPID for imaging & dosimetry

6.2.1.2 Prototype EPID configuration

The prototype EPID configuration utilizes a segmented plastic scintillator array in place of the copper sheet and phosphor screen described above for the standard EPID (see Figure 6.1(b)). The segmented plastic scintillator array was constructed using square fibers, each having a cross-sectional area of $1 \times 1 \text{mm}^2$ and 15 mm length. The fibers are oriented parallel to each other and as such are not focused toward a particular point in space (e.g., the radiation source). Figure 6.2 shows an image of the array. The fibers used (BCF-99-06A) are made of a polystyrene base and are the fiber analogue of BC-430 plastic scintillator. Each fiber was optically isolated from its neighbors by a polymethylmethacrylate (PMMA) cladding having a thickness of approximately 4% of the fiber width and a $10 - 15 \mu m$ thick coating of white Extra Mural Absorber (EMA). EMA is a coating applied to the outer surface of each fiber to minimize crosstalk between adjacent fibers by reflecting or attenuating light that escapes through the cladding. The top layer of the array (closest to the radiation source) was also covered with a thin film of reflective material (Vikuiti™ Enhanced Specular Reflector (ESR) from 3M, St Paul, MN). This ESR is a multilayer polymer film that redirects light arriving at the upper surface of the plastic array back into the fibers and toward the photodiodes. The plastic array comprised 150 $\times$ 150 such fibers, giving a total cross-sectional area of approximately $150 \times 150 \text{mm}^2$. This prototype was constructed by Saint-Gobain Crystals (Saint-Gobain Crystals, Hiram OH, USA) to our specifications. It should be noted that because the cross-sectional area of the plastic scintillating fibers is greater than the area of the photodiode pixels, misalignment of the fibers and photodiodes was inevitable.

To reassemble the research EPID into this configuration, the front aluminum cover and all components above the photodiode were first removed and the segmented plastic scintillator array was then placed directly on top of the photodiodes. When configured in this manner, the research EPID shall be referred to as the “experimental EPID” throughout the remainder of this paper. For some experiments it was necessary to stand the EPID on its side. In these cases, the plastic scintillator array was also positioned on its side and was placed on top of Styrofoam blocks to elevate it approximately to the level of the EPID’s center.
Figure 6.2: (a) Photograph of the segmented plastic scintillator array used in the experimental EPID. The segmented structure is clearly visible in the magnified view (b). Each segment has a cross-sectional area measuring $1 \times 1 \text{mm}^2$, is 15 mm long, and is surrounded by an optically reflective coating.

Tape was fixed from the outer frame of the plastic scintillator array to the EPID edges, to maintain direct contact between the plastic fibers and the photodiode array. It should be noted that any residual air gaps between the plastic scintillator array and the a-Si photodiode may affect the response of the experimental EPID. While measures were taken to minimize the occurrence of such air gaps, a detailed investigation of their effects on the detector response was beyond the scope of this study (see also Section 8.2.3.1).

### 6.2.2 Detector sensitivity, linearity and pixel noise

The detector sensitivity and linearity of dose response were measured at the central axis of a $10 \times 10 \text{cm}^2$ field for both EPID configurations. The EPID was placed on the treatment couch at a SDD of 100 cm with the gantry at $0^\circ$. Unless otherwise stated, the gain and integration times were adjusted to maximize the signal for each configuration without saturating the detector and the acquired images were both dark field and flood field corrected. The integrated pixel re-
6. A novel EPID for imaging & dosimetry

Responses were recorded for exposures of 1 – 100 MU and were calculated from the mean pixel value sampled over the central 24 × 24 pixels. Uncertainties were calculated as the standard deviation of the response within this region. The detector was operated in free running mode, where each frame is acquired at a preset frequency. An integrated image was obtained by simply summing the individual frames. The sensitivity of the experimental EPID was evaluated relative to that of the standard EPID by comparing the integrated pixel values per MU for each configuration. The linearity of dose response for each configuration was calculated from the integrated pixel values per MU as a function of MU number over the range investigated. The pixel noise was calculated as the fractional uncertainty (standard deviation divided by mean) of the central 24 × 24 pixels.

6.2.3 Dose response evaluation

The dose responses of the standard and experimental EPID configurations were evaluated relative to that of an ionization chamber array in nontransit and transit dosimetry setups. The MatriXX ionization chamber array (IBA Dosimetry Asia Pacific, Beijing, China) was used with 1.2 cm of solid water build-up to give a measurement depth of 1.6 cm (approximately equivalent to that of the experimental EPID). The MatriXX ionization chamber array has a dose response equivalent to a Farmer ionization chamber in water-equivalent material for the experimental geometries used in this study [21,43–45].

In the nontransit setup, the off-axis response and field size response were measured for each detector. For these measurements, each detector was positioned on the treatment couch with the gantry at 0° and a SDD of 100 cm. The EPID response was calculated from frame-averaged images consisting of 30 frames.

In the transit setup, square sheets of solid water were used to create a phantom on the treatment couch, centered about the isocenter. The gantry was rotated to 90° and each detector was placed on its side on the treatment couch with a SDD of 160 cm (couch rotated 90°). This configuration was chosen because it provided a simple and reproducible way to position the phantom material, EPID and MatriXX detector. The off-axis response, field size response, and transmission factors for each detector were measured. The EPID response was calculated using
frame-averaged images with 50 frames.

For all measurements taken with the MatriXX array, the mean response was calculated within the central $3 \times 3$ ionization chambers for field sizes greater than $3 \times 3 \text{cm}^2$. For smaller fields, the response of the single central chamber was used. The MatriXX array was aligned with one central detector located at the central axis. For all EPID dosimetry measurements, response was calculated as the mean pixel value within the central $24 \times 24$ pixels, with uncertainties taken as the standard deviation of pixel values in this region, unless otherwise stated. EPID gain and integration times were adjusted to maximize the signal for each configuration without saturating the detector. Unless otherwise stated, acquired images were both dark field and flood field corrected.

6.2.3.1 Off-axis response

The nontransit off-axis dose response was measured with static $10 \times 10 \text{cm}^2$ fields centered at positions of 0, 5, 10, and 15 cm along the cross-plane collimator axis. The transit off-axis dose response was measured using static $7 \times 7 \text{cm}^2$ fields centered at positions of 0, 5, 10, 15, and 20 cm (defined at the EPID position with a SDD of 160 cm) along the cross-plane collimator axis with a solid water phantom thickness of 20 cm. The off-axis fields were created using asymmetric jaw settings. In both nontransit and transit configurations, the EPID was shifted in the cross-plane direction with each field such that the same region of the detector was being irradiated. This method was used to overcome field size limitations due to the $15 \times 15 \text{cm}^2$ square size of the segmented plastic scintillator array and to remove any response variations due to measuring with different regions of the detector. In order to maintain the off-axis response these images were not flood-field corrected. Reference measurements were taken using the MatriXX ionization chamber array, placed at the same positions along the cross-plane collimator axis. Each ionization chamber measurement was taken with irradiations of 100 MU. All measured detector responses were normalized relative to the 0 cm field position (when the $10 \times 10 \text{cm}^2$ field was symmetric about the beam central axis).
6.2.3.2 Field size response

For the nontransit field size response measurements, static square fields ranging in size from $2 \times 2$ to $15 \times 15$ cm$^2$ were centered on the beam central axis. For the EPID measurement taken with the smallest field size, the response was calculated as the mean value of the central $6 \times 6$ pixels. Reference measurements were taken by delivering 100 MU for each field size with the MatriXX ionization chamber array placed at the isocenter in the same orientation as the EPID. All measurements were normalized to the reference $10 \times 10$ cm$^2$ field.

For the transit setup, the solid water phantom had a fixed thickness of 20 cm. Static square fields ranging in size from $2 \times 2$ to $9 \times 9$ cm$^2$ were centered on the beam central axis. With the EPID positioned at a SSD of 160 cm, a field size measuring $9 \times 9$ cm$^2$ was the largest that could be used with the experimental EPID due to size limitations of the plastic scintillator array. For the EPID measurements response was calculated as described above. Reference measurements taken with the MatriXX detector were obtained by delivering 300 MU for each field size. All response measurements were normalized to the value measured with the $9 \times 9$ cm$^2$ field.

6.2.3.3 Transmission factors

Transmission factors were determined using static fields measuring $9 \times 9$ cm$^2$ and centered on the beam central axis. Sheets of solid water were positioned on the treatment couch to create solid water objects with thicknesses ranging from 0 to 40 cm in 10 cm increments and centered about the isocenter. For the EPID measurements response was determined as described above. Measurements taken with the MatriXX detector were obtained by delivering 200 MU (0 and 10 cm solid water thickness), 300 MU (20 cm solid water thickness), 400 MU (30 cm solid water thickness), or 500 MU (40 cm solid water thickness) and normalizing the response to the number of MU delivered. Transmission factors were then determined by normalizing the response per MU measured with each detector to the value obtained with no solid water in the beam.
6.2.4 Image quality evaluation

The imaging performance of the experimental EPID was evaluated and compared to that of the standard EPID. Imaging performance was measured using a PipsPro QC-3V image quality phantom with associated software (Standard Imaging, Inc., Middleton, WI) and an anthropomorphic head phantom to acquire images with each EPID configuration for qualitative and quantitative analysis. Unless otherwise stated, images were acquired with the gantry rotated to 90° and with the phantom centered on the isocenter on the treatment couch. Each EPID configuration was placed on its side on the treatment couch behind the phantom (with the couch rotated 90°). As discussed in Section 6.2.3, this setup was the most simple and reproducible method available. A gain setting of 4 pF and frame integration time of 133 ms were used and all images were both dark field and flood field corrected.

6.2.4.1 QC-3V Phantom

The PipsPro QC-3V image quality phantom and software were used to quantitatively compare the spatial resolution and contrast-to-noise ratio (CNR) of the standard and experimental EPIDs. The phantom was placed on its side on the treatment couch with its upper surface located 90 cm from the source. The EPID, also placed on its side, was located at a SDD of 100 cm.

For each EPID configuration, separate images of the phantom were acquired using a single frame and an average of 50 frames, representing low and high dose levels approximately equivalent to 1 and 66 MU, respectively. These dose levels represent the range of low and high dose imaging applications.

PipsPro software was used to calculate the spatial resolution and CNR for images acquired with each combination of EPID configuration and dose levels. The spatial resolution was calculated based on $f_{50}$ and $f_{30}$ (the spatial frequencies, measured in units of lp/mm, at which the relative modulation transfer function is equal to 50% and 30%, respectively). For details on the calculation of $f_{50}$, $f_{30}$ and CNR the reader is referred to Rajapakshe et al.\cite{Rajapakshe2016}. 
6. A novel EPID for imaging & dosimetry

6.2.4.2 Anthropomorphic phantom

Images of an anthropomorphic head phantom were acquired to qualitatively evaluate the performance of the experimental EPID configuration, relative to the standard configuration, when imaging an object representative of human anatomy. The head phantom was positioned in an anterior-posterior orientation on the treatment couch and was centered at the isocenter. The EPID was placed at a SDD of 160 cm. For each EPID configuration, an image of the phantom was acquired by averaging 50 individual frames.

6.3 Results and Discussion

6.3.1 Detector sensitivity, linearity and pixel noise

For a given detector gain and integration time setting, the sensitivity of the experimental EPID was found to be approximately 0.37 times that of the standard EPID. This is to be expected since the Gd$_2$O$_2$S:Tb has a significantly higher scintillation yield (on average 60 optical photons per keV of deposited energy\textsuperscript{[47]} than the plastic scintillator (on average eight optical photons per keV, as provided by the manufacturer). Furthermore, the total photon cross sections and electron linear stopping power for Gd$_2$O$_2$S:Tb are greater than those for the plastic scintillator over the energy range relevant for radiotherapy. While these properties are partially countered by increasing the thickness of the plastic relative to the phosphor screen, determining the relationship between fiber dimensions and the detector sensitivity is the subject of ongoing investigations. Uneven physical contact between the fibers and the photodiode array may also affect detector sensitivity, as well as the mismatch between refractive indices for the plastic scintillating fibers (1.6 as provided by the manufacturer) and the a-Si photodiodes (4.6 for the peak optical emission wavelength\textsuperscript{[48]}), respectively. Methods to improve the optical transfer efficiency between the fibers and photodiodes are the subject of ongoing work.

Figure 6.3 illustrates the pixel values per MU measured at the center of the standard and experimental EPIDs for beam deliveries ranging from 1 to 100 MU. Values have been normalized to that measured with a 100 MU beam delivery. It
was found that for deliveries greater than 8 MU, the standard and experimental EPID pixel values agreed very closely with a mean difference of only 0.1%. For deliveries with fewer MU, agreement between the standard and experimental EPIDs was still within 1.6%. Overall, both configurations exhibited a response that was highly linear with delivered dose. The maximum deviation from unity was 3.3% and 1.7% for the standard and experimental EPIDs, respectively, occurring for a dose delivery of 3 MU (taking the experimental EPID response at 6 MU as an outlier). For dose deliveries above 15 MU, the normalized pixel values per MU for both EPID configurations were all within 1% of unity. The dependence of the measured pixel values per MU on the number of MUs delivered has been previously reported to be a consequence of image lag and gain ghosting effects attributable to charge trapping in a-Si based EPIDs\cite{49–51}.

Images uncorrected for gain (i.e., without the application of a flood field correction) were much noisier for the experimental EPID than for the standard EPID, with fractional uncertainties within the central region reaching 14.4% and 0.2%, respectively. The additional noise for the prototype EPID is due to the size differ-
6. A novel EPID for imaging & dosimetry

ence between the cross-sectional area of the plastic scintillator fibers (1×1 mm$^2$) and the area of individual photodiode pixels (0.4×0.4 mm$^2$). This difference, combined with the irregularities in the array construction seen in Figure 6.2(b), results in an imperfect alignment of the fibers and detector pixels. After applying the gain correction, the pixel noise decreases to 0.2% and 0.05% for the experimental and standard EPIDs, respectively. The generation of Čerenkov light within the plastic scintillator of the experimental EPID is expected to be several orders of magnitude smaller than the scintillation signal\cite{42}. PSDs suffer from noise due to Čerenkov radiation generated in the optical fibers that transport the optical signal from the scintillator to a remote light sensor\cite{32}. The experimental EPID detector does not involve the transport of light outside of the radiation field to a remote light sensor and therefore does not suffer the same problem. Work is planned to extend a previously validated Monte Carlo model of a standard EPID to investigate this phenomenon and its impact on the prototype EPID design\cite{52}.

6.3.2 Dose response evaluation

6.3.2.1 Off-axis response

The detector response at positions away from the central axis of the x-ray beam is shown in Figure 6.4 for a reference MatriXX detector and the standard and experimental EPID configurations in nontransit and transit geometries. Note that for this and all subsequent figures, error bars that are not visible are smaller than their corresponding data points. At the greatest off-axis distance (15 cm) in the nontransit geometry, the experimental EPID’s response is only 1.9% greater than that of the ionization chamber, whereas the standard EPID’s response is 18% greater than that of the ionization chamber. It is well documented that the energy spectrum of a 6 MV clinical x-ray beam becomes softer at positions away from the central axis of the beam\cite{53}. Given the increased sensitivity to low energy radiation exhibited by Gd$_2$O$_2$S:Tb relative to water, the off-axis over-response observed with the standard EPID was therefore to be expected. The close agreement in off-axis response observed between the experimental EPID and the ionization chamber provides support that the prototype EPID responds in a more water-equivalent manner, making it more suitable for dosimetry ap-
6.3. Results and Discussion

The large off-axis over-response that occurs when using the standard EPID implies that portal dose images acquired with a standard EPID must be manipulated in order to convert the dose measured in the EPID to an equivalent dose-in-water measurement. Using a water-equivalent dosimeter such as the experimental EPID greatly simplifies dosimetry calculations, since manipulations to the portal dose images are not required.

![Graphs showing off-axis ratios measured using the MatriXX detector (ion chamber), the standard EPID and the experimental EPID.](image)

Figure 6.4: Comparison of (a) nontransit and (b) transit off-axis ratios measured using the MatriXX detector (ion chamber), the standard EPID and the experimental EPID.

In the transit geometry, the greatest difference in response between the experimental EPID and the ionization chamber array was 1.2%, occurring at the off-axis distance of 15 cm. The greatest difference in response between the standard EPID and the ionization chamber array was 5.6%, occurring at the greatest off-axis distance of 20 cm. These results are still consistent with a softer x-ray spectrum at positions away from the central axis; however, the presence of the phantom acts to change the incident beam energy spectrum by increasing the ratio of scattered radiation to primary radiation incident on the detectors. This causes the observed flattening of the off-axis ratios relative to those observed.
in the nontransit geometry. Despite the lower off-axis ratios measured in the transit configuration, the experimental EPID is still seen to respond in a more water-equivalent manner than the standard EPID.

6.3.2.2 Field size response

Figures 6.5(a) and (b) show the responses of a reference MatriXX ionization chamber array and the standard and experimental EPIDs with beam field size, in nontransit and transit geometries, respectively. In the nontransit geometry, the field size response of the experimental EPID more closely matches that of the MatriXX detector than does the response of the standard EPID. The maximum percent differences in response between the experimental EPID and the MatriXX, and between the standard EPID and the MatriXX, were 1.2\% and 6.2\%, respectively, across all the studied field sizes. Once again, the disagreement observed between the response of the standard EPID and the MatriXX may be attributed to the increased sensitivity of the standard EPID to low energy radiation. As the x-ray beam field size increases, there is a greater proportion of low energy x-rays within the beam spectrum which causes the standard EPID to over-respond relative to the ionization chambers and experimental EPID.

Figure 6.5(b) shows that even in a transit dosimetry configuration, the field size response of the experimental EPID more closely matches that of the ionization chambers than does the field size response of the standard EPID. In this case, the maximum percent differences in response between the experimental EPID and ionization chambers, and between the standard EPID and ionization chambers were 1.2\% and 5.0\%, respectively. In both the nontransit and transit configurations, the experimental EPID was therefore found to exhibit a field size response that was water-equivalent whereas the standard EPID was not.

6.3.2.3 Transmission factors

Transmission factors measured with the reference ionization chamber array and the standard and experimental EPIDs in a transit dosimetry configuration are shown in Figure 6.6. The transmission factors measured with the experimental EPID closely match those measured with the MatriXX ionization chamber array,
6.3. Results and Discussion

Figure 6.5: Comparison of (a) nontransit and (b) transit field size output factors measured using a MatriXX ionization chamber array, the standard EPID and the experimental EPID.

with a maximum percent difference in response of only 2.5%. Agreement between the response measured using the standard EPID and the MatriXX ionization chamber array was worse, with a maximum percent difference in response of 13%. Once again, the difference in response between the standard and experimental EPIDs may be attributed to the higher sensitivity of the standard EPID to the low energy component of the x-ray beams. By increasing the thickness of solid water in the beam, more of the low energy photons are filtered from the beam and the standard EPID’s response becomes closer to that of the experimental EPID and ionization chamber.\footnote{Further to the primary beam’s hardening as it passes through the phantom, low energy scattered photons are also being generated in the solid water. However, this generation of low energy photons has a much slower ‘buildup’ as compared to the total dose buildup as seen in a classical percentage depth dose curve. These details, while not included in the published version of this chapter\cite{54}, have been included here for completeness.} However, the very close agreement observed between the experimental EPID and the ionization chamber provides further evidence that this prototype EPID maintains a water-equivalent dosimetric response.
Figure 6.6: Comparison of transmission factors measured using a MatriXX ionization chamber array, the EPID in its standard configuration and the experimental EPID with the segmented plastic scintillator.

These dose response evaluation results demonstrate that the experimental EPID exhibits a significantly more water-equivalent response than the standard EPID in both nontransit and transit configurations. As Linac mounted kV systems for 2D and 3D imaging become standard, the clinical demand for MV imaging has reduced. Meanwhile, the justification and potential for EPID-based in vivo dosimetry continues to grow\cite{4,55}. It is possible that the primary function of EPIDs will shift more towards dosimetry in the future. The prototype detector reported in this study serves as a promising example of a water-equivalent EPID that would be better suited for clinical dosimetry than current EPIDs.
6.3.3 Image quality evaluation

6.3.3.1 QC-3V phantom

Figure 6.7 shows images of the QC-3V image quality phantom taken with the standard and experimental EPIDs. The most immediately apparent observation when comparing these images is that the image acquired using the experimental EPID suffers from additional blurring relative to that taken with the standard EPID. This blurring primarily results from the misalignment between the fibers and photodiode pixels.

Figure 6.7: High dose (50 frame-averaged) images of the QC-3V image quality phantom (with dark and flood field corrections applied) taken with (a) the standard EPID and (b) the experimental EPID.

Table 6.1 summarizes the quantitative comparison of the image quality obtained using the standard and experimental EPIDs. Images identical to those shown in Figure 6.7, acquired as frame-averaged images with either 1 or 50 individual frames, were analyzed using PipsPro software to calculate their CNR and spatial resolution ($f_{50}$ and $f_{30}$).

The CNR of the images taken with the standard EPID were greater than that of the images taken with the experimental EPID. For both detectors, the CNR increased with dose (the number of frames), as expected. The decreased CNR
Table 6.1: Quantitative comparison of the CNR and spatial resolution of the standard and experimental EPID configurations.

<table>
<thead>
<tr>
<th>EPID configuration</th>
<th>CNR</th>
<th>Spatial resolution</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>1 frame</td>
<td>50 frames</td>
</tr>
<tr>
<td>Standard</td>
<td>195</td>
<td>1269</td>
</tr>
<tr>
<td>Experimental</td>
<td>47.8</td>
<td>819</td>
</tr>
</tbody>
</table>

measured in images taken with the experimental EPID is consistent with this detector’s reduced sensitivity and increased noise, as measured relative to the standard EPID. However, the differential increase in CNR with number of frames is larger for the experimental EPID (\( \Delta \text{CNR}/\text{CNR} = 16.1 \)) than the standard EPID (\( \Delta \text{CNR}/\text{CNR} = 5.51 \)) over the range 1–50 frames.

The spatial resolution was also better in images acquired using the standard EPID relative to those obtained with the experimental EPID. This result agrees with the qualitative description of the images in Figure 6.7 and again is largely a consequence of the greater cross-sectional area of the scintillating fibers relative to the photodiode pixels in the experimental EPID.

6.3.3.2 Anthropomorphic phantom

Anterior-posterior projection images of an anthropomorphic head phantom acquired using the standard and experimental EPIDs are shown in Figure 6.8. These images provide an example of the ability for the standard and experimental EPIDs to visualize anatomical structures. In the image taken using the experimental EPID, the structure of the plastic scintillating fiber array is manifested as a persistent grid-like pattern overlaying the image of the phantom.

Based on these image quality results, it is hypothesized that the imaging performance of the experimental EPID configuration may be improved by optimizing certain features of the plastic scintillator array. Features including the fiber dimensions, structural uniformity of the array, alignment of the fibers and photodiodes, and focusing the fibers to the radiation source likely impact the spatial resolution of acquired images. Optimization of the fiber lengths, mate-
6.4 Conclusion

In this study, a prototype EPID employing an array of plastic scintillating fibers in place of the standard copper plate and Gd$_2$O$_2$S:Tb phosphor screen has been developed and the first experimental measurements have been reported. In contrast to the standard EPID, the prototype EPID exhibited a near water-equivalent dosimetric response. While the imaging performance of the standard EPID was superior to that of the prototype, there are promising opportunities for design optimization of the prototype to improve imaging performance while maintaining a water-equivalent dose response. Monte Carlo radiation transport simulations will be used in future studies to quantify how features such as the dimensions of the plastic scintillating fibers and alignment of the fibers with photodiode pixels may improve image quality. The performance of the prototype reported in this work demonstrates the feasibility and potential of this experimental EPID as a
6. A novel EPID for imaging & dosimetry

next-generation device for simultaneous imaging and dosimetry in radiotherapy.

Acknowledgments

The authors would like to acknowledge funding support from the Cancer Institute NSW (Research Equipment Grant 10/REG/1-20) and Cancer Council NSW (Grant ID RG 11-06). S.J.B. would also like to thank The University of Sydney and the Institute of Medical Physics for scholarship support, as well as the Liverpool and Macarthur Cancer Therapy Centers for additional financial support. Thanks also to Rob Saunders of Nucletron for facilitating prototype design discussions with Saint-Gobain Crystals. The authors report no conflicts of interest in conducting the research.
References


References


42. Teymurazyan, A. & Pang, G. Monte Carlo simulation of a novel water-equivalent electronic portal imaging device using plastic scintillating fibers. *Medical Physics*
References


Optimisation of the imaging and dosimetric characteristics of an EPID employing plastic scintillating fibres using Monte Carlo simulations

The work presented in this chapter describes the design and implementation of a Monte Carlo model developed to optimise the prototype EPID.
7. MC optimisation of a prototype EPID

Statement of joint authorship

This work has been published in *Physics in Medicine and Biology* as:


<table>
<thead>
<tr>
<th>Author</th>
<th>Specific involvements</th>
</tr>
</thead>
<tbody>
<tr>
<td>S.J. Blake</td>
<td>Developed, built and ran simulation code, analysed results, wrote manuscript (80%)</td>
</tr>
<tr>
<td>A.L. McNamara</td>
<td>Aided simulations, aided analysis, edited manuscript (7%)</td>
</tr>
<tr>
<td>P. Vial</td>
<td>Aided analysis, edited manuscript (3%)</td>
</tr>
<tr>
<td>L. Holloway</td>
<td>Aided analysis, edited manuscript (3%)</td>
</tr>
<tr>
<td>Z. Kuncic</td>
<td>Intellectual contribution toward Monte Carlo modelling, aided simulations, aided analysis, edited manuscript (7%)</td>
</tr>
</tbody>
</table>
Abstract

A Monte Carlo model of a novel electronic portal imaging device (EPID) has been developed using GEANT4 and its performance for imaging and dosimetry applications in radiotherapy has been characterised. The EPID geometry is based on a physical prototype under ongoing investigation and comprises an array of plastic scintillating fibres in place of the metal plate/phosphor screen in standard EPIDs. Geometrical and optical transport parameters were varied to investigate their impact on imaging and dosimetry performance. Detection efficiency was most sensitive to variations in fibre length, achieving a peak value of 36% at 50 mm using 400 keV x-rays for the lengths considered. Increases in efficiency for longer fibres were partially offset by reductions in sensitivity. Removing the extramural absorber surrounding individual fibres severely decreased the modulation transfer function (MTF), highlighting its importance in maximising spatial resolution. Field size response and relative dose profile simulations demonstrated a water-equivalent dose response and thus the prototype’s suitability for dosimetry applications. Element-to-element mismatch between scintillating fibres and underlying photodiode pixels resulted in a reduced MTF for high spatial frequencies and quasi-periodic variations in dose profile response. This effect is eliminated when fibres are precisely matched to underlying pixels. Simulations strongly suggest that with further optimisation, this prototype EPID may be capable of simultaneous imaging and dosimetry in radiotherapy.
7. MC optimisation of a prototype EPID

7.1 Introduction

Patient-specific radiotherapy treatment deliveries are becoming increasingly complex and treatment plans often involve steep dose gradients that are sensitive to intra-fraction patient motion and inter-fraction changes in patient anatomy. Dose verification of treatment deliveries thus presents a highly desirable means of monitoring treatment progression and catching potential errors occurring throughout the radiotherapy process.

Modern amorphous silicon ($a$-Si) electronic portal imaging devices (EPIDs) are frequently used for imaging applications in radiotherapy\[1\]. A common example includes verifying patient setup prior to treatment delivery\[2\] and they have also been applied to real-time tracking of tumour position using fiducial markers\[3\]. With such image-guided radiotherapies becoming increasingly commonplace, the megavoltage (MV) imaging capabilities of EPIDs have made them indispensable devices in radiotherapy.

Interest in using modern $a$-Si EPIDs for dose verification continues to grow\[4-6\]. This is largely due to their high spatial resolution, real-time readout capabilities, resilience to radiation-induced damage and seamless integration with modern clinical linear accelerators\[4,7,8\]. The high atomic number ($Z$) components that are used to optimise EPID imaging performance, however, cause a non-water equivalent dose response that complicates the interpretation of portal images for dosimetry\[4,9,10\]. While some groups have successfully used $a$-Si EPIDs for pre-treatment and in-vivo dosimetry\[11\], EPID dosimetry is still far from becoming routine in clinical practice. The complex dose response presents a major challenge for the development of standardized commercial software support.

The detective quantum efficiency (DQE) of an imaging system is a widely accepted measure of the efficiency with which a detector converts incident x-rays into an image signal\[1\]. Novel EPIDs have been developed that exhibit DQEs significantly greater than those of standard EPIDs. Examples include EPIDs using thick, segmented phosphors\[12\] and crystal scintillators\[13,14\] in place of the standard metal plate and phosphor screen. Using an array of CsI:Tl crystals 40 mm thick, Sawant et al. reported zero spatial frequency DQEs up to 22% relative to $\approx 1 - 3\%$ for standard EPIDs\[14\]. The authors proposed that a DQE
up to 50% may be achievable with further optimisation. Prototypes employing thick CsI:Tl, BGO and LYSO crystal scintillators have also been characterised and applied to low-dose MV cone beam CT imaging\(^{15,16}\). However while these various prototypes drastically improve upon the current DQE, they do not offer detectors with water-equivalent responses.

Our group has previously characterised a novel, first-generation prototype EPID utilising an array of plastic scintillator fibres in place of the metal plate and phosphor screen of standard EPIDs for simultaneous imaging and dosimetry in radiotherapy\(^{17,18}\). Plastic scintillators respond linearly to dose and independently of dose rate, are resilient to radiation-induced damage and exhibit a water-equivalent dose response\(^{19,20}\). A range of plastic scintillation dosimeters have been developed for radiotherapy applications to take advantage of these characteristics\(^{21,22}\). One study modeled a similar plastic scintillator fibre EPID and reported zero spatial frequency DQEs between 4 – 37% at 6 MV depending on fibre length\(^{23}\). While our first-generation prototype exhibited a reduced sensitivity and spatial resolution relative to standard EPIDs, it exhibited a water-equivalent dose response and many aspects of the prototype were identified for future optimisation of imaging performance\(^{17}\).

In this study we develop a MC model of a novel \(\alpha\)-Si EPID employing a plastic scintillator fibre array (PSA) based on a second-generation prototype designed and recently purchased by our group. This second-generation prototype is hypothesised to exhibit a water-equivalent response because it utilises the same plastic scintillator material as previously studied with the first generation prototype, albeit with a different physical geometry. With this model, we investigated the effects of varying the detector geometry and optical transport parameters on its response, specifically to characterise the prototype’s imaging and dosimetry capabilities. The ultimate goal of this work was to determine the set of parameters that will optimise the detector response and to quantify the sensitivity of the response to sub-optimal parameter values.


7. MC optimisation of a prototype EPID

7.2 Methods and Materials

7.2.1 Monte Carlo source model and EPID geometry

All MC simulations were performed on a computer cluster comprised of 252 × 2.67 GHz CPUs, with parallel processing implemented using openMPI. *

A MC source model of a 6 MV photon beam was used to generate phase space files for the different field sizes investigated (3 cm × 3 cm to 10 cm × 10 cm). A detailed description of this source model may be found in earlier publications.[24,25] Briefly, egsnrc (V4 2.3.1)[26] was used with user code BEAMnrc (V4 2.3.1)[27] to build a model of an Elekta Synergy 6 MV photon beam. The source model was used to generate a phase space file for each beam field size by scoring all particles that traversed a plane 89.5 cm from the target. These files were then used as input for a GEANT4 MC model of the PSA-EPID. The distance from the target to the PSA-EPID photodiode plane was fixed at 100 cm.

The PSA-EPID model was developed using the GEANT4 MC toolkit (version 9.6 patch 02)[28,29] and was based on a previously validated model of a standard EPID that was also developed using GEANT4[24,25]. The standard EPID model comprised a series of uniform layers representing the individual detector components and was based on the 2D PerkinElmer flat panel imager XRD 1640 AN CS, which has a cross-sectional area measuring 41 cm × 41 cm and 0.4 mm pixel pitch. The key components of the standard EPID model were the copper buildup layer, Gd₂O₂S:Tb phosphor screen and array of a-Si photodiodes. Geometrical and material compositions were based on manufacturer specifications (PerkinElmer, Santa Clara, CA). The array of a-Si pixels was modeled as a uniform, 0.1 mm thick layer of a-Si supported by a 1 mm SiO₂ substrate. The pixelated structure and nonunity pixel fill factor were not explicitly modeled.

The standard EPID model was modified into the PSA-EPID configuration by replacing all layers above the a-Si photodiode array, including the Cu plate, Gd₂O₂S:Tb phosphor screen and aluminum cover, with a 2D array of plastic scintillator fibres. The photodiode layer and all downstream components remained as previously defined in the standard EPID model. The geometry of the PSA

*(http://www.open-mpi.org/)
7.2. Methods and Materials

Figure 7.1: Schematic (not to scale) of the PSA-EPID MC model. (a) Beam’s-eye view of a polystyrene fibre with PMMA cladding surrounded by an EMA. Dashed lines delineate the outer edges of the fibres. (b) A simulated event. Green x-ray, blue electron and magenta optical photon tracks propagate through the PSA-EPID.

was based on a physical prototype currently under experimental investigation by our group, with a total cross-sectional area measuring 15 cm × 15 cm (see Figure 7.1).

Several different fibre geometries were simulated in this study to investigate the impact of fibre geometry on imaging performance and dosimetric response. Fibres in the PSA were square in shape and were composed of a scintillating polystyrene core surrounded by a 20 µm thick PMMA cladding and a 15 µm thick layer of extra-mural absorber (EMA) that was used to prevent optical cross-talk between neighboring fibres (see Figure 7.1). Details of the fibre materials and their physical properties were provided by the manufacturer (Saint-Gobain Crystals, Hiram OH, USA) and are listed in Table 7.1. Throughout the remainder of this paper, the term “reference parameters” refers to the use of simulation parameters with values listed therein. On the other hand, “non-reference parameters” refers to the use of alternate values (listed in Table 7.2) that were chosen to investigate how variations in certain geometrical and physical parameters would affect simulated PSA-EPID imaging and dosimetric response.

The reference fibre length of 30 mm was chosen to match the length of fibres in our physical prototype. Additional fibre lengths of 15 and 50 mm were simulated
7. MC optimisation of a prototype EPID

for comparison. The reference fibre cross-sectional area (including the cladding and EMA layers) measured 0.4 mm × 0.4 mm as this represented an idealised geometry where the individual fibres were precisely aligned to the underlying a-Si photodiode pixels. It was hypothesized that this configuration would optimize spatial resolution. However, because the cross-sectional area of fibres in the physical prototype measured 0.5 mm × 0.5 mm, a PSA geometry with this fibre area was simulated for comparison.

For greater control over the optical processes occurring between the cladding and EMA layers, a GEANT4 G4OpticalSurface was defined at this interface for each fibre. A G4OpticalSurface is used to empirically specify surface characteristics that affect the nature of incident optical photon absorption and scattering, such as surface reflectivity or roughness. Since the purpose of the EMA is to absorb incident optical photons and thus prevent them from being transmitted into neighbouring fibres, the G4OpticalSurface was defined as a perfect absorber. The boundaries between the fibre core/cladding and cladding/EMA were assumed to be smooth, which is a valid assumption for fibres with diameters > 0.1 mm \[^{23,30}\]. The impact of the EMA layer on the detector’s response was investigated by performing simulations with and without this G4OpticalSurface present.

7.2.2 Simulated physics processes

The standard GEANT4 electromagnetic physics classes with a range cut of 1 mm were used for all PSA-EPID simulations (decreasing this parameter to 0.01 mm had a negligible effect on the resulting line spread function (LSF)).

Optical photons were generated in the fibre cores via scintillation and Čerenkov radiation. These photons may undergo boundary processes (reflection and refraction), incoherent (Rayleigh) scattering and bulk absorption. The refractive index \( n \), scattering length \( \mu \) and absorption length \( l \) were specified for each medium and the scintillation yield \( SY \) with optical emission spectrum \( \lambda(E) \) (for photon energy \( E \) and with peak wavelength \( \lambda_{\text{peak}} \)) was specified for the scintillator core. These optical transport parameters and reference values used in this study are summarised in Table 7.1.

Optical properties for the plastic scintillator were obtained from the PSA
7.2. Methods and Materials

Table 7.1: Summary of the reference properties of the PSA-EPID fibres and photodiodes used in the MC model. Unless otherwise stated, values were provided by the manufacturer (Saint-Gobain Crystals, Hiram OH, USA). Properties listed in parentheses had no information provided.

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fibre length</td>
<td>30 mm</td>
</tr>
<tr>
<td>Fibre cross-sectional area</td>
<td>0.4 mm × 0.4 mm</td>
</tr>
<tr>
<td>Core material</td>
<td>Polystyrene</td>
</tr>
<tr>
<td>Core density</td>
<td>1.06 g/cm²</td>
</tr>
<tr>
<td>Cladding material</td>
<td>PMMA</td>
</tr>
<tr>
<td>Cladding density</td>
<td>1.20 g/cm³</td>
</tr>
<tr>
<td>(EMA material)</td>
<td>Water</td>
</tr>
<tr>
<td>(EMA density)</td>
<td>1.00 g/cm³</td>
</tr>
<tr>
<td>Cladding refractive index, $n_{clad}$</td>
<td>1.49</td>
</tr>
<tr>
<td>Core refractive index, $n_{core}$</td>
<td>1.60</td>
</tr>
<tr>
<td>Core absorption length, $l_{core}$</td>
<td>3.5 m</td>
</tr>
<tr>
<td>(Core scattering length, $\mu_{core}$)</td>
<td>2.0 cm</td>
</tr>
<tr>
<td>Core scintillation yield, $SY$</td>
<td>7,100 photons/MeV</td>
</tr>
<tr>
<td>Optical emission spectrum, $\lambda(E)$</td>
<td>$\lambda(E) = \lambda_{peak} = 580$ nm</td>
</tr>
<tr>
<td>Photodiode refractive index, $n_{diode}(\lambda)$</td>
<td>0.46 – 5.187 a</td>
</tr>
<tr>
<td>Photodiode absorption length, $l_{diode}(\lambda)$</td>
<td>5.29 – 13,300 nm a</td>
</tr>
</tbody>
</table>

*aValues obtained from [31].
7. MC optimisation of a prototype EPID

Table 7.2: Analysis of the sensitivity of the PSA-EPID model to changes in geometric and optical transport parameters.

<table>
<thead>
<tr>
<th>Property</th>
<th>Values tested</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\mu_{\text{core}}$</td>
<td>1.0, 5.0, 10, 50 mm</td>
</tr>
<tr>
<td>EMA $G_4\text{OpticalSurface}$</td>
<td>Present, absent</td>
</tr>
<tr>
<td>Fibre length</td>
<td>15, 50 mm</td>
</tr>
<tr>
<td>Fibre area</td>
<td>0.5 mm $\times$ 0.5 mm</td>
</tr>
</tbody>
</table>

prototype manufacturer (Saint-Gobain Crystals, Hiram OH, USA). Material information for the EMA was not provided and was therefore specified as water-equivalent. This is a valid assumption given that we have previously demonstrated experimentally that the fibres comprising the PSA respond in a water-equivalent manner\cite{17}.

Optical parameters including $\mu$ and $l$ are difficult to measure directly and may not be precisely known due to, for example, variations in the PSA’s manufacturing process. One aspect of this study thus involved varying select optical transport parameters to observe any effects on the calculated quantities described in Section 7.2.3. The values of $n$, which are well known, and $l_{\text{core}}$, which has a nominal value more than two orders of magnitude greater than the fibre dimensions, were not included in this analysis. Information for $\mu_{\text{core}}$ was not provided therefore a reference value of 2 cm was assigned and several alternate values were investigated. The effects of varying geometrical parameters including the fibre length and cross-sectional area were also investigated. Table 7.2 includes the different optical transport and geometrical values that were utilised in addition to the reference values listed in Table 7.1.

### 7.2.3 Simulated imaging and dosimetric quantities

2D histograms scoring the spatial distribution of optical absorption events in the photodiode layer were created using ROOT (version 5.28.00) data-analysis software and were used to evaluate the imaging and dosimetric performance of the
PSA-EPID. This scoring was performed in 0.4 mm \times 0.4 mm bins to match the pixel pitch of the physical detector. Histograms were then analysed and post-processed using \textsc{root} or \textsc{matlab} (version R2011b). For simulations using a monoenergetic x-ray pencil beam, $10^6$ incident x-rays resulted in $\approx 0.4\%$ statistical uncertainty (the fractional uncertainty in the number of optical photons detected in the photodiode per incident x-ray). Those involving open field beams required $5 \times 10^7$ incident x-rays to achieve $\approx 1\%$ statistical uncertainty. Unless otherwise stated, the following quantities were calculated using the reference values (Table 7.1) and non-reference values (Table 7.2).

\subsection*{7.2.3.1 Imaging performance evaluation}

The detection efficiency, $\eta(E_\gamma)$, is the probability for an incident x-ray with energy $E_\gamma$ to generate at least one optical photon in the PSA that is absorbed by the photodiode layer. The detector sensitivity, $\xi(E_\gamma)$, is defined as the mean number of optical photons absorbed by the photodiode layer per incident x-ray as a function of $E_\gamma$. To calculate $\eta(E_\gamma)$ and $\xi(E_\gamma)$, pixel-sized monoenergetic pencil beams with energies from 0.2 to 6 MeV were normally incident on the central fibre. The fraction of incident x-rays that were subsequently detected and the mean number of optical photons absorbed per incident x-ray were scored.

An imaging detector’s modulation transfer function (MTF) characterises its spatial resolution and may be calculated from the 1D LSF\textsuperscript{[32,33]}. One method of calculating a detector’s LSF uses the angled slit technique\textsuperscript{[34]}. This was implemented in the model by sampling x-rays from a 6 MV spectrum derived from the source model. The \textsc{Geant4 General Particle Source} was used to create a 50 mm \times 0.08 mm beam normally incident on the PSA-EPID and angled by $\approx 2.5^\circ$ with respect to the pixel columns. The 2D spatial distribution of optical photons absorbed by the photodiode layer was scored and the MTF was calculated from the modulus of the Fourier transform for the normalised LSF.

\subsection*{7.2.3.2 Dose response evaluation}

The PSA-EPID’s dosimetric response was evaluated by comparing simulated field size factors and relative dose profiles to those simulated in a water phantom. The
7. MC optimisation of a prototype EPID

6 MV photon beam model described in Section 7.2.1 was used with square field side lengths ranging between 3 and 10 cm. The spatial distribution of optical photons absorbed in the photodiode layer was scored in a 2D histogram as this represents the true physical processes leading to image formation.

Field size factors were calculated from the mean response within the central 1 cm × 1 cm for each field size normalised to the response for the 10 cm × 10 cm field. Uncertainties were calculated as the standard deviation in the response within the central 1 cm × 1 cm. 1D relative dose profiles were extracted from the 2D histograms through the PSA-EPID’s central axis in the cross-plane direction, with the response normalised to that simulated at the central axis.

To confirm the PSA-EPID’s water-equivalent response, field size factors and dose profiles were similarly calculated in a water phantom measuring 41 cm × 41 cm and 40 cm depth. Energy deposited at a depth of 30 mm in water (100 cm from the target in the source model) was scored to ensure similar conditions of electronic equilibrium and build up between the PSA and the phantom.

7.3 Results

Aside from $\mu_{core}$, which did not significantly affect the PSA-EPID’s dosimetric response, the parameters investigated uniquely affected each of the calculated quantities.

7.3.1 Detection efficiency

The variations in detection efficiency with fibre length, cross-sectional area, $\mu_{core}$ and EMA $G_4OpticalSurface$ are presented in Figures 7.2(a), (b), (c) and (d) respectively. Subplots show the ratio between efficiencies calculated using non-reference and reference values. Efficiency typically decreases with increasing energy. This is a consequence of the reduced interaction probability and increased range for higher energy x-rays and secondary electrons, respectively, in the plastic scintillator.

The detection efficiency increases with increasing fibre length for all incident x-ray energies, which is consistent with longer fibres having greater sensitive
7.3. Results

volumes. The x-ray energy resulting in peak detection efficiency shifts from 0.2 MeV to 0.3 MeV and finally 0.4 MeV for the 15, 30 and 50 mm long fibres, respectively. This is because low energy x-rays tend to interact at shallower scintillator depths. Increasing fibre length may increase the probability of optical absorption within the scintillator, thus decreasing x-ray detection probability for these energies.

Increasing the fibre area and removing the EMA G4OpticalSurface similarly increased the detection efficiency for x-rays energies < 1 MeV. This is intuitive since increasing the fibre area reduces the probability that optical photons within the fibres will refract through the cladding and be absorbed by the EMA. Similarly, removing the EMA increases the number of optical photons that reach the photodiode layer.

The detection efficiency tends to decrease with decreasing \( \mu_{\text{core}} \), though this effect was most significant for \( \mu_{\text{core}} = 1 \) mm. Increasing \( \mu_{\text{core}} \) increases the average distance traveled by an optical photon between Rayleigh scattering events. Therefore, on average more optical photons reach the photodiode when the probability for optical Rayleigh scattering is reduced. This effect is greatest for the low energy x-rays because the range of \( \mu_{\text{core}} \) values tested is on the order of the fibre length (30 mm). Varying \( \mu_{\text{core}} \) therefore preferentially affects those optical photons generated at shallower depths in the plastic.

7.3.2 Detector sensitivity

Variations in detector sensitivity with fibre length and \( \mu_{\text{core}} \) are shown in Figures 7.3(a) and (b), respectively. Subplots show the ratio between non-reference and reference value detector sensitivities. Sensitivity increases continuously with increasing x-ray energy for all cases investigated. This is likely a consequence of two independent factors. Firstly, higher energy x-rays tend to interact deeper in the PSA and generate optical photons closer to the photodiode. Secondly, higher energy x-rays produce more optical photons on average when they interact and thus create a greater signal when detected.

Increasing the fibre length decreases sensitivity across all x-ray energies and this stems from the greater probability for optical absorption in longer fibres.
7. MC optimisation of a prototype EPID

Figure 7.2: X-ray detection efficiency as a function of energy for varying (a) fibre length, (b) fibre cross-sectional area, (c) $\mu_{\text{core}}$ and (d) EMA optical surface. Subplots show the ratio of the detector efficiency relative to that calculated using reference parameters.
7.3. Results

Figure 7.3: Detector sensitivity as a function of incident x-ray energy for varying (a) fibre length and (b) $\mu_{\text{core}}$. Subplots show the ratio of the detector sensitivity relative to that calculated using reference parameters.

Varying the fibre area and EMA $G4OpticalSurface$ had a negligible effect on the detector sensitivity (data not shown). Interestingly, decreasing values of $\mu_{\text{core}}$ resulted in increased sensitivity to low energy x-rays. One explanation is that as $\mu_{\text{core}}$ approaches the fibre width, fewer optical photons reach the EMA and are absorbed.

7.3.3 Modulation Transfer Function

The variations in the PSA-EPID MTF for different fibre lengths, cross-sectional areas, $\mu_{\text{core}}$ and EMA $G4OpticalSurfaces$ are shown in Figures 7.4(a) – (d). A greater MTF over higher spatial frequencies corresponds to improved spatial resolution.

Increasing fibre length decreases the MTF across all spatial frequencies as longer fibres offer a greater volume for x-ray and secondary electron scatter. Optical photons may thus be generated farther away from the x-ray source, degrading spatial resolution. These results agree with those of other studies reported in the literature\textsuperscript{[23]}. Similarly, increasing the fibre cross-sectional area decreased
the MTF at spatial frequencies above \( \approx 0.2 \text{mm}^{-1} \). By increasing the fibre area optical photons are spatially less confined and, in the case of photodiode pixels being smaller in area than the fibres, may be incident upon multiple pixels despite propagating along a single fibre. The MTF is less sensitive to variations in \( \mu_{\text{core}} \). Across the midrange spatial frequencies, increasing \( \mu_{\text{core}} \) increased the MTF only slightly.

Removing the EMA surrounding each fibre had the most significant impact on the PSA-EPID’s MTF and greatly decreased its spatial resolution across all spatial frequencies. Removing the \( G4\text{OpticalSurface} \) allowed optical photons to interact with the EMA as any other dielectric medium. Therefore, they were able to refract through the EMA into neighbouring fibres, which constituted optical cross talk and thus degraded spatial resolution.

### 7.3.4 Field size response

Figure 7.5 compares field size factors calculated for the PSA-EPID using reference parameters and a water phantom. All field size factors agree within statistical uncertainties with a maximum difference of 1.1\% occurring for the 3 cm \( \times \) 3 cm field size. These results demonstrate the PSA-EPID’s water equivalent response. Variations in optical transport parameters from Table 7.2 did not result in any significant changes to the reference field size factor calculations.

### 7.3.5 Relative dose profiles

Relative dose profiles calculated for the PSA-EPID and water phantom models are compared in Figure 7.6(a). For clarity, the 3 cm \( \times \) 3 cm, 5 cm \( \times \) 5 cm, 7 cm \( \times \) 7 cm and 10 cm \( \times \) 10 cm profiles have been vertically shifted by 0, 10, 20 and 30 arbitrary units, respectively. Comparing the PSA-EPID and water phantom profiles demonstrates very close agreement for all field sizes studied. \( \gamma \)-index analysis\[^{[35]} \] using 3%/3mm criteria yielded \( \geq 91\% \) of data points passing above a dose threshold of 10\% (data shown for 3 cm \( \times \) 3 cm and 10 cm \( \times \) 10 cm field sizes only). Above this threshold, a minimum of 89 data points (3 cm \( \times \) 3 cm field size) were compared for each profile in the \( \gamma \)-index analysis. Note that the sudden
7.3. Results

Figure 7.4: PSA-EPID MTF as a function of spatial frequency for varying (a) fibre length, (b) fibre area, (c) $\mu_{core}$ and (d) EMA optical surface.
Figure 7.5: Field size output factors calculated by scoring optical photons absorbed in the photodiode. Field size factors calculated at 30 mm depth in a water phantom are shown for comparison. The subplot illustrates the ratio of the PSA-EPID field size factors to those calculated in water.
drop in response occurring near ±75 mm corresponds to the outer edges of the scintillator array.

Figure 7.6(b) compares relative dose profiles using 5 cm × 5 cm and 10 cm × 10 cm field sizes for the PSA-EPID with variations in EMA G4OpticalSurface and fibre area. As demonstrated in Figure 7.4(d), removing the EMA degrades spatial resolution and when simulating open field beams, this manifests as severe profile rounding.

An interesting effect was observed for PSA-EPID profiles calculated with a fibre cross-sectional area of 0.5 mm × 0.5 mm. Increasing the fibre area eliminated the precise matching of individual fibres and photodiode pixels so that optical photons confined to a single fibre were incident upon multiple pixels. This resulted in quasi-periodic variations in the dose profiles most noticeable in the in-field region. A γ-index analysis using 3%/3mm criteria yielded only 73% and 41% of data points from the 5 cm × 5 cm and 10 cm × 10 cm profiles, respectively, with 0.5 mm × 0.5 mm fibre area agreeing to the reference profile above a 10% relative dose threshold.

7.4 Discussion

The results obtained provide valuable insight into factors affecting the PSA-EPID response and how it may be optimised. Simulating the detection efficiency, sensitivity and MTF quantified the detector’s imaging performance while field size factors and relative dose profiles characterised its dose response relative to water.

The PSA-EPID’s detection efficiency was most sensitive to fibre length, with an increase from 30 to 50 mm (∼ 2/3) producing an increase in peak efficiency of ∼ 43%. However, increasing fibre length also caused an approximately constant decrease in sensitivity of 30%, which may act to reduce the contrast of images acquired using a thicker scintillator. The continuously increasing sensitivity across 0.2 – 6 MeV for all fibre lengths agrees with results published by Teymurazyan et al.. Because of its water-equivalent response, the PSA-EPID does not exhibit the over-response to low energy radiation characteristic of standard EPIDs[23]. Simulation results published by Cremers et al. for a copper plate/phosphor screen instead showed a peak in sensitivity for x-ray energies less than ∼ 2 MeV owing
Figure 7.6: (a) Relative dose profiles calculated using the PSA-EPID model with reference parameters are compared to dose profiles calculated in a water phantom. The 3 cm × 3 cm, 5 cm × 5 cm, 7 cm × 7 cm and 10 cm × 10 cm profiles have been shifted vertically by 0, 10, 20 and 30 respectively for clarity. The subplot shows a γ-index analysis (3%/3mm). (b) Relative dose profiles calculated using reference and non-reference EMA and fibre cross-sectional area for the 5 cm × 5 cm and 10 cm × 10 cm field sizes. The 10 cm × 10 cm profiles have been shifted vertically by 20.
7.4. Discussion

to the large cross-section of the high-Z components$^{[36]}$. One potential method
to further increase efficiency is by using an optical coupling agent to increase
transmission between from the PSA fibres to the photodiodes. This will be the
subject of future investigation.

Simulations of the detector MTF quantified the loss in spatial resolution that
occurs for scintillators of increasing thickness. Moreover, these simulations have
demonstrated the importance of an effective EMA surrounding each fibre to main-
tain spatial resolution and the impact of element-to-element mismatch between
fibres and the underlying photodiode pixels. This latter result is an important
demonstration of the improvements in spatial resolution that may be realised by
optimising the physical dimensions of our prototype and matching them to the
photodiode pixel pitch. The variation in MTF for differing $\mu_{\text{core}}$ is also important.
Since information concerning this parameter is not precisely known, $\mu_{\text{core}}$ may be
considered a “free” model parameter that may be determined empirically by com-
paring simulated MTFs to experimental measurements. This process, which may
aid in validating the model, is currently under investigation.

Differences in the 6 MV energy spectra occurring for photon beams with
different field sizes causes variation in field size and off-axis dose profile response.
The increases in these quantities relative to water observed with standard EPIDs
have been greatly reduced with the PSA-EPID, further demonstrating its water-
equivalent response.

Dose profile results suggest that a quasi-periodic variation in response may be
observed experimentally using the PSA-EPID prototype. Others have reported
similar misalignment issues while investigating thick, segmented crystal scintilla-
tors$^{[14,15]}$. El-Mohri et al. proposed a novel binning technique to minimise the
undesirable effects of detector misalignment, however it was most effective when
applied to scintillators exhibiting mechanical hardness, high density and high re-
fractive index$^{[16]}$. Investigating the misalignment effect and correction methods
experimentally is the subject of future investigation.
7. MC optimisation of a prototype EPID

7.5 Conclusions

This study reports on a Monte Carlo investigation of the imaging and dosimetry performance of a novel \(\alpha\)-Si EPID employing an array of plastic scintillating fibres (PSA) in place of the standard EPID metal plate and phosphor screen. We have found that the increase in the PSA-EPID’s detection efficiency with fibre length is partially offset by a reduction in sensitivity to x-rays, which decreases with increasing fibre length. The energy at which x-ray detection efficiency peaks was found to increase with fibre length, from \(\approx 0.2\) MeV for 15 mm length to \(\approx 0.4\) MeV for 50 mm length. Our results suggest that self-absorption of optical photons in the fibres becomes important for lengths approaching 50 mm and therefore that the optimum fibre length is less than 50 mm. The MTF is acutely sensitive to the presence of extramural absorber (EMA), which prevents optical crosstalk between the fibres. The trade-off between improving detection efficiency while decreasing spatial resolution when using thicker scintillators is well known. Based on these results, however, the use of an absorbing EMA and scintillator core with increased Rayleigh scattering length are recommended to maximise both detection efficiency and spatial resolution. The optical transport parameters studied did not significantly influence the PSA-EPID’s water-equivalent dose response, which is important so that we may continue to optimise the PSA-EPID’s imaging performance without affecting its water-equivalency. However, the alignment of the PSA fibre cross-sectional area with the photodiode pixel size affects the profile shape within the open field region. This is an important consideration for experimental prototype development as performance is optimised when the fibres are precisely aligned to the underlying imaging pixels. Future work will involve validating our model against experimental measurements so that it may be used to optimise the design of a next-generation EPID capable of simultaneous imaging and dose verification in radiotherapy.

Acknowledgments

The authors would like to acknowledge funding support from the Cancer Institute NSW (Research Equipment Grants 10/REG/1-10 and 10/REG/1-20) and
Cancer Council NSW (Grant ID RG 11-06). S.J.B. would also like to thank The University of Sydney and the Institute of Medical Physics for scholarship support, as well as the Liverpool and Macarthur Cancer Therapy Centers for additional financial support. The authors report no conflicts of interest in conducting this research.
References


References


References


A next-generation EPID for simultaneous imaging and dosimetry in radiotherapy

The work presented in this chapter includes the experimental demonstration and model validation of the second-generation prototype EPID employing plastic scintillating fibres.
8. A next-generation EPID: measurements & model validation

Statement of joint authorship

This work is in preparation for submission to *Radiotherapy & Oncology* as:


<table>
<thead>
<tr>
<th>Author</th>
<th>Specific involvements</th>
</tr>
</thead>
<tbody>
<tr>
<td>S.J. Blake</td>
<td>Developed, built and ran simulation code, aided experimental design, performed experiments, analysed results, wrote manuscript (80%)</td>
</tr>
<tr>
<td>A.L. McNamara</td>
<td>Performed experiments, edited manuscript (3%)</td>
</tr>
<tr>
<td>P. Vial</td>
<td>Experimental design, performed experiments, edited manuscript (10%)</td>
</tr>
<tr>
<td>L. Holloway</td>
<td>Aided experimental design, edited manuscript (2%)</td>
</tr>
<tr>
<td>Z. Kuncic</td>
<td>Intellectual contribution toward Monte Carlo modelling, aided experimental design, edited manuscript (5%)</td>
</tr>
</tbody>
</table>
Abstract

Background and Purpose: This study reports the first experimental measurements of a novel, second-generation prototype water-equivalent electronic portal imaging device (EPID) designed for simultaneous imaging and dose verification in radiotherapy. Measurements were performed both to characterise the detector’s performance and empirically validate a Monte Carlo (MC) of the prototype that will be used for ongoing detector optimisation.

Materials and Methods: The prototype EPID utilises an array of plastic scintillating fibres in place of the metal plate/phosphor screen used in standard amorphous silicon EPIDs and the MC model, developed using GEANT4, is based on its design. Experiments were performed using a clinical 6 MV photon beam to measure the prototype’s modulation transfer function (MTF), field size output factors and relative dose profiles for static, open fields. These quantities were likewise simulated using the MC model and comparison with measured results empirically validated unknown model parameters.

Results: Field size factor and profile measurements demonstrated the prototype EPID’s water-equivalent response. Dose profiles exhibited quasi-periodic variations resulting from element-to-element mismatch between the scintillating fibres and underlying photodiode pixels. The presence of a thin air gap between the scintillator and photodiode plane caused a more rapid fall off of the MTF relative to an idealised configuration without an air gap.

Conclusions: A novel water-equivalent EPID has been characterised experimentally for imaging and dosimetry applications in radiotherapy. A MC model has also been validated and simulation results suggest that further improvement in spatial resolution may be realised through ongoing detector optimisation.
8. A next-generation EPID: measurements & model validation

8.1 Introduction

Improvements in our ability to modulate the shape and intensity of high-energy x-ray beams in modern radiotherapy enable clinicians to offer highly conformal and patient-specific therapies. However, the increasing prevalence of steep dose gradients in modern therapies places a high level of importance on ensuring correct patient positioning and monitoring of intra-fraction motion\[1\]. The ability to perform routine \textit{in vivo} patient dosimetry would give clinicians a means to verify that treatments are delivered as intended and would serve as a useful tool to catch errors in dose delivery and identify those patients that may benefit from adaptations to their treatment plan. This study proposes a novel detector based on modern amorphous silicon (\textit{a}-Si) electronic portal imaging devices (EPIDs) that may be used for simultaneous imaging and dose verification in radiotherapy.

Megavoltage (MV) \textit{a}-Si EPIDs are based on active matrix flat panel imaging (AMFPI) technology and first became commercially available in the year 2000\[2,3\]. Since their commercial inception, \textit{a}-Si EPIDs have evolved to become one of the most frequently used imagers in radiotherapy clinics, finding applications in image-guided radiotherapy\[4,5\], linear accelerator (linac) quality assurance (QA)\[6,7\] and \textit{in vivo} patient dosimetry\[8,9\]. There are also several features that make \textit{a}-Si EPIDs ideally suited for routine \textit{in vivo} dosimetry; they are readily supplied by linac vendors, are typically mounted to the gantry directly opposite the primary MV therapy beam, offer real-time readout capabilities, respond linearly to integral dose and independently of dose rate and are highly resistant to radiation-induced damage\[10–16\]. The primary drawback to using modern \textit{a}-Si EPIDs for dosimetry applications is their well-characterised non water-equivalent response, which complicates the calibration of these detectors against reference water-equivalent dosimeters\[12,14,17–19\]. Their non water-equivalent response stems primarily from the high atomic number ($Z$) metal plate and phosphor screen components, which causes them to over-respond to low energy radiation relative to water.

An x-ray imager’s detective quantum efficiency (DQE) is a measure of how well it transfers an input x-ray signal into an output image. A theoretically perfect imager with a DQE of 100% would process an input signal without degradation
such that the resulting image’s quality is dependent solely on characteristics of the input signal\cite{3}. Several studies have shown that modern $a$-Si EPIDs have DQEs on the order of 1-3\%\cite{3,20,21}, therefore significant effort has gone into developing novel EPIDs with improved DQE. Several such high-DQE EPID prototypes have been reported in the recent literature, most of which are based on the principle of replacing current metal plate/phosphor screen components with thick, segmented scintillators\cite{22–29}. These designs drastically increase the detector sensitivity relative to standard EPIDs that employ thin phosphor screens while striving to maintain high spatial resolution. The segmented scintillator design, however, introduces an imaging artifact that stems from an inevitable misalignment between the individual segments of the scintillators and the imaging pixels of the underlying detector\cite{25,30,31}. Depending on the specific geometry under consideration and the pixel pitch of the detector being used, this element-to-element mismatch can significantly degrade image quality. At present, one group has presented novel post-processing binning algorithms to minimise the loss in image quality resulting from this geometrical mismatch\cite{31}. Another disadvantage to the thick, segmented scintillators proposed by these groups is that they do not improve upon the current non water-equivalent response of standard EPIDs since they too employ high-$Z$ materials.

Plastic scintillators have been well characterised with respect to their water-equivalent dosimetric response and have frequently been used in clinical dosimetry applications. Several different prototype dosimeters employing plastic scintillator have been reported in the literature\cite{32–40}. Teymurazyan and Pang (2012) proposed a novel water-equivalent EPID employing plastic scintillating fibres and used Monte Carlo (MC) simulations to characterise both its water-equivalent response and its imaging performance in terms of the modulation transfer function (MTF) and zero spatial frequency DQE\cite{41}. Based on their model, they predicted the water-equivalent EPID to have a higher DQE and better energy response than current copper plate/phosphor screen EPIDs. The EPID’s water-equivalent response eliminates the over-sensitivity to low energy x-rays. However, their model predicted the spatial resolution to be worse than that of the standard EPID and they commented that the increased thickness of the plastic scintillator would make such proposed detectors heavier and bulkier than standard EPIDs.
The previous characterisation of the imaging and dosimetry performance of our first-generation prototype EPID employing plastic scintillation fibres represents the first experimental validation of such a thick, segmented plastic scintillator in the literature\textsuperscript{[42]}. The primary advantage to the prototype EPID design over that of standard $a$-Si EPIDs and those proposed for high-DQE imaging is its demonstrated water-equivalent response. The plastic scintillating fibres of the second-generation prototype array reported in this study were fabricated using the same materials as those in the first-generation prototype. Therefore, it is hypothesized that this second-generation prototype will continue to exhibit a water-equivalent response.

The principal aim of this study was to characterise the imaging and dosimetry performance of a second-generation prototype of a novel EPID employing an array of plastic scintillating fibres (PSA). Experiments were performed to evaluate the water-equivalent response of the prototype EPID and to quantify the detector’s line spread function (LSF) and MTF. These measurements were also used to validate a previously developed MC model of the prototype EPID\textsuperscript{[13]}. By validating a MC model of the PSA-EPID against our physical prototype, we will be able to further quantify the detector response and investigate potential configurations to further optimise its overall performance in simultaneous imaging and dosimetry.

## 8.2 Methods and Materials

### 8.2.1 Description of the prototype detector

The scintillation detector investigated in this study is a second-generation prototype that is based on an earlier design previously characterised by our group\textsuperscript{[42]}. It comprises an array of plastic scintillating fibres constructed by and purchased from Saint-Gobain Crystals (Saint-Gobain Crystals, Hiram OH, USA) that is used with a standard configuration $a$-Si EPID purchased specifically for research purposes. The research $a$-Si EPID was a PerkinElmer XRD 1640 AN CS flat panel imaging device (PerkinElmer, Santa Clara, CA) that was previously used to characterise our group’s first PSA prototype\textsuperscript{[42]}. It employs a 1024 $\times$ 1024
array of $a$-Si photodiode pixels each having a pitch of 0.4 mm and total combined area measuring $\approx 41 \times 41 \text{ cm}^2$. As a research detector, the front aluminium cover of the $a$-Si EPID may be taken off and the underlying copper sheet, optical filter and gadolinium oxysulfide phosphor screen removed to expose the array of photodiode pixels.

A photo of the PSA along with a schematic illustrating its structure and dimensions are shown in Figure 8.1. The array comprises square plastic scintillating fibres measuring $0.5 \times 0.5 \text{ mm}^2$ in cross-sectional area and 30 mm in length. Each fibre comprising the array is identical and consists of a polystyrene scintillating core doped with organic fluors surrounded by a PMMA cladding. A thin extramural absorber (EMA) is painted on the outer surface of the cladding and acts to absorb optical photons refracting through the cladding to prevent optical cross talk between fibres. The array contains $300 \times 300$ parallel fibres, which results in a total area of approximately $150 \times 150 \text{ mm}^2$. A rigid, plastic frame surrounds the array and offers structural support. For comparison, the first generation prototype previously characterised by our group contained fibres with an identical chemical and structural makeup, however measuring $1 \times 1 \text{ mm}^2$ in cross-sectional area and with a length of 15 mm.

Figure 8.1: (a) Photo of the second-generation prototype array of plastic scintillating fibres. The individual fibre components and dimensions are illustrated in the schematic, (b).
8. A next-generation EPID: measurements & model validation

As described further in Section 8.2.3, experiments were performed with the aluminium cover, copper sheet, optical filter and phosphor screen of the standard configuration a-Si EPID removed and replaced by the PSA. The PSA was placed in direct contact with the photodiode panel such that the combined PSA-EPID constituted an indirect x-ray detection configuration. X-rays and secondary electrons interacting within the scintillator produce optical photons that are channeled along individual fibres via total internal reflection. Those optical photons reaching the photodiodes may generate electron-hole pairs within the a-Si, leading to a build up of charge that may be integrated and subsequently read out to form a digital image.

8.2.2 Overview of the Monte Carlo model

A MC model of the PSA-EPID was developed using GEANT4 and is shown in Figure 8.2. While a detailed description and characterisation of this model has been described in another paper\textsuperscript{[43]}, an overview is presented here for completeness. GEANT4 was chosen for the modeling component of this study because of its ability to self-consistently simulate both x-ray and optical photon transport relevant for such indirect-detection imagers. Several others have similarly used GEANT4 to study optical transport within phosphor screens\textsuperscript{[44,45]}, crystal scintillators\textsuperscript{[46,47]} and plastic scintillators\textsuperscript{[41]}.

A previously validated model of a standard a-Si EPID was modified by replacing all components upstream of the photodiode plane with an array of plastic scintillating fibres\textsuperscript{[42]}. The PSA model was based on the physical prototype described in Section 8.2.1 with material properties for the fibres and cladding provided by the manufacturer (Saint-Gobain Crystals, Hiram OH, USA). The square fibres were modeled as previously described with a PMMA cladding thickness of 20 µm and EMA thickness of 15 µm. Geometrical and material properties for the a-Si photodiode array and all underlying components that comprised the research EPID were also provided by the manufacturer (PerkinElmer, Santa Clara, CA). The a-Si photodiodes were modeled as a uniform 1 mm thick layer of a-Si with a 1 mm SiO\textsubscript{2} substrate.

The standard electromagnetic GEANT4 physics classes were used to simulate
8.2. Methods and Materials

Figure 8.2: GEANT4 model of the plastic scintillating fibre array and underlying a-Si EPID (not to scale). For clarity, only the photodiode array and substrate are shown.

photoelectric absorption, Compton scattering, pair production, bremsstrahlung radiation, impact ionisation, electron/positron annihilation, scintillation and Čerenkov radiation. Optical physics classes were used to simulate optical boundary processes (reflection and refraction), incoherent (Rayleigh) scattering and bulk absorption. To properly simulate optical transport, the correct optical properties must be assigned to the relevant materials. These include the refractive indices $n$, Rayleigh scattering lengths $\mu$ and bulk absorption lengths $l$ for all materials that optical photons propagate through. In addition, the scintillation yield $SY$ with optical emission spectrum $\lambda(E)$ (for photon energy $E$ with peak wavelength $\lambda_{\text{peak}}$) must be specified for the scintillator. Unless otherwise stated, all surfaces were treated as specular optical photon reflectors. Where possible, these optical properties were obtained from the manufacturer otherwise values were obtained from the literature. A summary of all relevant material and optical transport parameters and the values used throughout this study is listed in Table 8.1.

Material information was not provided for the EMA layer, therefore its material was simply assigned as water. Its function to absorb optical photons that
8. A next-generation EPID: measurements & model validation

Table 8.1: Summary of the reference properties of the PSA-EPID fibres and photodiodes used in the MC model. Unless otherwise stated, values were provided by the manufacturer. Properties listed in parentheses had no information provided.

<table>
<thead>
<tr>
<th>Property</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fibre length</td>
<td>30 mm</td>
</tr>
<tr>
<td>Fibre cross-sectional area</td>
<td>$0.4 \times 0.4 \text{mm}^2$</td>
</tr>
<tr>
<td>Core material</td>
<td>Polystyrene</td>
</tr>
<tr>
<td>Core density</td>
<td>1.06 g/cm$^2$</td>
</tr>
<tr>
<td>Cladding material</td>
<td>PMMA</td>
</tr>
<tr>
<td>Cladding density</td>
<td>1.20 g/cm$^3$</td>
</tr>
<tr>
<td>(EMA material)</td>
<td>Water</td>
</tr>
<tr>
<td>(EMA density)</td>
<td>1.00 g/cm$^3$</td>
</tr>
<tr>
<td>Core refractive index, $n_{\text{core}}$</td>
<td>1.49</td>
</tr>
<tr>
<td>Core absorption length, $l_{\text{core}}$</td>
<td>3.5 m</td>
</tr>
<tr>
<td>(Core scattering length, $\mu_{\text{core}}$)</td>
<td>2.0 cm</td>
</tr>
<tr>
<td>Core scintillation yield, $SY$</td>
<td>7,100 photons/MeV</td>
</tr>
<tr>
<td>Optical emission spectrum, $\lambda(E)$</td>
<td>$\lambda(E) = \lambda_{\text{peak}} = 580 \text{nm}$</td>
</tr>
<tr>
<td>Scintillation time decay constant, $t$</td>
<td>16.8 ms</td>
</tr>
<tr>
<td>Photodiode refractive index, $n_{\text{diode}}(\lambda)$</td>
<td>$0.46 - 5.187 , \alpha$</td>
</tr>
<tr>
<td>Photodiode absorption length, $l_{\text{diode}}(\lambda)$</td>
<td>$5.29 - 13,300 , \text{nm} , \alpha$</td>
</tr>
</tbody>
</table>

*Values obtained from\cite{48}.*
are transmitted through the cladding layer and thereby prevent optical cross

talk between fibres was treated by assigning a $G4OpticalSurface$ to the interface

between the cladding and EMA layers. GEANT4 allows users to specify such sur-

faces to have more control over the types of optical boundary processes taking

place. This $G4OpticalSurface$ was therefore defined to absorb all incident optical

photons. The impact of using this $G4OpticalSurface$ on the resulting detector ef-

ficiency, sensitivity and modulation transfer function was previously investigated

in a separate study\cite{43}. Briefly, the $G4OpticalSurface$ was found to increase the
detector efficiency for x-ray energies below $\approx 1$ MeV and was shown to be crucial
in maintaining high spatial resolution.

8.2.3 Experimental measurements

Measurements were performed using the PSA-EPID to both characterise its re-
sponse and quantify its performance for imaging and dosimetry. These measure-
ments also provided a series of data against which to validate the MC model. The
following subsections describe the experimental procedures followed to measure

the detector LSF, MTF, field size output factors and relative dose profiles. The

LSF and MTF are commonly used metrics for quantifying the spatial resolution

of imaging systems whereas the field size response and relative dose profiles are

used to investigate the variations in the detector’s response with incident x-ray

energy spectra.

A 6 MV Elekta Synergy linac (Elekta, Crawley, UK) was used to generate

photon beams for all measurements. Unless otherwise stated, all beam field sizes

and positions are defined with respect to the isocentre, located at a distance of 100

cm from the target. The PerkinElmer software package XIS (PerkinElmer, Santa

Clara, CA) was interfaced with the research EPID to facilitate image acquisition.

To minimise the fluctuations in residual dark current, the EPID was left to

warm up for approximately 20 minutes prior to each set of measurements. Dark

field images were acquired by integrating the EPID signal for 30 frames with

the radiation source off and were updated after each portal image was acquired.

When necessary, flood field images were acquired as frame-averaged exposures

of a region on the EPID using an open field larger than the measurement area
of interest. For a given measurement, dark field images were subtracted from the raw image and the result was divided by the flood field to correct for non-uniformities in pixel response. Dead pixel corrections were also applied to all corrected images.

8.2.3.1 Line spread and modulation transfer functions

The angled-slit technique was used to measure the PSA-EPID’s LSF\(^{[49-52]}\). This method has previously been used by others to measure the LSF of MV imagers including standard EPIDs\(^{[20,51]}\) and novel EPIDs employing thick, segmented phosphor and crystal scintillators\(^ {\cite{23,52}}\). Measuring the LSF involves the generation of a photon beam incident on a closely spaced pair of thick, tungsten blocks to form a narrow beam of radiation with sub-pixel width. Since the MTF may be calculated by taking the Fourier transform of the LSF, the following procedure describes the experimental determination of the LSF.

To measure the detector LSF, it was necessary to rotate the linac gantry and the treatment couch to 90° and 270°, respectively. The treatment couch was retracted and a portable cart supporting two machine-polished tungsten blocks on a translation stage was positioned near the isocentre. Each block measured 17.5 × 11.5 × 8 cm\(^3\) (thickness × height × width) with the largest dimension oriented parallel to the linac beam central axis. A piece of paper was placed between the blocks and a clamp was used to press the blocks together, forming a slit approximately 0.08 mm wide. The width of the slit was verified with a series of metal shims with known thicknesses. The translation stage was used to rotate and translate the blocks such that the linac beam’s central axis was centred on and oriented parallel to the slit opening, using the in-room lasers for guidance. The translation stage itself was slightly elevated on one side to allow the slit to be slightly angled (≈ 4° with respect to vertical) to obtain a sub-pixel sampled LSF. The research EPID, with all components above the a-Si photodiodes removed (including the aluminium cover, copper sheet, opaque optical filter and phosphor screen), was then placed on its side facing the linac target and in line with the blocks and the beam central axis. Relative to the linac target, the proximal face of the blocks were positioned at a distance of 110 cm and the EPID was positioned
8.2. Methods and Materials

at 143 cm. A clamp was used to secure the EPID to the couch in this orientation.

Finally, the PSA was placed on its side on top of a Styrofoam support and was positioned against the photodiodes. Individual fibre axes were therefore perpendicular to the surface of the photodiode plane and parallel to the beam central axis. An attempt was made to ensure good contact between the scintillator and photodiode surfaces by fixing tape from the edges of the PSA to the outer frame of the research EPID. While alternative methods may have resulted in improved contact between the PSA and the photodiodes (such as the use of an optical coupling agent), these may have also had an unintended and permanent effects on the detector and hence were not investigated. Finally, all light sources within the linac bunker were turned off and the entire PSA-EPID apparatus was covered with an opaque sheet to prevent any residual ambient light from affecting the detector’s signal. A schematic illustrating this experimental setup is shown in Figure 8.3.

Figure 8.3: Diagram of the experimental setup used to measure the PSA-EPID LSF using the angled slit technique. (a) side-on view and (b) beam’s-eye view.

To calculate the detector LSF a set of three different images was acquired. Image acquisition was repeated several times to ensure measurement consistency. First, the raw LSF images were obtained by irradiating the tungsten blocks with
a vertically oriented $5 \times 2 \text{cm}^2$ open x-ray field, centred on the narrow spacing between the blocks. The narrow beam of radiation that passed between the blocks and was incident on the PSA-EPID constituted the angled slit that generated the raw LSF image. A second image was then acquired after slightly rotating the translation stage that supported the tungsten blocks in the plane of the floor so that the angled slit was no longer in line with the beam central axis. The same $5 \times 2 \text{cm}^2$ field was used to irradiate the blocks, however with the slit out of alignment no primary signal was measured with the PSA-EPID. These “background” images were used to subtract any background signal from the raw LSF images that may have resulted from x-ray scatter and transmission. Finally, the blocks were completely removed from the linac beam and a flood field correction image was acquired by irradiating the PSA-EPID geometry with a larger, $7 \times 7 \text{cm}^2$ open field. The raw and background images were independently dark field and flood field corrected before the background image was subtracted from the raw image. Finally, a dead pixel correction was applied.

All image processing was performed using MATLAB (version R2011b) and in-house code was written to calculate the detector’s LSF. The slit angle $\theta_{\text{slit}}$ relative to the vertical columns of image pixels was first determined by sampling pixels with peak responses from the corrected image. Each horizontal row in the image was then shifted by a distance $y_i \tan \theta_{\text{slit}}$ where $y_i$ is the vertical distance between row $i$ and the row passing through the centre of the slit. A 1D sub-sampled LSF was finally calculated by summing along the resulting column vectors. The MTF was then calculated from the modulus of the Fourier transform for the normalized LSF by using MATLAB’s built-in Fast Fourier Transform (FFT) algorithm.

8.2.3.2 Field size output factors

Field size output factors were measured both for the PSA-EPID and a MatriXX 2D array of ionisation chambers (IBA Dosimetry Asia Pacific, Beijing, China) by placing the detectors on the treatment couch and rotating the linac to a gantry angle of $0^\circ$. The detectors were centred on the beam central axis and the source to detector distance (SDD) was fixed at 100 cm. The detector plane was considered to be at the position of the photodiodes for the PSA-EPID and at the indicated
8.2. Methods and Materials

The PSA-EPID field size response was measured by using the PSA-EPID to acquire a series of static, open field images for varying field size. Images for square fields ranging from $2 \times 2$ to $15 \times 15 \text{cm}^2$ were acquired and the mean response within the central $1 \times 1 \text{cm}^2$ of the open field region was calculated. Field size output factors were then determined by normalising the response for each field to that measured with the reference $10 \times 10 \text{cm}^2$ field size and measurement uncertainties were taken as the standard deviation in the response within the central region of interest.

To evaluate the water-equivalency of the PSA-EPID’s response, field size output factors were also measured using a MatriXX array of ion chambers with $27 \text{mm}$ solid water buildup – a configuration that has previously been shown to be water-equivalent$^{[42]}$. The MatriXX was slightly offset to align a single ionisation chamber with the beam central axis and a series of three measurements were taken by delivering 100 monitor units (MU) for each field size. The mean response and standard deviation were then calculated for each field size and field size output factors were determined by normalising measurements to that measured with the $10 \times 10 \text{cm}^2$ field size.

8.2.3.3 Relative dose profiles

A set of open field images for the $5 \times 5$ and $10 \times 10 \text{cm}^2$ field sizes were also acquired using the PSA-EPID to calculate relative dose profiles. Profiles were extracted from the 2D images along a 1D slice in the cross plane direction through the centre of the open field. Each profile was then normalised to the response measured at the central axis. The profiles were measured to investigate any changes in PSA-EPID response at distances away from the central axis and to investigate the profile penumbra shape measured for several field sizes using a segmented detector.

An artifact arises in the dose profiles measured with the PSA-EPID owing to a mismatch between individual scintillating fibres and the underlying photodiode pixels. This mismatch has been previously reported for other prototype EPIDs using segmented scintillators$^{[25,30,31]}$ and is largely due to the $0.5 \times 0.5 \text{mm}^2$ cross-
sectional area of the scintillating fibres being greater than the 0.4 mm \(a\)-Si pixel pitch. In an idealised geometry, each fibre would be precisely matched to a single \(a\)-Si to maximise the detector’s spatial resolution. An initial attempt to remove this artifact from the dose profiles was taken by applying flood field corrections to the PSA-EPID images. Others have reported alternative methods to minimise the effect of the mismatching artifact, however the application of these methods was beyond the scope of the current study\textsuperscript{[31]}.

### 8.2.4 Simulated quantities

The MC model was used to simulate the same quantities described throughout Section 8.2.3 and measured experimentally using the physical PSA-EPID prototype. A previously validated source model of a 6 MV Elekta Synergy (Elekta, Crawley, UK) linac was used for all simulations\textsuperscript{[42]}. The spatial distribution of optical photons that reached and were absorbed by the photodiodes was scored in 2D histograms using ROOT (version 5.28.00) data-analysis software to quantify the PSA-EPID response. The histogram bin size was 0.4 mm to match the spatial resolution of the physical \(a\)-Si photodiode array. These histograms were then analysed and post-processed using in-house code developed in ROOT and MATLAB (version R2011b).

#### 8.2.4.1 Line spread and modulation transfer functions

The LSF and MTF were simulated using the GEANT4 General Particle Source class to define a geometrical x-ray source with the same physical dimensions as the experimental angled slit \((50 \times 0.08 \text{ mm}^2 \text{ at an angle of } \approx 2.5^\circ)\). X-ray energies were sampled from a 6 MV clinical spectrum and were normally incident on the PSA-EPID’s surface. The simulation of \(10^7\) primary histories was sufficient to achieve a statistical uncertainty less than 1% in the number of optical photons absorbed per incident x-ray at the beam central axis.

The 2D histogram scoring optical photon absorption events in the photodiode was analysed using the same process described in Section 8.2.3.1 for the experimental LSF images. Once the sub-sampled 1D LSF was calculated, a Fourier transform was used to generate the simulated MTF.
Owing to the difficulty of ensuring good optical contact between the PSA and photodiodes when performing the LSF measurement (as described in Section 8.2.3.1), the detector LSF was also simulated with a thin, uniform air gap present between the PSA and the photodiode layer in the MC model. The air gap was specified to have an arbitrary uniform thickness of 0.5 mm as this was believed to be a good estimate of the maximum possible spacing that may have been present in the experimental setup. The LSF and MTF that were simulated with the 0.5 mm air gap were then compared to those simulated without any air gap present to quantify the gap’s impact on these imaging metrics.

### 8.2.4.2 Field size output factors

Field size output factors were simulated using static, open fields measuring $3 \times 3$, $5 \times 5$, $7 \times 7$ and $10 \times 10 \text{cm}^2$. Clinical open field x-ray beams were generated using EGSnrc and BEAMnrc and saved as output phase space files. These files were then read into the GEANT4 model of the PSA-EPID.

In a manner similar to that performed experimentally, the mean number of optical photons absorbed within the central $1 \times 1 \text{cm}^2$ region of the 2D histogram was calculated for each field size. Field size output factors were then determined by normalising these values to the response simulated with the $10 \times 10 \text{cm}^2$ reference field size. Uncertainties were calculated as the standard deviation in the response within the central region for each field size.

### 8.2.4.3 Relative dose profiles

Relative dose profiles were simulated using the same static, open $5 \times 5$ and $10 \times 10 \text{cm}^2$ field size phase space files previously described for the field size output factor calculations. The simulation of $5 \times 10^7$ primary histories was sufficient to achieve a statistical uncertainty of $\approx 1\%$ in the number of optical photons absorbed per incident x-ray at the beam central axis. Profiles were calculated by extracting 1D slices from the 2D histograms through the central axis of the open field in the cross-plane direction. Profiles were normalised to the response calculated at the central axis. To simulate the effect of the geometrical mismatch between the plastic scintillating fibres and the underlying photodiode pixels, dose
profiles were also simulated in an idealised PSA geometry using a $0.4 \times 0.4 \text{mm}^2$ fibre cross-sectional area. Throughout the remainder of this paper, profiles simulated using the true $0.5 \times 0.5 \text{mm}^2$ fibre area and the idealised $0.4 \times 0.4 \text{mm}^2$ area will be referred to as the “uncorrected” and “corrected” simulation profiles, respectively.

8.3 Results and discussion

8.3.1 Line spread and modulation transfer functions

The experimentally measured and simulated LSFs, normalised to the peak response at the detector central axis, are illustrated in Figure 8.4. Due to the experimentally difficult nature of measuring the detector LSF in the MV energy range, there is significant noise in the measured response at distances more than approximately 5–10 mm away from the central axis. A median filter was used to help improve upon this noise without affecting the overall shape of the experimental LSF. The experimentally measured and simulated MTFs are illustrated in Figure 8.5 and were calculated by taking the Fourier transform of the LSFs shown in Figure 8.4.

The simulation LSF and MTF were calculated using optical transport parameters that were obtained from the manufacturer or the literature where possible. Information concerning the Rayleigh scattering length in plastic scintillator and the optical nature of the fibre surfaces were not provided; therefore a range of potential values (between $1 - 50 \text{mm}$, see also Chapter 7) was investigated to empirically validate these parameters against the measured data. It was found that a Rayleigh scattering length on the order of $\approx 1 \text{mm}$ resulted in optimal agreement with measurements when specifying the EMA to act as a pure optical absorber with zero reflectivity.

Interestingly, it was found that the introduction of a 0.5 mm thick layer of air inserted at the contact surface between the PSA and the photodiode plane drastically improved agreement with the measured results. In an idealised geometry without the air gap present, the simulated LSF was narrower and, consequently, the MTF was increased across all spatial frequencies. The value of $f_{50}$ (the spatial
8.3. Results and discussion

Figure 8.4: Comparison of the experimentally measured LSF and simulated LSFs calculated with and without a 0.5 mm air gap between the scintillator and photodiode plane. LSFs were normalised to their peak values at the detector central axis. For clarity, only every third data point has been plotted for each curve.

frequency at which the normalised MTF assumes a value of 0.5) for the experimental MTF was 0.12 mm\(^{-1}\). Values for the simulation MTFs with and without the 0.5 mm air gap present were and 0.13 and 0.18 mm\(^{-1}\) respectively.

While efforts were made to reduce the presence of air between the scintillator and photodiode during measurements, it is certainly reasonable to expect a small air gap to be present. Only a minimal amount of pressure was applied to the scintillator to maintain close contact with the photodiodes so as not to damage either component of the detector. This data has also been shown in Figures 8.4 and 8.5 to demonstrate the potential improvements in spatial resolution that may be realised by further optimising the combined PSA-EPID detector design. There are, however, several additional factors that may also contribute towards a decrease in the measured MTF at higher spatial frequencies, relative to that simulated without an air gap. One potential factor is whether the EMA itself is truly 100% absorptive. Light leakage through the EMA would cause optical cross-talk, which would also contribute towards a decreased MTF. The imaging
8. A next-generation EPID: measurements & model validation

Panel itself may also have a thin protective coating that could behave like an air gap and allow optical spread between the scintillator and photodiodes. Such factors are worthy of additional investigation.

Another aspect of the PSA’s design that is worthy of further investigation is the orientation of individual fibres with respect to the incident primary x-ray beam. The PSA prototype described in this study used parallel fibres in line with the primary beam’s central axis. However as a means of improving spatial resolution when using imagers with thick scintillators, others have suggested a so-called ‘focussed’ fibre geometry whereby the individual fibres are uniquely angled to match the off-axis divergence exhibited by clinical linear accelerator beams\cite{24,41}. Sawant et al. have estimated that beam divergence through a non-focussed 40 mm thick segmented array of crystal scintillators may cause losses in the MTF up to 15% in the periphery of a $40 \times 40$ cm$^2$ field\cite{24}. Noting the 30 mm thickness of the PSA prototype used in this study, it is reasonable to assume that improvements in both imaging and dosimetry performance may be realised by moving towards a focussed geometry.

8.3.2 Field size response

The field size output factors measured using the reference water-equivalent MatriXX ionisation chamber array and the prototype PSA EPID are shown in Figure 8.6. Field size factors simulated using the MC model are also shown for comparison. Error bars for each set of data have been included, though for the experimental measurements they were smaller than the markers used to illustrate the data.

The maximum percent difference between the MatriXX and PSA EPID measurements was 0.86% and occurred for the smallest field size ($2 \times 2$ cm$^2$). The close agreement between these measurements demonstrates the near water-equivalent response of this prototype detector. The maximum percent difference between the simulated and measured PSA-EPID field size factors was 1.0% and occurred for the $3 \times 3$ cm$^2$ field size. This close agreement between the simulation data and the experimental measurements serves as validation for the field size response simulated using the MC model.
8.3. Results and discussion

Figure 8.5: Comparison of the experimentally measured MTF and simulated MTFs calculated with and without a 0.5 mm air gap between the scintillator and photodiode plane. MTFs were calculated from the LSFs shown above in Figure 8.4.

8.3.3 Relative dose profiles

Uncorrected and corrected relative dose profiles measured using the prototype PSA-EPID and simulated with the MC model are shown in Figures 8.7 and 8.8 respectively. The significant variations in the measured and simulated response within the open field region of the uncorrected profiles stems from an unavoidable geometrical mismatch between the PSA fibres and the photodiode pixel array. Due to differences in the plastic fibre and a-Si photodiode areas, optical photons confined to a single fibre may in fact be incident on multiple photodiode pixels. A further complication exists with the physical prototype in that it is currently impractical to precisely align single rows or columns of fibres with rows or columns of pixels. The net result of these effects is that of quasi-periodic variations in the response between neighboring pixels, despite being irradiated by a uniform radiation field. The variation appears more regularly periodic in the simulation data owing to the consistent and repetitive nature of the fibre/photodiode mis-
alignment in this geometry. The magnitude of signal variation is approximately equal between the simulation and measured profiles for both field sizes studied.

This work investigated simple methods that may be applied to correct for this element-to-element mismatching artifact. Experimentally, one approach is simply to apply a flood field correction to the raw image. Assuming an unchanging detector setup, any observed variations in detector response due to a mismatch between the scintillator and photodiode panel will also be present in a flood field image. Therefore, applying the flood field correction will simply divide out this variation in response and result in a smooth dose profile. Due to the computational burden of simulating a large flood field with optical transport in the PSA, corrected dose profiles were instead simulated by changing the fibre cross-sectional area to $0.4 \times 0.4 \text{mm}^2$, resulting in perfect fibre-to-photodiode alignment.

Undesirable consequences of this flood field correction approach however include the removal of the detector off-axis response due to changes in the incident
8.3. Results and discussion

Figure 8.7: Comparison of the uncorrected relative dose profiles measured using the prototype EPID and simulated using the MC model for the (a) 5 × 5 and (b) 10 × 10 cm² open field sizes. Note the large signal variation in the open-field region resulting from mismatching between the PSA and the underlying photodiode array.

beam spectra at distances away from the central axis. Due to the generally softer beam energy away from the central axis, dose profiles in water typically exhibit a slight increase in response in this region (often referred to as the “horns” of the profile). The corrected simulation profiles will retain the correct off-axis response variations because no flood field correction was applied. While the corrected experimental and simulation profiles agree almost perfectly for the 5 × 5 cm² field size, the effect of removing the dose horns through the application of the flood field correction is more pronounced for the 10 × 10 cm² field size. A γ-index comparison between the experimental and simulated profiles for these field sizes quantitatively demonstrates this same effect\(^\text{[53]}\). Using 2%/2mm criteria and a relative dose threshold of 10%, 97.8% and 84.2% of data points passed for the 5 × 5 and 10 × 10 cm² fields, respectively.

Because the incident x-ray energy spectrum is relatively uniform across the 5 × 5 cm² field, the flood field correction approach to removing the variations in response for the experimental profile is sufficient. However, an alternative
8. A next-generation EPID: measurements & model validation

Figure 8.8: Comparison of the corrected relative dose profiles measured using the prototype EPID and simulated using the MC model for the (a) 5 × 5 and (b) 10 × 10 cm² open field sizes. The mismatching artifact present in the uncorrected images was removed by applying a flood field correction to experimental images and re-simulating the open field response using 0.4 × 0.4 mm² fibres to precisely match the photodiode array. Subplots indicate a γ-index analysis comparing the experimental and simulated profiles with 2%/2 mm criteria.

The approach should be used for larger field sizes including the 10 × 10 cm² field so as not to remove the off-axis response. While the investigation of alternative correction methods was beyond the scope of this study, others have reported novel binning correction techniques to improve the spatial resolution of similar thick, segmented scintillator detectors\cite{31}. While those authors used a photodiode array with a significantly higher resolution than the detector used in this study, the impact of using similar binning techniques to correct for the mismatching artifact with this detector will be the subject of future investigation.

8.4 Conclusions

This study reports the first experimental measurements taken using a novel, second-generation water-equivalent EPID employing an array of plastic scintillat-
8.4. Conclusions

ring fibres. Measurements were used firstly to characterise the detector’s response and secondly to empirically validate a Monte Carlo model of the prototype based on simulations quantifying the detector’s spatial resolution and dose response. As hypothesised, the prototype EPID exhibited a water-equivalent non-transit dose response that matched measurements performed using a MatriXX ion chamber array. Simulation results of the detector’s line spread function and modulation transfer function suggest the presence of a thin air gap between the scintillator and photodiode array possibly present during the measurements. Future work will endeavor to improve upon the quality of images acquired using this second-generation prototype by investigating methods to correct for the artifact introduced by scintillator fibre and photodiode pixel misalignment. Furthermore, the Monte Carlo model will be used to investigate potential configurations to further optimise the imaging and dosimetric response of a next-generation prototype.

Acknowledgments

The authors would like to acknowledge funding support from the Cancer Institute NSW (Research Equipment Grant 10/REG/1-20) and Cancer Council NSW (Grant ID RG 11-06). S.J.B. would also like to thank The University of Sydney and the Institute of Medical Physics for scholarship support, as well as the Liverpool and Macarthur Cancer Therapy Centers for additional financial support. The authors report no conflicts of interest in conducting this research.
References


References


References


(2013). URL http://scitation.aip.org/content/aapm/journal/medphys/40/9/10.1118/1.4816657. 220, 222, 229, 230


References


Conclusions and future work

Conclusions

The investigations contributing to this thesis have taken several steps forward in the pursuit to develop a next-generation device capable of simultaneous imaging and dose verification in radiotherapy. While recent studies have reported on high efficiency detectors for megavoltage (MV) imaging, the prototype EPID developed, optimised and evaluated in this work represents the first high efficiency MV imager to be physically constructed that also exhibits a demonstrated water equivalent response. Furthermore, the Monte Carlo (MC) model of the prototype EPID developed and validated here provides a valuable tool for ongoing optimisation of next-generation prototypes.

As a foundation upon which to develop a MC model of the prototype EPID, models of a clinical 6 MV photon source (Chapter 3) and a standard copper plate/phosphor screen a-Si EPID were first developed and experimentally validated. While previous studies have reported similar models, the model devel-
9. Conclusions and future work

veloped here is the first to characterise optical photon transport self-consistently with x-ray and electron transport in the standard EPID by using the GEANT4 MC toolkit. Results further confirmed that although simulating optical transport is necessary to predict the EPID point spread function and hence imaging performance, energy deposition in the phosphor screen is an accurate surrogate for predicting EPID dose response to large fields in non-transit (Chapter 4) and transit (Chapter 5) configurations.

Characterising the imaging and dosimetry performance of the prototype EPID constitutes the first experimental demonstration reported in the literature of indirect MV x-ray detection using a segmented low-density scintillator (Chapter 6). Results demonstrating its water-equivalent response in non-transit and transit configurations were perhaps of greatest relevance considering the goal of developing a novel EPID suitable for applications in dosimetry as well as imaging. While images acquired with this first-generation prototype were of an inferior quality relative to images acquired with a standard EPID, they were nonetheless capable of visualising gross anatomical features in an anthropomorphic phantom and several opportunities for imaging performance optimisation were identified.

Having benchmarked the initial prototype array of plastic scintillating fibres, a second-generation prototype was developed using fibres that were longer and narrower in an attempt to improve sensitivity and spatial resolution. A MC model based on the dimensions of the second-generation prototype was developed and its sensitivity to a range of geometrical and optical transport parameters was studied to determine an optimised configuration (Chapter 7). Simulation of the extramural absorber that surrounds each fibre in the prototype array was found to be critically important for simulations predicting the detector modulation transfer function (MTF) and open field dose response. Element-to-element mismatch between scintillating fibres and photodiode pixels was also found to be important, causing the MTF to decrease at high spatial frequencies and introducing quasi-periodic variations in open field dose profiles. Measurements performed using the second-generation prototype further demonstrated its water-equivalent response and were used to validate the MC model (Chapter 8). A simple method to improve upon the quasi-periodic profile response resulting from element-to-element mismatch was also demonstrated to be suitable for small fields.
Future work

Ongoing optimisation of the prototype EPID is required to continue improving upon its imaging performance. For this type of detector to be used clinically for simultaneous imaging and dose verification, its performance as an imager should ideally match or exceed current commercially available a-Si EPIDs. The results presented in this thesis confirm that the water equivalent dose response remains unaffected by changes in the optical properties of the plastic scintillator fibres. Therefore, further improvements in imaging performance may be realised through optimisation of select optical properties (such as the optical yield, for example) without affecting the dose response. Opportunities for continued design optimisation also include the fabrication of fibres with still narrower physical dimensions to improve upon spatial resolution. Post-processing methods to correct images exhibiting element-to-element mismatch artifacts, including those referenced in Chapter 8, should also be investigated.

One particular design that warrants detailed investigation in future studies is the use of an optical coupling agent between the scintillator array and the photodiodes to maximise light detection. The discovery that the presence of air between the scintillator and photodiodes reduces the detector MTF (Chapter 8) suggests that optical coupling may offer further improvements in imaging performance. The exploration of alternative cladding materials with different optical properties is also warranted to investigate whether there is scope to decrease optical signal loss in the plastic scintillator fibres beyond that which is present in the current design. Another avenue for future work is the use of geometrically divergent scintillator arrays designed to match the divergence exhibited by clinical linear accelerator beams. A potential disadvantage to this design is that such a detector would exhibit optimal performance at a fixed distance from the accelerator target. Nevertheless the likely improvements in spatial resolution that would result by accounting for beam divergence in this manner are worthy of further exploration.

Ultimately, the feasibility of developing a novel MV imaging device with a water-equivalent dose response has been demonstrated through this work. Ongoing efforts to optimise the design of the existing prototype for radiotherapy
9. Conclusions and future work

imaging and dosimetry will now be able to progress at an accelerated rate by taking advantage of the validated MC model. It is anticipated that through these efforts and the development of advanced fabrication techniques, a next-generation EPID may be realised with the ability to serve as an essential clinical tool for comprehensive radiotherapy treatment verification.
Appendix – Abstracts from conference presentations

This appendix contains, in chronological order, the abstracts for work presented in the form of posters and oral presentations at conferences throughout candidature.
Towards the development of a comprehensive model of an electronic portal imaging device for advanced radiotherapy applications

S. Blake, P. Vial, L. Holloway and Z. Kuncic

Oral presentation at the Cancer Research Network Symposium for Postgraduate Students in Sydney, Australia – December, 2010

Background: Electronic portal imaging devices (EPIDs) are of major significance to the field of radiation oncology due to their applications to external beam megavoltage (MV) radiotherapy. Approximately half of all cancer patients receive radiotherapy during treatment, the majority of which is delivered externally. The amorphous silicon (a-Si) flat panel EPID shows enormous promise for its use as a 2-dimensional dosimeter due to its large, high-resolution detector area and real-time acquisition capabilities. As a dosimeter, it would serve to verify accurate treatment delivery and to indicate when treatment adaptations may be advantageous. This work represents the first stage of an ongoing study to investigate the physical processes occurring within EPIDs, including the effects of optical scattering on image quality and dosimetry.

Methods: Data from the Phase-space database for external beam radiotherapy (International Atomic Energy Agency, IAEA) was used with the GEANT4 software toolkit to construct a Monte Carlo model of a Siemens Primus linear accelerator (linac) 6 MV photon source. Dose profiles and percent depth dose (PDD) curves were extracted from simulations of dose in water and compared to experimental measurements. A preliminary EPID model was developed to incorporate both high energy radiation and optical photon transport.

Results: Mean agreement in dose profiles inside the open beam was within 0.5%. Mean agreement in PDD curves beyond depth of dose maximum was within 1.1% (local percent difference). The radiation transport of both high energy and optical photons were visualized in the EPID simulation. Further work is under way to experimentally validate the EPID model.
Conclusions: The comparison of simulated dose in water with measurements indicates that the IAEA phase-space represents an accurate model of a linac source. We have demonstrated the feasibility of developing a comprehensive EPID model incorporating both high energy and optical physics.
Towards the development of a comprehensive model of an electronic portal imaging device using Geant4

S. Blake, P. Vial, L. Holloway and Z. Kuncic

Poster presentation at the Engineering and Physical Sciences in Medicine and the Australian Biomedical Engineering Conference (EPSM-ABEC) annual meeting in Melbourne, Australia – December, 2010

Objective: This work represents the first stage of an ongoing study to investigate the physical processes occurring within electronic portal imaging devices (EPIDs), including the effects of optical scattering on image quality and dosimetry. The objective of this work was to develop an initial Monte Carlo model of a linear accelerator (linac) beam and an EPID. The ability to simulate the radiation transport of both high energy and optical photons in a single Monte Carlo model was tested.

Methods: Data from the Phase-space database for external beam radiotherapy (International Atomic Energy Agency, IAEA) was used with the GEANT4 toolkit to construct a model of a Siemens Primus linac 6 MV photon source. Dose profiles and percent depth dose (PDD) curves were extracted from simulations of dose in water and compared to experimental measurements. A preliminary EPID model was developed to incorporate both high energy radiation and optical photon transport.

Results: Agreement in dose profiles inside the open beam was within 1.6%. Mean agreement in PDD curves beyond depth of dose maximum was within 6.1% (local percent difference). The radiation transport of both high energy and optical photons were simulated and visualized in the EPID model. Further work is required to experimentally validate the EPID model.

Conclusions: The comparison of simulated dose in water with measurements indicates that the IAEA phase-space may represent an accurate model of a linac source. We have demonstrated the feasibility of developing a comprehensive EPID model incorporating both high energy and optical physics in GEANT4.
Modelling of a radiotherapy linac beam and portal imager using Geant4: A feasibility study

S. Blake, P. Vial, L. Holloway and Z. Kuncic

Oral presentation at the 5th Student Research Symposium of the ACT/NSW Branch of the Australasian College of Physical Scientists and Engineering in Medicine in Sydney, Australia – December, 2010

Introduction: This work represents the first stage of an ongoing study to investigate the physical processes operating within electronic portal imaging devices (EPIDs), including the effects of optical scattering on image quality and dosimetry. The objective of this work was to develop a Monte Carlo (MC) model of a linear accelerator (linac) beam and an EPID using the GEANT4 MC toolkit[1]. We tested the capability of GEANT4 to model both the high energy and optical physics relevant for EPIDs, which have not previously been modelled simultaneously.

Methods: A preliminary EPID model was developed to incorporate high energy and optical photon transport. Data from the IAEA Phase-space database for external beam radiotherapy was used with GEANT4[2] to build a Siemens Primus linac 6 MV source model. To benchmark the quality of this phase-space (phsp) source, relative dose profiles and percent depth dose (PDD) curves were extracted from simulations of dose in water and compared to measurements with a CC13 compact ionization chamber (IC). Uncertainty in the mean energy deposited along the beam central axis was calculated using the method described by Walters et al.[3]

Results: Mean agreement in dose profiles inside the open beam was within 1.6%. Mean agreement in PDD curves beyond depth of dose maximum ($d_{\text{max}}$) was within 1.0% (local percent difference). Latent uncertainties in the energy deposited along the beam central axis using the phsp files are between approximately 1–3%. The radiation transport of both high energy and optical photons were simulated and visualized in the EPID model.
Appendix

Discussion: The comparison of simulated dose in water with measurements indicates that the IAEA phsp may represent an accurate model of a linac source. To exclude dose-volume effects inherent in IC measurements, agreement between simulation and experimental data was only evaluated within the open field of the beam (profiles) and beyond $d_{\text{max}}$ (PDDs). To further benchmark the linac source model, work is required to experimentally validate the energy deposited outside of the open field (profiles) and within the build-up region (PDDs).

Conclusions: We have demonstrated the feasibility of developing a comprehensive EPID model incorporating high energy physics in GEANT4. Future work will include refining and testing the optical physics in our model to optimize imaging and dosimetry capabilities.

References:


Developing a model of an electronic portal imaging device using Geant4

S. Blake, P. Vial, L. Holloway, P. Greer and Z. Kuncic

*Oral presentation at the 1st GEANT4 Australian School and User Workshop in Wollongong, Australia – April, 2011*

**Introduction:** Electronic portal imaging devices (EPIDs) are silicon-based detectors that are used regularly in radiotherapy for imaging and, more recently, dosimetry. This work forms part of an ongoing study[1,2,3] to investigate the physical processes operating within EPIDs, including the effects of optical scintillation and scattering on image quality and dosimetry. The objective of this work is to develop and benchmark a preliminary Monte Carlo (MC) model of a linear accelerator (linac) beam and an EPID using the GEANT4 MC toolkit[4]. We tested the capability of GEANT4 to model both the high energy and optical physics relevant for EPIDs, which have not previously been modelled simultaneously.

**Materials and methods:** A preliminary linac and EPID model was developed to incorporate both high energy and optical photon transport. Data from the IAEA Phase-space database for external beam radiotherapy was used with GEANT4[5] to build a Siemens Primus linac 6 MV source model. Phase-space (phsp) files for five square-shaped beam field sizes were used (3 × 3, 5 × 5, 10 × 10, 15 × 15 and 20 × 20 cm²). To benchmark the quality of these phsp sources, relative dose profiles and percent depth dose (PDD) curves were extracted from simulations of dose in water and compared to measurements taken using a CC13 compact ionization chamber (IC). Uncertainty in the mean energy deposited along the beam central axis was calculated using the method described by Walters *et al.*[6] Latent uncertainty in the IAEA phsp files has been estimated using the method described by Sempau *et al.*[7]

**Results:** The radiation transport of both high energy and optical photons were simulated and visualized in the EPID model. Mean agreement in dose profiles inside the open beam was within 1.6% when benchmarking the linac
source model. Furthermore, mean agreement in PDD curves beyond depth of dose maximum ($d_{\text{max}}$) was within 1.0% (local percent difference). Latent uncertainties in the energy deposited along the beam central axis using the phsp files are within 1–3%.

**Conclusions:** We have developed a preliminary EPID model incorporating both high energy and optical physics in GEANT4. The comparison of simulated dose in water with measurements indicates that the IAEA phsp may represent an accurate model of a linac source. To further benchmark the linac source model, work is required to experimentally validate the energy deposited outside of the open field (profiles) and within the build-up region (PDDs). Future work will also include refining and testing the optical physics in our model to optimize imaging and dosimetry performance. This performance may be characterized by simulating the Point Spread Function, Modulation Transfer Function, and Detective Quantum Efficiency of the EPID in this model.

**References:**

Self-consistent Monte Carlo simulation of x-ray and optical photon transport in electronic portal imaging devices

S. Blake, P. Vial, L. Holloway, P. Greer and Z. Kuncic

Oral presentation at the International Workshop on Recent Advances in Monte Carlo Techniques for Radiation Therapy in Montreal, Canada – June, 2011

**Purpose:** To self-consistently model x-ray and optical photon transport within an indirect-detection electronic portal imaging device (EPID) using Monte Carlo (MC) methods and to quantify the effect of optical scatter on the output signal.

**Materials and Method:** A generic indirect-detection EPID model was developed using the GEANT4 MC toolkit. The EPID was modeled as a series of uniform slabs with thicknesses and material properties obtained from published literature. The standard GEANT4 electromagnetic and optical physics modules were incorporated into the model to self-consistently simulate both x-ray and optical photon transport relevant for indirect-detection EPIDs. Preliminary model response was investigated using simulations of a narrow monoenergetic beam of 1 MeV photons normally incident on the EPID surface. The beam width was equal to the photodiode pixel pitch of 0.4 mm, generating a line of radiation incident on the EPID surface. Particle hits and energy deposition were scored in the gadolinium oxysulfide scintillator and amorphous silicon photodiode layers. Optical and x-ray photons were scored separately in the photodiode layer to measure their relative effects on the output signal. Line spread functions (LSFs) were generated to indicate the distribution of hits and energy deposited across the scintillator and photodiode planes.

**Results:** Preliminary LSFs have been generated for energy deposition events scored in the scintillator and optical photon hits scored in the photodiode. Current work involving the validation of optical transport within this model will be presented.
Appendix

Conclusions: Initial LSF simulations suggest a small but non-negligible contribution of optical photon scatter to the output EPID signal. Modeling of optical photon transport may therefore be important when simulating imager performance for an indirect-detection EPID. Validation of the optical transport modeling is required to more accurately quantify imager LSFs.
An investigation into optical photon transport effects on electronic portal imaging performance using Geant4

S. Blake, P. Vial, L. Holloway, P. Greer and Z. Kuncic

*Oral presentation in the Young Investigators Symposium at The 2011 Joint American Association of Physicists in Medicine (AAPM)/Canadian Organization of Medical Physicists (COMP) annual meeting in Vancouver, Canada – August, 2011*

**Purpose:** To develop a comprehensive Monte Carlo (MC) model of an indirect-detection electronic portal imaging device (EPID) that can self-consistently quantify the effect of optical blur on the output signal.

**Method and Materials:** A model of an indirect-detection EPID was developed using the GEANT4 MC toolkit. The EPID was modeled as a series of uniform slabs with thicknesses and material properties obtained from published literature. The model also included a slab of solid water backscatter material directly beyond the EPID rear housing. The standard electromagnetic and optical physics GEANT4 modules were incorporated into the model to simultaneously simulate both high energy and optical photon transport relevant for indirect-detection EPIDs. A narrow, monoenergetic beam of 1 MeV photons was used to generate a line of radiation normally incident on the EPID surface. The beam width was equal to the pixel pitch of 0.4 mm used for scoring particle hits and energy deposition in the gadolinium oxysulfide scintillator and amorphous silicon photodiode layers. Optical and gamma photons were scored separately in the photodiode layer to measure their relative effects on the output signal. Line spread functions (LSFs) were generated indicating the distribution of hits and energy deposited across the scintillator and photodiode planes.

**Results:** The LSFs for optical photon hits in the photodiode array and energy deposition events in the scintillator had a FWHM of approximately 4.7 mm and 0.82 mm, respectively. This indicates a significant increase in image blurring due to optical photon scatter.
Appendix

Conclusions: Our results indicate that modeling optical photon transport may be important when simulating imager performance for an indirect-detection EPID. Further analysis of calculated LSFs, including determination of the detector modulation transfer function, is required to further quantify imager performance.
Preliminary investigation of optical photon transport in electronic portal imaging devices

S. Blake, P. Vial, L. Holloway and Z. Kuncic

Poster presentation at the Engineering and Physical Sciences in Medicine and the Australian Biomedical Engineering Conference (EPSM-ABEC) annual meeting in Darwin, Australia – August, 2011

Objective: To quantify the effect of optical photon transport on the output signal of an indirect-detection electronic portal imaging device (EPID) using the GEANT4 Monte Carlo (MC) toolkit.

Methods: The GEANT4 MC toolkit has been used to develop a model of a generic indirect-detection EPID based on geometrical data and material properties obtained from published literature. X-ray and optical photon transport were modeled self-consistently within the EPID using the GEANT4 electromagnetic and optical physics modules. Narrow beams of 1 MeV monoenergetic x-rays, normally incident on the EPID surface, were used in a preliminary investigation of the detector response. The beam width was equal to the photodiode pixel pitch of 0.4 mm. Line spread functions (LSFs) of energy deposition events and photon hits were scored in the gadolinium oxysulfide scintillator and amorphous silicon photodiode layers. X-ray and optical photons were scored separately to measure their relative effects on the output signal.

Results: The full width at half maximum (FWHM) and full width at tenth maximum (FWTM) of the LSF scoring energy deposited in the scintillator are 0.82 mm and 1.16 mm respectively. The FWHM and FWTM of the LSF scoring optical photon hits in the photodiode are 0.94 mm and 2.10 mm respectively.

Conclusions: Preliminary results suggest a small but non-negligible contribution of optical photon scatter to the output EPID LSF signal. Modeling optical photon transport may therefore be important when simulating indirect-detection EPID imaging performance. Validation of the optical transport modeling parameters is required to more accurately quantify EPID LSFs.
Monte Carlo investigation of optical photon transport effects on electronic portal imaging device dosimetric response

S. Blake, P. Vial, L. Holloway, P. Greer and Z. Kuncic

Oral presentation at the 6th Student Research Symposium of the ACT/NSW Branch of the Australasian College of Physical Scientists and Engineering in Medicine in Sydney, Australia – December, 2011

Introduction: Amorphous silicon (a-Si) electronic portal imaging devices (EPIDs) have been demonstrated to be suitable for radiotherapy dosimetry applications due in part to their high spatial resolution and real-time data acquisition capabilities. Such EPIDs indirectly detect radiation by means of a Gd$_2$O$_2$S:Tb phosphor screen that converts incident radiation into optical photons that are detected by the a-Si photodiode array. The phosphor screen improves the detection efficiency of the EPID; however it results in an over-response to low energy radiation when compared to more water equivalent dosimeters. Many previous studies using Monte Carlo (MC) methods to model the dosimetric response of a-Si EPIDs have not fully accounted for the transport of optical photons within these detectors. The goal of this work was to develop a MC model of an a-Si EPID that could self-consistently simulate both x-ray and optical photon transport and to thereby investigate the effects of optical photon transport on EPID dosimetric response.

Methods: A model of an indirect-detection a-Si EPID was developed using the Geant4 MC toolkit. The EPID was modeled as a series of uniform slabs with thicknesses and material properties based on specifications from the manufacturer of a research detector used in this study. The standard electromagnetic and optical physics Geant4 classes were incorporated into the model. Phase-space data for a 6 MV Elekta Synergy photon source was used to generate static open fields ($5 \times 5$ and $10 \times 10\,\text{cm}^2$) incident on the EPID surface. Optical photon absorption events were scored in the a-Si photodiode layer. Experimental EPID images were also obtained by exposing a research EPID to Elekta Synergy 6 MV
photon beams for the aforementioned open field sizes. Normalized simulation profiles scoring the optical photons absorbed in the photodiode were compared to normalized profiles extracted from experimental EPID images.

**Results:** Local percent difference (LPD) between the normalized simulation profiles scoring optical photon absorption events in the photodiode and the normalized experimental EPID images was calculated within the open field region for each field size studied. The mean and maximum LPDs were 0.23% and 2.8% respectively for the $5 \times 5 \text{cm}^2$ open field and 1.7% and 5.9% respectively for the $10 \times 10 \text{cm}^2$ field.

**Conclusions:** A MC model of an $a$-Si EPID that self-consistently simulates x-ray and optical photon transport within the detector was developed using GEANT4. Preliminary results comparing normalized profiles scoring optical photon absorption events in the photodiode to normalized profiles extracted from experimental images indicate mean agreement within the open field region of less than 2%. Similar investigations of dosimetric EPID response to larger field sizes are underway. Furthermore, to better quantify optical photon transport effects in this model current work is focusing on scoring energy deposition events in the phosphor layer and comparing resulting profiles with those reported in this study.
Monte Carlo modeling of optical photon transport effects on electronic portal imaging device dosimetric response

S. Blake, P. Vial, L. Holloway, P. Greer and Z. Kuncic

Oral presentation at the Electronic Patient Imaging 2012 (EPI2k12) conference in Sydney, Australia – March, 2012

Introduction: Amorphous silicon (a-Si) electronic portal imaging devices (EPIDs) have been demonstrated to be suitable for radiotherapy dosimetry applications due in part to their high spatial resolution, real-time data acquisition capabilities and resilience to radiation-induced damage. Commercially available a-Si EPIDs, however, contain a number of non water-equivalent components that complicate their use for dosimetry. Such EPIDs indirectly detect radiation by means of a gadolinium oxysulfide phosphor screen that converts incident radiation into optical photons that are detected by the a-Si photodiode array. While this phosphor screen is used to improve the detective quantum efficiency of the EPID, it results in an over-response to low energy radiation in comparison to more water equivalent dosimeters. Previous studies have used Monte Carlo (MC) simulation methods to model the dosimetric response of a-Si EPIDs. However, the majority of these studies have not fully accounted for the transport of optical photons within these detectors. The goal of this work was to develop a MC model of an a-Si EPID that could self-consistently simulate both x-ray and optical photon transport relevant for indirect-detection EPIDs. This model was then used to investigate the effects of optical photon transport on EPID dosimetric response.

Methods and Materials: A model of an indirect-detection a-Si EPID was developed using the GEANT4 MC toolkit. The EPID was modeled as a series of uniform slabs with thicknesses and material properties based on specifications from the manufacturer of a research detector used in this study. The standard electromagnetic and optical physics GEANT4 classes were incorporated into the model. Phase-space data for a 6 MV Elekta Synergy photon source was used to generate a static open 10 × 10 cm² field incident on the EPID surface. Energy deposition events were scored in the gadolinium oxysulfide phosphor layer...
and optical photon absorption events were scored in the $\alpha$-Si photodiode layers. Experimental EPID images were also obtained by exposing a research EPID to Elekta Synergy 6 MV photon beams for the $10 \times 10 \text{ cm}^2$ open field. Normalized cross-plane profiles scoring energy deposition events in the phosphor and optical photons absorbed in the photodiode were compared with normalized profiles extracted from experimental EPID images.

**Results:** Local percent difference (LPD) between the normalized simulation profiles and the normalized experimental EPID profiles was calculated within the open field region. The mean and maximum LPD were 1.5% and 5.6% respectively between the simulation profile scoring optical photon absorption events in the photodiode and the experimental profile. The mean and maximum LPD were 1.1% and 3.3% respectively between the simulation profile scoring energy deposition events in the phosphor and the experimental profile.

**Conclusions:** A MC model of an $\alpha$-Si EPID that self-consistently simulates x-ray and optical photon transport within the detector has been developed using GEANT4. Preliminary results comparing normalized profiles scoring optical photon absorption events in the photodiode and energy deposition events in the phosphor to normalized profiles extracted from experimental images indicate mean agreement within the open field region of less than 2%. Similar investigations of dosimetric EPID response to smaller and larger field sizes are underway. Furthermore, to better quantify optical photon transport effects in this model, current work is focusing on evaluating potential differences between the scoring methods outside of the open field region of the beam.
Appendix

Sensitivity analysis of an electronic portal imaging device Monte Carlo model to variations in optical transport parameters

S. Blake, P. Vial, L. Holloway, A. McNamara, P. Greer and Z. Kuncic

Poster presentation at The 2012 American Association of Physicists in Medicine (AAPM) annual meeting in Charlotte, NC USA – August, 2012

Purpose: To investigate the sensitivity of a Monte Carlo (MC) model of a standard clinical amorphous silicon (a-Si) electron portal imaging device (EPID) to variations in optical photon transport parameters.

Methods: The GEANT4 MC toolkit was used to develop a comprehensive model of an indirect-detection a-Si EPID incorporating x-ray and optical photon transport. The EPID was modeled as a series of uniform layers with properties specified by the manufacturer (PerkinElmer, Santa Clara, CA) of a research EPID at our centre. Optical processes that were modeled include bulk absorption, Rayleigh scattering, and boundary processes (reflection and refraction). Model performance was evaluated by scoring optical photons absorbed by the a-Si photodiode as a function of radial distance from a point source of x-rays on an event-by-event basis (0.025 mm resolution). Primary x-ray energies were sampled from a clinical 6 MV photon spectrum. Simulations were performed by varying optical transport parameters and the resulting point spread functions (PSFs) were compared. The optical parameters investigated include: x-ray transport cutoff thresholds; absorption path length; optical energy spectrum; refractive indices; and the ‘roughness’ of boundaries within phosphor screen layers.

Results: The transport cutoffs and refractive indices studied were found to minimally affect resulting PSFs. A monoenergetic optical spectrum slightly broadened the PSF in comparison with the use of a polyenergetic spectrum. The absorption path length only significantly altered the PSF when decreased drastically. Variations in the treatment of boundaries noticeably broadened resulting PSFs.
Conclusion: Variation in optical transport parameters was found to affect resulting PSF calculations. Current work is focusing on repeating this analysis with a coarser resolution more typical of a commercial a-Si EPID to observe if these effects continue to alter the EPID PSF. Experimental measurement of the EPID line spread function to validate these results is also underway.
Towards a Next-Generation Electronic Portal Device for Radiotherapy Imaging and Dosimetry

S. Blake, P. Vial, L. Holloway, A. L. McNamara, P. B. Greer and Z. Kuncic

Oral presentation at the Cancer Research Network Symposium for Postgraduate Students in Sydney, Australia – November, 2012

Introduction: Electronic portal imaging devices (EPIDs) are x-ray detectors that have many clinical applications in radiation oncology. While optimized for imaging uses, including verification of patient positioning prior to treatment, EPIDs may also be used to verify the radiation dose delivered during radiotherapy treatments. EPIDs use a phosphor screen to convert incident x-rays into optical photons which are then detected by a photodiode array. Studies using Monte Carlo (MC) simulations to model EPID response, however, often neglect optical transport within these detectors. In this study a comprehensive MC model of an EPID was developed to self-consistently simulate x-ray and optical transport, and thereby investigate optical transport effects on EPID response.

Methods: The GEANT4 MC software toolkit was used to develop a model of an EPID incorporating x-ray and optical transport. A clinical 6 MV photon beam source was also developed and integrated into the model. This was validated using experimental images acquired with a research EPID at the Liverpool and Macarthur Cancer Therapy Centres. Energy deposited in the phosphor and optical absorption events in the photodiode were recorded in response to photon beams of varying size. Beam profiles were then normalized and compared to quantify optical transport effects within the detector. Recently the MC model has been modified to incorporate CT images, enabling a patient/phantom geometry to be positioned in the beam line (transit configuration).

Results: More than 94% of all data points for the simulated EPID dose profiles agreed with experimental measurements. Differences in model response resulting from optical transport were not found to be statistically significant.

Conclusions: Optical photon transport contributed a negligible change in EPID dosimetric response for a non-transit configuration. Current work is inves-
tigating optical transport effects in transit dosimetry, as well as next-generation EPIDs utilizing different materials to optimize their use for radiotherapy dosimetry applications.
Appendix

Towards a Next-Generation Electronic Portal Device for Dual-Mode Imaging and Dosimetry

S. Blake, P. Vial, L. Holloway, A. L. McNamara, P. B. Greer and Z. Kuncic

Poster presentation at the 20th Australian Institute of Physics Congress in Sydney, Australia – December, 2012

Introduction: Amorphous silicon (a-Si) electronic portal imaging devices (EPIDs) are flat panel x-ray detectors that have found many clinical applications particularly in radiation oncology. They are most frequently used for patient imaging in image-guided radiotherapy to verify patient positioning prior to treatment. More recently, EPIDs have also been demonstrated to be suitable for dosimetry applications, such as to verify the radiation dose delivered during radiotherapy treatment[1]. EPIDs employ a metal plate and phosphor scintillator screen to indirectly detect x-rays and thereby increase their efficiency. X-rays and electrons interact in the phosphor to generate optical photons which are then absorbed by an a-Si photodiode array. Optical photons are absorbed by individual photodiode pixels, leading to a charge buildup that can be read out to form a digital image. Many studies using Monte Carlo (MC) simulations to simulate EPID response have made simplifying assumptions regarding optical transport within the detectors. In most cases, the dose deposited in the phosphor is the quantity used to predict dose response and optical transport is either empirically corrected for or neglected, e.g. [2]. In this study, a comprehensive MC model of an a-Si EPID was developed to investigate optical transport effects on dosimetric response. This is the first EPID model to self-consistently model both x-ray and optical photon transport.

Methods: The GEANT4 MC radiation transport toolkit was used to develop a detailed model of a research EPID. Electromagnetic and optical physics classes were used to simulate radiation transport in the EPID. Material and optical properties were based on information provided by the manufacturer or literature values. A model of a clinical 6 MV photon beam source was also developed and integrated into the simulation. To validate the model, experimental images
were acquired using the research EPID under the same irradiation conditions. Histograms scoring the 2D spatial distribution of energy deposition and optical absorption events (0.4 × 0.4 mm² resolution) were scored in the phosphor and photodiode planes, respectively. These histograms were then normalized and compared to evaluate the effects of optical photon transport within the detector. Histograms were also compared with experimental images to observe differences in agreement between scoring methods. Initial simulations were performed in a non-transit dosimetry configuration – that is, with nothing between the radiation source and the EPID. Recently the MC model has been modified to enable a transit dosimetry configuration with a phantom based on a DICOM computed tomography image set positioned in the beam line.

Discussion and conclusions: Optical photon transport was found to contribute a non-negligible amount of signal towards the EPID response for irradiations in non-transit dosimetry. Optical scatter, predominantly within the phosphor layer, acted to slightly broaden dose profiles relative to those obtained by scoring energy deposited in the phosphor layer. Improved agreement with experimental images was obtained when scoring optical absorption events. Current work is investigating the impact of optical transport on EPID response for transit dosimetry.

References:


Introduction: Amorphous silicon (a-Si) electronic portal imaging devices (EPIDs) are x-ray detectors that have realized many clinical applications in radiation oncology. Commercially available EPIDs use a metal plate and phosphor screen to convert incident x-rays into optical photons which are then detected by an a-Si photodiode array. While this detection scheme optimizes image contrast and noise properties, the high atomic number materials result in a non water-equivalent dosimetric response. By replacing the metal plate/phosphor screen with an array of water-equivalent plastic scintillating fibers, it is hypothesized that the EPID may be used to simultaneously image the patient while verifying the radiation dose delivered during radiotherapy treatments.

In this study, the imaging and dosimetric capabilities of a research EPID were evaluated for both the standard clinical configuration and the proposed configuration which employed a prototype plastic scintillator array (PSA). In addition, a comprehensive Monte Carlo (MC) model of the a-Si EPID was developed to self-consistently simulate x-ray and optical transport within the standard and prototype EPID configurations. This model will facilitate the optimization of the proposed design and thereby play an important role in the development of a next-generation EPID capable of simultaneous imaging and dosimetry in radiotherapy.

Methods: The imaging performance of the standard and prototype EPIDs was evaluated experimentally by acquiring images of a QC-3V phantom to quantify the resulting contrast-to-noise ratio (CNR) and spatial resolution of each configuration. The dosimetric response of the standard and prototype EPIDs was evaluated experimentally by measuring field size response in non-transit and
transit geometries. Dosimetric response was compared to that of an ion chamber in solid water to determine the water-equivalency of each configuration.

The GEANT4 MC toolkit was used to develop a model of an EPID incorporating electromagnetic and optical transport. A clinical 6 MV photon beam source was developed and integrated into the model. The model may also incorporate clinical computed tomography images to create a patient/phantom geometry positioned in the beam line. The energy deposited in the phosphor/PSA was scored to approximate EPID response in each configuration.

Results: Experimental images acquired using the QC-3V phantom indicate that the prototype EPID configuration has decreased CNR and spatial resolution relative to the standard configuration. Measurements and simulations have, however, demonstrated that the prototype configuration responds in an approximately water-equivalent manner whereas the standard configuration overresponds to low energy radiation relative to an ion chamber.

Conclusions: The prototype EPID employing a PSA responds in an approximately water-equivalent manner, suggesting that images acquired using such a detector may be useful for applications in dosimetry. The decreased CNR and spatial resolution of the prototype configuration may be improved upon by optimizing the design of the PSA, in part through the use of MC simulations. Ongoing investigations are using the MC model to quantify optical transport effects within the PSA on detector response in the prototype configuration.
Appendix

Investigation of optical transport within a novel plastic scintillator imaging device

S. Blake, A. L. McNamara, P. Vial, L. Holloway, P. Greer and Z. Kuncic

Oral presentation at the 2nd GEANT4 Australian School and Monte Carlo Workshop in Wollongong, Australia – April, 2013

**Purpose:** Amorphous silicon (a-Si) electronic portal imaging devices (EPIDs) are x-ray detectors frequently used in radiotherapy imaging and dosimetry applications. Standard EPIDs employ a copper plate and Gd_{2}O_{2}S:Tb phosphor screen to convert x-rays into optical photons that may be detected using an array of a-Si photodiodes. This indirect-detection system optimizes the EPID sensitivity to incident x-rays, however the high atomic number materials cause a non water-equivalent response that is not ideal for dosimetry. An alternative configuration that replaces the copper and phosphor screen with a low-density plastic scintillator is under ongoing investigation by our group[1]. This plastic scintillator exhibits a water-equivalent response, albeit with a reduced x-ray detection efficiency that necessitates the use of a thick scintillator. In this study, Monte Carlo (MC) simulations were used to characterize optical transport within two alternate configurations of this novel EPID, employing either a single block of plastic scintillator or a segmented array of plastic scintillating fibers.

**Methods:** GEANT4 was previously used to develop and experimentally validate a model of a standard a-Si EPID[2]. In the present study, the EPID model was modified by replacing all of the components above the a-Si photodiode layer (including the copper plate and phosphor screen) with a layer of plastic scintillator. The plastic scintillator geometries are based on experimental prototypes currently under investigation by our group. One design utilises a 15 × 15 cm^2 block of plastic scintillator with uniform thicknesses ranging from 1–50 mm. A second design utilises a 15 × 15 cm^2 segmented array of plastic scintillating fibers, each having a cross-sectional area of 1 × 1 mm^2 and 15 mm thickness. Material properties were based on specifications provided by the manufacturer of the experimental prototypes. The standard GEANT4 electromagnetic and optical
physics classes were used. Energy deposited in the scintillator and optical photons absorbed by the $a$-Si photodiode were separately scored to quantify optical transport effects on detector response. The detector imaging performance was evaluated for each design by using a point source of monoenergetic x-rays (0.1 – 10 MeV) to calculate the x-ray detection efficiency, point spread function (PSF) and modulation transfer function (MTF). Dosimetric response was investigated for each design by using phase-space files of 6 MV x-ray beams with varying field size to calculate field size factors and relative beam profiles.

**Results:** Simulations of the EPID PSF demonstrated noticeable signal blurring from optical scattering within the plastic scintillator. Increased blurring was observed for thicker blocks of plastic scintillator and the optical scattering resulted in dose profiles that were distinctly rounded with very wide penumbral regions. The blurring was, however, significantly reduced in the segmented array design and the relative dose profiles had much steeper penumbrae. Validation of optical transport parameters against experimental measurements is currently under investigation.

**Conclusions:** Explicit modelling of optical transport is important when using MC simulations to predict the response of novel EPIDs incorporating thick plastic scintillators. The segmented array of plastic scintillating fibers greatly limits the lateral spread of optical photons over the unsegmented block, resulting in improved spatial resolution and steeper penumbrae in dose profiles. Further work is required to validate the EPID models against measurements taken with the experimental prototypes.

**References:**


Monte Carlo simulation of electromagnetic and optical transport within $a$-Si electronic portal imaging devices

S. Blake, A. L. McNamara, P. Vial, L. Holloway, P. Greer and Z. Kuncic

*Poster presentation at the International Conference on the Use of Computers in Radiation Therapy in Melbourne, Australia – May, 2013*

**Purpose:** Amorphous silicon ($a$-Si) electronic portal imaging devices (EPIDs) are x-ray detectors frequently used in radiotherapy imaging and dosimetry applications. Standard EPIDs employ a Gd$_2$O$_2$S:Tb phosphor screen to convert x-rays into optical photons albeit with a non water-equivalent response. In this study, Monte Carlo (MC) simulations were used to characterize optical transport within the standard EPID and a novel EPID that improves the water-equivalent response by replacing the phosphor screen with a plastic scintillator array (PSA).

**Methods:** The GEANT4 MC toolkit was used to develop $a$-Si EPID models (standard and novel configurations) incorporating electromagnetic and optical transport. Energy deposited in the phosphor/PSA and optical photons absorbed by the $a$-Si photodiode were compared to quantify differences in simulated response. The imaging performance and dosimetric response of each configuration was investigated using a point x-ray source and open field x-ray beams, respectively. Patient/phantom CT images were recently integrated into the model for transit dosimetry simulations.

**Results:** Simulations of the standard EPID point spread function demonstrated noticeable signal blurring from optical scattering within the phosphor screen. This blurring was not noticeable in open field beam profiles at the standard EPID pixel size of 0.4 mm. Characterization of optical transport in the novel EPID is currently under investigation.

**Conclusions:** Explicit modelling of optical transport may be important when using MC simulations to predict EPID imaging performance. Electromagnetic transport alone may, however, suffice when predicting EPID dose response to open fields. Further work is required to evaluate optical transport effects in the novel EPID configuration.
Characterising optical photon transport in novel electronic portal imaging devices employing plastic scintillator

S. Blake, A. L. McNamara, P. Vial, L. Holloway, P. Greer and Z. Kuncic

Oral presentation at the Geant4 2013 International Users Conference in Bordeaux, France – October, 2013

**Background:** Amorphous silicon (a-Si) electronic portal imaging devices (EPIDs) are x-ray detectors frequently used in radiotherapy for patient imaging. Interest in using EPIDs for dosimetry is growing, however, due to their high 2d spatial resolution and real-time readout capabilities[1]. Standard EPIDs use a copper plate and gadolinium oxysulfide phosphor screen to convert x-rays into optical photons that are detected with an array of a-Si photodiodes. While this configuration results in high x-ray detection efficiency, the high atomic number materials cause a non water-equivalent response that is not ideal for dosimetry[1,2]. A novel configuration that replaces the copper and phosphor screen with a 2d array of water-equivalent plastic scintillator (PS) fibres is under ongoing investigation by our group[2]. This study uses Monte Carlo (MC) simulations to characterise optical transport in this prototype EPID and model its dosimetric response to megavoltage (MV) photon beams.

**Methods:** The GEANT4 MC toolkit[3] was previously used to develop and experimentally validate a model of a 6 MV clinical photon beam and standard a-Si EPID[4]. This model was modified by replacing the components above the photodiode layer with two separate PS geometries. First, a simple PS block with variable thickness was used with the x-ray source to calculate EPID dose profiles. Optical transport in the PS was characterised by varying optical transport parameters and observing any resulting effects on the profiles. Reference scattering ($\mu$) and absorption ($l$) lengths were taken to be 2 cm and 3.5 m respectively from manufacturer specifications. Second, a realistic model of the PS fibre array was developed to simulate the response of the prototype EPID. The array consisted of $150 \times 150$ optically isolated fibres, each measuring $1 \times 1 \text{mm}^2$ in area and 15
mm in length.

**Results:** Dose profiles simulated for select values of \( \mu \) and \( l \) (15 mm thick PS block and \( 5 \times 5 \text{ cm}^2 \) x-ray beam) are shown in Figure 1. A profile simulated using the PS fibre array with reference \( \mu \) and \( l \) is shown for comparison. The percentage difference between each PS block profile and the reference profile is also shown. Since \( l \) is much greater than the PS thickness, varying \( l \) did not affect profile shapes. As \( \mu \) is on the order of the PS thickness, profiles were sensitive to changes in this parameter. Increasing \( \mu \) acted to broaden dose profiles. This effect was greater for thicker scintillators and larger field sizes. The PS fibre array drastically reduced lateral optical scatter, resulting in a flatter profile.

![Figure 1: (Upper) Relative dose profiles simulated for select values of \( \mu \) and \( \tau \). (Lower) Percentage difference relative to reference profile with \( \mu = 2 \text{ cm} \) and \( \tau = 3.5 \text{ m} \).](image)

**Conclusions:** GEANT4 has been used to develop a MC model of a novel EPID incorporating PS detectors. A preliminary characterisation of optical transport in a PS block demonstrates dose profile sensitivity to variations in the optical scattering length. Ongoing work involves validation of the optical transport parameters and the EPID model’s dosimetric response against experimental measurements. Simulations and measurements of the EPID line spread function will also be used for further model validation.
References:


Towards a next-generation electronic portal device for simultaneous imaging and dose verification in radiotherapy

S. Blake, P. Vial, L. Holloway and Z. Kuncic

Oral presentation at The 2014 American Association of Physicists in Medicine (AAPM) annual meeting in Austin TX, USA – July 2014

Purpose: This work forms part of an ongoing study to develop a next-generation electronic portal imaging device (EPID) for simultaneous imaging and dose verification in radiotherapy. Monte Carlo (MC) simulations were used to characterize the imaging performance of a novel EPID that has previously been demonstrated to exhibit a water-equivalent response. The EPID’s response was quantified in several configurations and model parameters were empirically validated against experimental measurements.

Methods: A MC model of a novel a-Si EPID incorporating an array of plastic scintillating fibers was developed. Square BCF-99-06A scintillator fibers with PMMA cladding (Saint-Gobain Crystals) were modelled in a matrix with total area measuring $150 \times 150\,\text{mm}^2$. The standard electromagnetic and optical physics GEANT4 classes were used to simulate radiation transport from an angled slit source (6 MV energy spectrum) through the EPID and optical photons reaching the photodiodes were scored. The prototype’s modulation transfer function (MTF) was simulated and validated against experimental measurements. Several optical transport parameters, fiber lengths and thicknesses of an air gap between the scintillator and photodiodes were investigated to quantify their effects on the prototype’s detection efficiency, sensitivity and MTF.

Results: Simulated EPID response was more sensitive to variations in geometry than in the optical parameters studied. The MTF was particularly sensitive to the introduction of a 0.5 – 1.0 mm air gap between the scintillator and photodiodes, which lowered the MTF relative to that simulated without the gap. As expected, increasing the fiber length increased the detector efficiency and sensitivity while decreasing the MTF.
Conclusions: A model of a novel water-equivalent EPID has been developed and benchmarked against measurements using a physical prototype. We have demonstrated the feasibility of this new device and are continuing to optimize the design to achieve an imaging response that warrants the development of a next-generation prototype.
A next-generation EPID for simultaneous imaging and dosimetry in radiotherapy

S. Blake, P. Vial, L. Holloway and Z. Kuncic

Oral presentation at EPI2k14 - The 13th International Conference on Electronic Patient Imaging in Aarhus, Denmark – September 2014

Summary: A novel prototype electronic portal imaging device (EPID) has been characterised experimentally for imaging and dosimetry applications in radiotherapy. A Monte Carlo (MC) model of the prototype has also been developed for ongoing detector optimisation. The prototype EPID exhibits a water-equivalent dose response and simulations of the modulation transfer function have identified geometrical and optical transport parameters that are important for maintaining spatial resolution.

Introduction: A novel EPID with a water-equivalent dose response has been designed for simultaneous imaging and dose verification applications in radiotherapy. This work presents an updated prototype based on a previously characterised first-generation water-equivalent detector[1]. The first measurements taken with this prototype are reported along with a preliminary characterisation of the detector performance based on MC simulations using a model of the detector prototype. Measurements were used both to experimentally characterise the prototype’s imaging and dosimetric response and to empirically validate the MC model, which will be used for ongoing detector optimisation.

Methods and Materials: To facilitate a water-equivalent dose response and thereby render the prototype EPID more suitable for applications in dosimetry, a plastic scintillating fibre array (PSA) was used in place of the metal plate/phosphor screen in standard amorphous silicon (a-Si) EPIDs. The MC model, developed using GEANT4, is based on the prototype’s design and incorporates an array of $0.5 \times 0.5 \times 30 \text{ mm}^3$ (width $\times$ height $\times$ length) scintillating fibres with total area measuring $150 \times 150 \text{ mm}^2$ (Figure 2). This array was placed in direct contact with the a-Si photodiode panel, which has a nominal pixel pitch of 0.4 mm (the impact of fibre and a-Si pixel mismatch was investigated using...
Experiments were performed using a clinical 6 MV photon beam to measure the prototype PSA-EPID’s modulation transfer function (MTF), field size output factors and relative dose profiles for static, open fields. These quantities were likewise simulated using the MC model. The fibre cross-sectional area, extra-mural absorber (EMA) and optical transport parameters were varied in the MC model to investigate their impact on detector performance.

**Figure 2:** (a) Photo of the second-generation prototype array of plastic scintillating fibres. The individual fibre components and dimensions are illustrated in the schematic, (b).

**Results:** Field size response and relative dose profiles demonstrated a water-equivalent PSA-EPID dose response (Figure 3). Measured and simulated field size response and profiles agreed within statistical and experimental uncertainties. Dose profiles exhibited quasi-periodic variations that resulted from element-to-element mismatch between the scintillating fibres and underlying photodiode pixels. This mismatch also reduced the MTF for high spatial frequencies. When the MC model was used to precisely match scintillating fibres to the underlying pixels, the quasi-periodic variation in profile response was eliminated and the MTF was increased for high spatial frequencies. Removal of the EMA surrounding each scintillating fibre caused a severe decrease in the simulated MTF, highlighting the importance of minimising optical cross talk between fibres in maintaining spatial resolution. Of the optical transport parameters investigated, only the Rayleigh scattering length had a quantifiable impact on the simulated MTF.
MTF, where increases in this parameter over the range $1 - 50 \text{ mm}$ caused a slight increased in the MTF.

![Field size output factors measured using a water-equivalent MatriXX array of ion chambers with solid water buildup are compared to those measured and simulated using the PSA-EPID prototype and MC model.](image)

Figure 3: Field size output factors measured using a water-equivalent MatriXX array of ion chambers with solid water buildup are compared to those measured and simulated using the PSA-EPID prototype and MC model.

**Discussion:** The dose response exhibited by this prototype PSA-EPID gives support that a water-equivalent EPID may be developed to be better suited for clinical dosimetry applications than current commercially available portal imagers. Since its value is not precisely known for plastic scintillator, the Rayleigh scattering length may be treated as a “free” parameter to empirically tune simulated MTFs to measurements. A recent study by El-Mohri *et al.* has proposed novel binning algorithms to correct element-to-element mismatch artefacts and the application of these methods to our prototype is currently under investigation[2].

**Conclusion:** A novel water-equivalent EPID has been characterised experimentally for imaging and dosimetry applications in radiotherapy. A MC model has also been validated and simulation results suggest that further improvements in detector performance may be realised through ongoing detector optimisation.

**Acknowledgements:** The authors acknowledge funding from the Cancer Institute NSW (Research Equipment Grant 10/REG/1-20) and Cancer Council NSW (Grant ID RG 11-06). SJB also thanks The University of Sydney, the Institute of Medical Physics and the Liverpool and Macarthur Cancer Therapy
Centers for scholarship support.

References:
