KILOVOLTAGE RANGE X-RAY CHARACTERISATION AND DIAGNOSTIC APPLICATIONS OF THE MOSKIN DOSIMETER

Nathan Kenneth Thorpe

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KILOVOLTAGE RANGE X-RAY CHARACTERISATION AND DIAGNOSTIC APPLICATIONS OF THE MOSKIN DOSIMETER

A Thesis Submitted in Partial Fulfilment of the Requirements for the Award of the Degree of Doctor of Philosophy

from

UNIVERSITY OF WOLLONGONG

by

Nathan Kenneth Thorpe
BMedRad (Hons)

School of Physics
Faculty of Engineering and Information Sciences

2020
CERTIFICATION

I, Nathan Kenneth Thorpe, declare that this thesis, submitted in partial fulfilment of the requirements for the award of Doctor of Philosophy, in the School of Physics, Faculty of Engineering and Information Sciences, University of Wollongong, is wholly my own work unless otherwise referenced or acknowledged. The document has not been submitted for qualifications at any other academic institution.

Nathan Kenneth Thorpe
25 March 2020
Dedicated to

My family who have supported me throughout this chapter of my life
and to my cat for providing moral support late into many nights
# Table of Contents

<table>
<thead>
<tr>
<th>Section</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>List of Tables</td>
<td>x</td>
</tr>
<tr>
<td>List of Figures/Illustrations</td>
<td>xvi</td>
</tr>
<tr>
<td>ABSTRACT</td>
<td>xx</td>
</tr>
<tr>
<td>Acknowledgements</td>
<td>xxi</td>
</tr>
<tr>
<td><strong>1 Radiation Induced Risk in Radiology</strong></td>
<td>1</td>
</tr>
<tr>
<td>1.1 Interventional Cardiology Procedures</td>
<td>5</td>
</tr>
<tr>
<td>1.2 The Risks of Radiation-induced Injury during Interventional Procedures</td>
<td>9</td>
</tr>
<tr>
<td>1.2.1 Classifications of Radiation-Induced Injury</td>
<td>9</td>
</tr>
<tr>
<td>1.2.2 Radiation-induced injuries in the Catheterisation Lab</td>
<td>11</td>
</tr>
<tr>
<td>1.2.3 Sources of Radiation Exposure in Angiography</td>
<td>15</td>
</tr>
<tr>
<td>1.3 Radiation Safety and Exposure Limitations</td>
<td>16</td>
</tr>
<tr>
<td>1.3.1 Measurement and Assessment of Patient Dose</td>
<td>16</td>
</tr>
<tr>
<td>1.3.2 Limitations on Occupational Exposures</td>
<td>18</td>
</tr>
<tr>
<td>1.4 Alternative measures of radiation exposure</td>
<td>20</td>
</tr>
<tr>
<td>1.4.1 Radiochromic Film Dosimeters</td>
<td>20</td>
</tr>
<tr>
<td>1.4.2 Indirect Dosimetry Solutions</td>
<td>21</td>
</tr>
<tr>
<td>1.4.3 Dose contouring software packages</td>
<td>22</td>
</tr>
<tr>
<td>1.4.4 Solid-State Dosimetry: Band Theory Basics</td>
<td>23</td>
</tr>
<tr>
<td>1.4.5 Luminescent Dosimeters</td>
<td>25</td>
</tr>
<tr>
<td>1.4.6 Electronic Dosimeters</td>
<td>28</td>
</tr>
<tr>
<td>1.4.7 MOSFET Dosimeters</td>
<td>29</td>
</tr>
<tr>
<td>1.5 Why aren’t direct dosimetry solutions broadly deployed during interventional procedures? The Potential of the MO\Skin dosimetry solution</td>
<td>31</td>
</tr>
<tr>
<td><strong>2 Development of Real-time Dosimetry Solutions for Application in the Coronary Catheterisation Laboratory</strong></td>
<td>33</td>
</tr>
<tr>
<td>2.1 Introduction to MOSFET dosimetry solutions</td>
<td>34</td>
</tr>
<tr>
<td>2.1.1 Solid-State Dosimetry: Development of electronic dosimeters</td>
<td>34</td>
</tr>
<tr>
<td>2.1.2 MOSFET-type Radiation Dosimeters</td>
<td>35</td>
</tr>
<tr>
<td>2.1.3 The MO\Skin Dosimeter</td>
<td>39</td>
</tr>
<tr>
<td>2.1.4 The MO\Skin Readout System: Clinical Semiconductor Dosimetry System</td>
<td>40</td>
</tr>
<tr>
<td>2.1.5 The MO\Skin Readout System: OneTouch Dosimetry System</td>
<td>42</td>
</tr>
</tbody>
</table>
# TABLE OF CONTENTS

2.2 Methodologies for Diagnostic Characterisation of the MOSkin Dosimeter

2.2.1 Lifetime Sensitivity of the MOSkin Dosimeter ..... 43
2.2.2 MOSkin Characterisation Using Clinical C-arm Beam Qualities ..... 44
2.2.3 Simulated MOSkin Energy Dependence for Diagnostic X-Ray Spectra ..... 51
2.2.4 MOSkin Energy and Angular Dependence Using Standard Diagnostic Beam Qualities ..... 52

2.3 Results of the MOSkin Characterisation Study

2.3.1 Lifetime Linearity of MOSkin Sensitivity ..... 56
2.3.2 Investigation of MOSkin response to clinical c-arm beam qualities ..... 57
2.3.3 Simulated MOSkin Energy Dependence for Diagnostic X-ray Spectra ..... 59
2.3.4 MOSkin Energy Dependence Using Standard Diagnostic Beam Qualities ..... 67
2.3.5 MOSkin response to varying angle of irradiation incidence using standard diagnostic beam qualities ..... 68

2.4 MOSkin Characterisation Discussion ..... 69

2.5 Conclusions ..... 77

3 Development of Dose Minimisation Strategies via the Optimisation of Operator Modifiable Parameters Selected During Procedure

3.1 Clinical Equipment, Patient Representative Phantom Volumes, defining ‘Dose’ Delivery and Identifying Operator Modifiable Parameters ..... 79
3.1.1 Measuring ‘Dose’ in the Coronary Catheterisation Laboratory ..... 80
3.1.2 Indexation of Operator Modifiable Parameters ..... 81

3.2 Specific Methodologies for Monitoring Each Operator Modifiable Parameter

3.2.1 Field of View ..... 83
3.2.2 Beam Collimation ..... 83
3.2.3 Wedge Filtration ..... 85
3.2.4 Source to Image-receptor Distance, Table Height and Air Gap ..... 87
3.2.5 Phantom Thickness ..... 89
3.2.6 C-arm Angulation using the CIRS 702 Atom Phantom ..... 91

3.3 The Impact of Operator Modifiable Parameters on Dose Delivery

3.3.1 The Impact of Field of View on Dose Delivery ..... 94
3.3.2 The Impact of Beam Collimation on Dose Delivery ..... 94
3.3.3 The Impact of Wedge Filtration on Dose Delivery ..... 94
3.3.4 The Impact of Source to Image-receptor Distance, Table Height and Air Gap on Dose Delivery ..... 96
3.3.5 The Impact of Phantom Thickness on Dose Delivery ..... 97
3.3.6 The Impact of c-arm Angulation using the CIRS 702 Atom Phantom ..... 100

3.4 Relative Importance of Operator Modifiable Parameters ..... 102
### TABLE OF CONTENTS

3.4.1 Summary of Experimental Data and Subsequent Clinical Recommendations .................................................. 102
3.4.2 Impact of Field Related Operator Modifiable Parameters ........................................................................... 104
3.4.3 Impact of C-arm Gantry related Operator Modifiable Parameters ................................................................. 106
3.4.4 Impact of Patient Thickness and C-arm angulation ....................................................................................... 108
3.4.5 The Importance of clear terminology ........................................................................................................... 111
3.4.6 Limitations of this study ............................................................................................................................... 112
3.5 Conclusions ...................................................................................................................................................... 114

4 Evaluating C-arm System Performance and the Efficacy of C-arm Systems by Measuring Dose and Image Quality Simultaneously ........................................................................................................................................ 116

4.1 Introduction to Evaluating C-arm System Performance ...................................................................................... 117
4.1.1 Defining C-arm System Performance through Image Quality ...................................................................... 117
4.1.2 Factors that Influence C-arm System Performance ..................................................................................... 118

4.2 Simultaneous Measurement of Dose Delivery and Image Quality .................................................................. 118
4.2.1 The NEMA XR 21 Protocol ......................................................................................................................... 118
4.2.2 CIRS Model 901 Phantom Configurations ................................................................................................. 120
4.2.3 Quantifying Image Quality: Iodine Contrast ............................................................................................... 121
4.2.4 Quantifying Image Quality: Spatial Resolution ......................................................................................... 122
4.2.5 Quantifying Image Quality: Working Thickness Range ................................................................................ 123
4.2.6 Quantifying Image Quality: Static and Dynamic Motion Target Resolution ................................................. 124
4.2.7 Quantifying Image Quality: Digital Assessment ........................................................................................... 125
4.2.8 Comparison of Image Quality Scores Between Protocols ........................................................................... 126

4.3 C-arm Specific Methodologies .......................................................................................................................... 127
4.3.1 The Philips ‘Clarity’ Upgrade ........................................................................................................................ 127
4.3.2 The GE Innova 2100IQ c-arm system and the GE ‘Dose Map’ Upgrade ......................................................... 130
4.3.3 Comparing the GE Innova 2100IQ and GE IGS 520 c-arm systems ............................................................... 131

4.4 Summary of the Results ..................................................................................................................................... 133
4.4.1 The Philips System Upgrade ........................................................................................................................ 134
4.4.2 The GE System Upgrade ............................................................................................................................. 136
4.4.3 The Upgraded GE System ........................................................................................................................... 138
4.4.4 The GE Intersystem Comparison .................................................................................................................. 140

4.5 Discussion .......................................................................................................................................................... 142
4.5.1 Clinical impact of the Philips upgrade results ............................................................................................... 142
4.5.2 Clinical impact of the GE upgrade results .................................................................................................... 146
4.5.3 Direct system comparison: Innova 2100IQ vs IGS 520 .................................................................................. 150
4.5.4 Developing a better definition of ‘Image Quality’ ......................................................................................... 152
4.5.5 The efficacy of this study’s methodology ..................................................................................................... 154
4.5.6 Air Kerma and the importance of implementing real-time surface dosimetry solutions ............................ 157

4.6 Conclusions ....................................................................................................................................................... 161
## TABLE OF CONTENTS

<table>
<thead>
<tr>
<th>Chapter</th>
<th>Title</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>5</td>
<td>Development of Operator Dosimetry Solutions and Methodologies to</td>
<td>163</td>
</tr>
<tr>
<td></td>
<td>Address Emerging Health Concerns</td>
<td></td>
</tr>
<tr>
<td></td>
<td>5.1 Operator Specific Dosimetry Solutions</td>
<td>164</td>
</tr>
<tr>
<td></td>
<td>5.1.1 Radiation-induced risks to interventional clinicians</td>
<td>164</td>
</tr>
<tr>
<td></td>
<td>5.1.2 Radiation-induced injury to Clinician Eye Lens</td>
<td>165</td>
</tr>
<tr>
<td></td>
<td>5.1.3 Requirements for eye lens specific dose monitoring</td>
<td>166</td>
</tr>
<tr>
<td></td>
<td>5.2 Methodology</td>
<td>170</td>
</tr>
<tr>
<td></td>
<td>5.2.1 Calibration of the TLD-100 Eye-D Dosimeter</td>
<td>170</td>
</tr>
<tr>
<td></td>
<td>5.2.2 Measuring Eye Lens Dose from Several Recommended Reference Points</td>
<td>172</td>
</tr>
<tr>
<td></td>
<td>5.3 Results</td>
<td>174</td>
</tr>
<tr>
<td></td>
<td>5.3.1 TLD Eye-D calibration values</td>
<td>176</td>
</tr>
<tr>
<td></td>
<td>5.3.2 Comparison of Eye Lens Dose Measurement Points</td>
<td>179</td>
</tr>
<tr>
<td></td>
<td>5.3.3 H_{\mu}(3) MOSkin Calibrations</td>
<td>182</td>
</tr>
<tr>
<td></td>
<td>5.4 Discussion</td>
<td>183</td>
</tr>
<tr>
<td></td>
<td>5.4.1 The Accuracy of the TLD Eye-D Calibration</td>
<td>183</td>
</tr>
<tr>
<td></td>
<td>5.4.2 Operator Phantom Eye Lens Doses</td>
<td>183</td>
</tr>
<tr>
<td></td>
<td>5.4.3 Performance of the H_{\mu}(3) MOSkin enclosure</td>
<td>185</td>
</tr>
<tr>
<td></td>
<td>5.4.4 Consideration for Future Operator Dosimetry MOSkin Design</td>
<td>187</td>
</tr>
<tr>
<td></td>
<td>5.5 Conclusions</td>
<td>187</td>
</tr>
<tr>
<td>6</td>
<td>Floating Gate MOSFET Dosimeters: A Prospective Study</td>
<td>188</td>
</tr>
<tr>
<td></td>
<td>6.1 Overview of Floating Gate MOSFET devices</td>
<td>188</td>
</tr>
<tr>
<td></td>
<td>6.2 FG-MOSPET Characterisation methodologies</td>
<td>191</td>
</tr>
<tr>
<td></td>
<td>6.2.1 FG-MOSPET Characterisation: Output Stability and Temperature Dependence</td>
<td>192</td>
</tr>
<tr>
<td></td>
<td>6.2.2 FG-MOSPET Characterisation: Mega-Voltage range Gamma Characterisation</td>
<td>193</td>
</tr>
<tr>
<td></td>
<td>6.2.3 FG-MOSPET Characterisation: Kilovoltage Range X-ray Characterisation</td>
<td>194</td>
</tr>
<tr>
<td></td>
<td>6.3 Results of the FG-MOSPET Characterisation</td>
<td>195</td>
</tr>
<tr>
<td></td>
<td>6.3.1 Output Stability and Temperature Dependence</td>
<td>195</td>
</tr>
<tr>
<td></td>
<td>6.3.2 Megavoltage Range Gamma-ray Sensitivity</td>
<td>197</td>
</tr>
<tr>
<td></td>
<td>6.3.3 Kilovoltage Range X-ray Sensitivity</td>
<td>200</td>
</tr>
<tr>
<td></td>
<td>6.4 Discussion of FG-MOSPET Performance</td>
<td>201</td>
</tr>
<tr>
<td></td>
<td>6.4.1 FG-MOSPET Output Stability</td>
<td>201</td>
</tr>
<tr>
<td></td>
<td>6.4.2 FG-MOSPET Sensitivity</td>
<td>201</td>
</tr>
<tr>
<td></td>
<td>6.4.3 FG-MOSPET Desensitisation</td>
<td>202</td>
</tr>
<tr>
<td></td>
<td>6.5 FG-MOSPET Conclusions</td>
<td>204</td>
</tr>
<tr>
<td>7</td>
<td>Conclusions and the Future of Diagnostic Dosimetry Solutions</td>
<td>206</td>
</tr>
</tbody>
</table>
# TABLE OF CONTENTS

## A Philips ‘Clarity’ Upgrade Evaluation

- **A.0.1** Overview of the Clarity Upgrade: Acquisition Mode .......................... 210
- **A.0.2** Comparison of the high detail acquisition mode protocols: ‘Clarity Boost’ and ‘Cardiac 4’ ................................................................. 211
- **A.0.3** Comparison of the high detail acquisition mode protocols: ‘Clarity Normal’ and ‘Cardiac 4’ ................................................................. 213
- **A.0.4** Comparison of low dose acquisition mode protocols: ‘Cardiac Low’ and ‘Clarity Low’ ................................................................. 215
- **A.0.5** Overview of the Clarity Upgrade: Fluoroscopy mode protocols . 217
- **A.0.6** Comparison of the fluoroscopy mode dose level 1 protocols: ‘Clarity’ and ‘Cardiac 4’ ................................................................. 218
- **A.0.7** Comparison of the fluoroscopy mode dose level 2 protocols: The ‘Clarity’ and ‘Cardiac 4’ protocols ................................................................. 220
- **A.0.8** Comparison of the fluoroscopy mode dose level 1 protocols: The ‘Clarity’ and ‘Cardiac Low’ protocols ................................................................. 223
- **A.0.9** Comparison of the low fluoroscopy mode dose level 2 protocols: The ‘Clarity’ and ‘Cardiac Low’ protocols ................................................................. 226

## B GE ‘Dose Map’ Upgrade Evaluation

- **B.0.1** Overview of the GE Upgrade Evaluation ........................................... 230
- **B.0.2** Comparison of the high detail acquisition mode dose level 1 protocols: ‘Standard’ and ‘CoroPlus’ ................................................................. 231
- **B.0.3** Comparison of the low dose rate acquisition mode dose level 1 protocols: ‘Low’ and ‘RDLS’ ................................................................. 233
- **B.0.4** Comparison of the post-upgrade acquisition mode dose level 1 protocols: ‘Low’ and ‘Standard’ ................................................................. 236
- **B.0.5** Comparison of the post-upgrade acquisition mode dose level 1 protocols: ‘Very Low’ and ‘Low’ ................................................................. 239
- **B.0.6** Comparison of the SHC and EHC high detail acquisition mode dose level 1 protocols: ‘Standard’ and ‘EHC-IQ+’ ................................................................. 242
- **B.0.7** Comparison of the SHC and EHC low dose rate acquisition mode dose level 1 protocols: ‘Low’ and ‘EHC-RDLS’ ................................................................. 244
- **B.0.8** Comparison of the post-upgrade acquisition mode dose level 2 protocols: ‘Standard’ and ‘CoroPlus’ ................................................................. 247
- **B.0.9** Comparison of the post-upgrade acquisition mode dose level 2 protocols: ‘Low’ and ‘RDLS’ ................................................................. 249
- **B.0.10** Comparison of the post-upgrade acquisition mode dose level 2 protocols: ‘Low’ and ‘Standard’ ................................................................. 251
- **B.0.11** Comparison of the post-upgrade acquisition mode dose level 2 protocols: ‘Very Low’ and ‘Low’ ................................................................. 253
- **B.0.12** Comparison of the SHC and EHC high detail acquisition mode dose level 2 protocols: ‘Standard’ and ‘EHC-IQ+’ ................................................................. 256
- **B.0.13** Comparison of the SHC and EHC low dose rate acquisition mode dose level 2 protocols: ‘Low’ and ‘EHC-RDLS’ ................................................................. 258
<table>
<thead>
<tr>
<th>Section</th>
<th>Title</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>B.0.14</td>
<td>Comparison of the post-upgrade fluoroscopy mode dose level 1 protocols: ‘Standard’ and ‘CoroPlus’</td>
<td>261</td>
</tr>
<tr>
<td>B.0.15</td>
<td>Comparison of the post-upgrade fluoroscopy mode dose level 1 protocols: ‘Low’ and ‘RDLs’</td>
<td>264</td>
</tr>
<tr>
<td>B.0.16</td>
<td>Comparison of the post-upgrade fluoroscopy mode dose level 1 protocols: ‘Low’ and ‘Standard’</td>
<td>267</td>
</tr>
<tr>
<td>B.0.17</td>
<td>Comparison of the post-upgrade fluoroscopy mode dose level 1 protocols: ‘Low (15 FPS)’ and ‘Low’</td>
<td>270</td>
</tr>
<tr>
<td>B.0.18</td>
<td>Comparison of the post-upgrade fluoroscopy mode dose level 1 protocols: ‘Very Low’ and ‘Low’</td>
<td>273</td>
</tr>
<tr>
<td>B.0.19</td>
<td>Comparison of the post-upgrade acquisition mode dose level 2 protocols: ‘Low’ and ‘Standard’</td>
<td>275</td>
</tr>
<tr>
<td>B.0.20</td>
<td>Comparison of the post-upgrade acquisition mode dose level 2 protocols: ‘Low (15 FPS)’ and ‘Low’</td>
<td>277</td>
</tr>
<tr>
<td>B.0.21</td>
<td>Comparison of the post-upgrade acquisition mode dose level 2 protocols: ‘Very Low’ and ‘Low’</td>
<td>280</td>
</tr>
</tbody>
</table>
## List of Tables

<table>
<thead>
<tr>
<th>Table</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>2.1</td>
<td>Comparison of the Beam Area produced at Image Receptor by Diagonal Field Settings and Orthogonal Field Settings</td>
<td>48</td>
</tr>
<tr>
<td>2.2</td>
<td>Narrow Spectrum Series Beam Qualities as specified by ISO 4037-1 [111]</td>
<td>49</td>
</tr>
<tr>
<td>2.3</td>
<td>RQR Spectrum Series Beam Qualities as specified by IEC 61267 [112]</td>
<td>50</td>
</tr>
<tr>
<td>2.4</td>
<td>Calibration of the MOSkin dosimeter using the Philips Allura Xper FD20 c-arm system in acquisition mode</td>
<td>60</td>
</tr>
<tr>
<td>2.5</td>
<td>Calibration of the MOSkin dosimeter using the Philips Allura Xper FD20 c-arm system in fluoroscopy mode</td>
<td>60</td>
</tr>
<tr>
<td>2.6</td>
<td>Calibration of the MOSkin dosimeter using the GE Innova 2100IQ c-arm system in acquisition mode</td>
<td>61</td>
</tr>
<tr>
<td>2.7</td>
<td>Calibration of the MOSkin dosimeter using the GE Innova 2100IQ c-arm system in fluoroscopy mode</td>
<td>61</td>
</tr>
<tr>
<td>2.8</td>
<td>Coefficients for quadratic trendlines that describe MOSkin behaviour during exposure to kilovoltage range beam qualities over four filtration values</td>
<td>66</td>
</tr>
<tr>
<td>2.9</td>
<td>Experimental and Published MOSkin calibration factors for radiotherapeutic megavoltage beam qualities</td>
<td>70</td>
</tr>
<tr>
<td>2.10</td>
<td>Experimental and Published MOSkin calibration factors for diagnostic kilovoltage beam qualities</td>
<td>70</td>
</tr>
<tr>
<td>3.1</td>
<td>Averaged Impact on Dose Delivery from Manipulation of Operator Modifiable Parameters</td>
<td>102</td>
</tr>
<tr>
<td>4.1</td>
<td>Air cylinder and aluminium cylinder thicknesses embedded within each quadrant of the CIRS model 901 phantom for c-arm WTR assessment</td>
<td>124</td>
</tr>
<tr>
<td>4.2</td>
<td>Pre- and post- upgrade protocols installed on the Philips c-arm system commissioned at Sutherland Heart Clinic</td>
<td>128</td>
</tr>
<tr>
<td>4.3</td>
<td>Pre- and post- upgrade protocols installed on the GE c-arm systems commissioned at Sutherland Heart Clinic and Eastern Heart Clinic</td>
<td>132</td>
</tr>
<tr>
<td>4.4</td>
<td>Summary of the change in Philips c-arm system performance after the Clarity upgrade</td>
<td>135</td>
</tr>
<tr>
<td>4.5</td>
<td>Summary of the change in GE Innova 2100IQ c-arm system performance after the post-processing upgrade</td>
<td>137</td>
</tr>
<tr>
<td>4.6</td>
<td>Summary of the upgraded GE Innova 2100IQ protocol comparison</td>
<td>139</td>
</tr>
</tbody>
</table>
### LIST OF TABLES

<table>
<thead>
<tr>
<th>Table</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>4.7</td>
<td>Summary of the comparison of the GE Innova 2100\textsuperscript{IQ} and GE IGS 520 c-arm systems</td>
<td>141</td>
</tr>
<tr>
<td>4.8</td>
<td>Air Kerma to ESD ratio as recorded for each acquisition mode measurement performed during the c-arm system performance study</td>
<td>159</td>
</tr>
<tr>
<td>4.9</td>
<td>Air Kerma to ESD ratio as recorded for each fluoroscopy mode measurement performed during the c-arm system performance study</td>
<td>160</td>
</tr>
<tr>
<td>6.1</td>
<td>FG-MOSFET Sensitivities for each 10 Gy Irradiation Period</td>
<td>199</td>
</tr>
<tr>
<td>6.2</td>
<td>Comparison of FG-MOSFET characteristics to MOSkin Dosimeter</td>
<td>203</td>
</tr>
<tr>
<td>A.1</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Cardiac 4’ and ‘Clarity Boost’ acquisition mode protocols for all phantom and imaging configurations</td>
<td>212</td>
</tr>
<tr>
<td>A.2</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity Normal’ and ‘Cardiac 4’ acquisition mode protocols for all phantom and imaging configurations</td>
<td>214</td>
</tr>
<tr>
<td>A.3</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity Low’ and ‘Cardiac Low’ acquisition mode protocols for all phantom and imaging configurations</td>
<td>216</td>
</tr>
<tr>
<td>A.4</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac 4’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations</td>
<td>219</td>
</tr>
<tr>
<td>A.5</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac 4’ fluoroscopy mode protocols in the dose level 2 setting for all phantom and imaging configurations</td>
<td>222</td>
</tr>
<tr>
<td>A.6</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac Low’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations</td>
<td>225</td>
</tr>
<tr>
<td>A.7</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac Low’ fluoroscopy mode protocols in the dose level 2 setting for all phantom and imaging configurations</td>
<td>228</td>
</tr>
<tr>
<td>B.1</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Standard’ and ‘CoroPlus’ acquisition mode protocols in the dose level 1 setting for all phantom and imaging configurations</td>
<td>232</td>
</tr>
<tr>
<td>B.2</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘RDLS’ acquisition mode protocols in the dose level 1 setting for all phantom and imaging configurations</td>
<td>235</td>
</tr>
<tr>
<td>B.3</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘Standard’ acquisition mode protocols in the dose level 1 setting for all phantom and imaging configurations</td>
<td>238</td>
</tr>
<tr>
<td>B.4</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Very Low’ and ‘Low’ acquisition mode protocols in the dose level 1 setting for all phantom and imaging configurations</td>
<td>241</td>
</tr>
<tr>
<td>B.5</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Standard’ and ‘EHC-IQ+’ acquisition mode protocols in the dose level 1 setting for all phantom and imaging configurations.</td>
<td>243</td>
</tr>
<tr>
<td>B.6</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘EHC-RDLS’ acquisition mode protocols in the dose level 1 setting for all phantom and imaging configurations.</td>
<td>246</td>
</tr>
<tr>
<td>B.7</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Standard’ and ‘CoroPlus’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.</td>
<td>248</td>
</tr>
<tr>
<td>B.8</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘RDLS’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.</td>
<td>250</td>
</tr>
<tr>
<td>B.9</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘Standard’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.</td>
<td>252</td>
</tr>
<tr>
<td>B.10</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Very Low’ and ‘Low’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.</td>
<td>255</td>
</tr>
<tr>
<td>B.11</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Standard’ and ‘EHC-IQ+’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.</td>
<td>257</td>
</tr>
<tr>
<td>B.12</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘EHC-RDLS’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.</td>
<td>260</td>
</tr>
<tr>
<td>B.13</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Standard’ and ‘CoroPlus’ and fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.</td>
<td>263</td>
</tr>
<tr>
<td>B.14</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘RDLS’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.</td>
<td>266</td>
</tr>
<tr>
<td>B.15</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘Standard’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.</td>
<td>269</td>
</tr>
<tr>
<td>B.16</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘Low (15 FPS)’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.</td>
<td>272</td>
</tr>
<tr>
<td>B.17</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Very Low’ and ‘Low’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.</td>
<td>274</td>
</tr>
<tr>
<td>B.18</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘Standard’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.</td>
<td>276</td>
</tr>
<tr>
<td>Table</td>
<td>Description</td>
<td></td>
</tr>
<tr>
<td>-------</td>
<td>-------------</td>
<td></td>
</tr>
<tr>
<td>B.19</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Low (15 FPS)’ and ‘Low’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations</td>
<td>279</td>
</tr>
<tr>
<td>B.20</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Very Low’ and ‘Low’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations</td>
<td>281</td>
</tr>
</tbody>
</table>
# List of Figures

<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.1</td>
<td>Percentage Share and Total of Revascularisation Procedures Performed in Australia (2000-2014)</td>
<td>6</td>
</tr>
<tr>
<td>1.2</td>
<td>Trends in Revascularisation Procedural Preference in OECD Countries (2000-2013)</td>
<td>6</td>
</tr>
<tr>
<td>1.3</td>
<td>Illustration of the Kerma Area Product</td>
<td>17</td>
</tr>
<tr>
<td>1.4</td>
<td>a) Cardiologist’s Personal Protective Equipment and b) Common Operator Exposure Monitoring Points</td>
<td>19</td>
</tr>
<tr>
<td>1.5</td>
<td>Radiopacity of Traditional Real-Time Dosimetry Systems</td>
<td>32</td>
</tr>
<tr>
<td>2.1</td>
<td>Cross-section of a generic p-channel type MOSFET Dosimeter</td>
<td>38</td>
</tr>
<tr>
<td>2.2</td>
<td>$I_D - V_G$ Characteristics of a MOSFET Dosimeter before/after irradiation</td>
<td>38</td>
</tr>
<tr>
<td>2.3</td>
<td>Photograph of the MO.Skin dosimeter with Clinical Semiconductor Dosimetry System</td>
<td>41</td>
</tr>
<tr>
<td>2.4</td>
<td>Photograph of the MO.Skin dosimeter with the wireless OneTouch System</td>
<td>41</td>
</tr>
<tr>
<td>2.5</td>
<td>PMMA Phantom with MO.Skin attached to underside positioned at the c-arm system rotational isocenter</td>
<td>45</td>
</tr>
<tr>
<td>2.6</td>
<td>Irradiation apparatus for characterisation of the MO.Skin dosimeter using standardised beam qualities using cylindrical PMMA water phantom positioned with measurement point 100 cm from x-ray source</td>
<td>54</td>
</tr>
<tr>
<td>2.7</td>
<td>MO.Skin dosimeter positioning on cylindrical phantom volume presented with a) phantom at 0° orientation and b) phantom at 90° orientation. with respect to the MOSkin pigtail</td>
<td>54</td>
</tr>
<tr>
<td>2.8</td>
<td>Diagram representing the scoring volumes used during air Kerma simulations as positioned in a) free air and b) with the phantom volume generated (yellow represents $H_p(0.07)$, green represents $H_p(3)$, purple represents $H_p(10)$ and grey arrows represent uniform incident radiation beam)</td>
<td>55</td>
</tr>
<tr>
<td>2.9</td>
<td>Irradiation apparatus for measuring angular response of the MOSkin dosimeter to standardised beam qualities at a fixed source to dosimeter distance of 1 meter (blue circle represents phantom in 0° incidence configuration, yellow circle represents phantom in 45° incidence configuration and orange circle represents phantom in 90° incidence configuration)</td>
<td>55</td>
</tr>
<tr>
<td>Figure</td>
<td>Description</td>
<td>Page</td>
</tr>
<tr>
<td>--------</td>
<td>-------------</td>
<td>-----</td>
</tr>
<tr>
<td>2.10</td>
<td>Lifetime Dependence of MOSkin Sensitivity on Accumulated Absorbed Dose using 6MV Linear Accelerator to a Total Dose of 30 Gy</td>
<td>58</td>
</tr>
<tr>
<td>2.11</td>
<td>MOSkin Sensitivity Expressed as a Function of Threshold Voltage</td>
<td>58</td>
</tr>
<tr>
<td>2.12</td>
<td>MOSkin Dose Rate Linearity using a Toshiba Infinix C-arm System producing a consistent beam quality at 93 kVp x-ray tube voltage</td>
<td>63</td>
</tr>
<tr>
<td>2.13</td>
<td>Energy and dose rate dependence of EBT2 film using 50, 70, 100 and 150 kVp orthovoltage beam qualities for doses of up to 200 cGy</td>
<td>63</td>
</tr>
<tr>
<td>2.14</td>
<td>Photon Spectra for x-ray tube potentials ranging from 60 kVp to 120 kVp with (A) no filter, (B) 2mmAl filter, (C) 5mmAl filter and (D) 10mmAl filter settings at a fixed x-ray tube current of 1mAs</td>
<td>64</td>
</tr>
<tr>
<td>2.15</td>
<td>MOSkin response to monoenergetic photons relative to the response at 300 keV as generated using Monte Carlo with logistics trendline fit to data</td>
<td>64</td>
</tr>
<tr>
<td>2.16</td>
<td>Relative MOSkin response to the simulated c-arm beam spectra</td>
<td>65</td>
</tr>
<tr>
<td>2.17</td>
<td>MOSkin calibration factors as established by scaling the relative MOSkin response data as scaled by and compared to clinically measured MOSkin Sensitivities</td>
<td>65</td>
</tr>
<tr>
<td>2.18</td>
<td>Energy dependence of the MOSkin dosimeter at native measurement depth with respect to ( H_p(0.07), H_p(3) ) and ( H_p(10) ) dose delivered to measurement point</td>
<td>67</td>
</tr>
<tr>
<td>2.19</td>
<td>MOSkin response to varying angle of incidence as irradiated using ( a) ) the Narrow( 80 ) beam quality and ( b) ) the RQR( 6 ) beam quality with angles simulated by ( c) ) rotating the cylindrical phantom volume</td>
<td>68</td>
</tr>
<tr>
<td>2.20</td>
<td>MOSkin radiation transparency and visibility as compared to paperclips on images acquired by the Phillips Allura Xper FD20 c-arm system</td>
<td>71</td>
</tr>
<tr>
<td>2.21</td>
<td>MOSkin response to standard beam qualities when varying ( a) ) ( \theta ) axis angulation and ( b) ) ( \Phi ) axis angulation with as simulated by ( c) ) rotating the cylindrical phantom volume</td>
<td>74</td>
</tr>
<tr>
<td>2.22</td>
<td>Comparison of MOSkin response at reference depths ( H_p(0.07), H_p(3) ) and ( H_p(10) ) during ( a) ) Narrow( 80 ) and ( b) ) RQR( 6 ) irradiation when varying the ( \theta ) axis with ( c) ) degree of angular dependence for each reference depth represented in terms of parabolic ( k )-values</td>
<td>76</td>
</tr>
<tr>
<td>3.1</td>
<td>The a) PMMA Slab Phantom and the b) CIRS 702 Atom Phantom</td>
<td>80</td>
</tr>
<tr>
<td>3.2</td>
<td>Diagrammatic representation of variables relevant to calculating cross-sectional area of the c-arm beam</td>
<td>82</td>
</tr>
<tr>
<td>3.3</td>
<td>a) No Collimation b) 73% of Primary Beam Area c) 52% of Primary Beam Area d) 22% of Primary Beam Area e) 9% of Primary Beam Area</td>
<td>84</td>
</tr>
<tr>
<td>3.4</td>
<td>a) acquisition with no additional filters b) acquisition with one filtration wedge applied c) acquisition with two filtration wedges applied separately d) acquisition with two filtration wedges overlaid e) acquisition with two filtration wedges partially overlaid f) acquisition with MOSkin dosimeters superimposed to represent the three dosimetric measurement points utilised</td>
<td>86</td>
</tr>
</tbody>
</table>
3.5 Diagram representing the operator modifiable parameters of SID, Table Height and Air Gap relative to the c-arm gantry ......................................................... 88
3.6 Diagrammatic representation of phantom thickness diminishing by 1 cm with each measurement for EPT ranging from 19-30 cm (PMMA phantom aligned with c-arm rotational isocentre) ........................................ 90
3.7 Diagrammatic representation of the CIRS 702 phantom with heart aligned to the c-arm gantry rotational isocentre and with ESD measure by the MOSkin ................................................................. 92
3.8 CT DICOM of the CIRS 702 Atom Phantom representing the phantom heart aligned to c-arm rotational isocenter presented as a) axial cross-section in plane of measurement points with LAO/RAO measurement points identified and as b) sagittal cross-section in plane of measurement points with CRA/CAU measurement points identified ........................................ 92
3.9 Impact of adjusting the FOV on the a) ESDR and b) DAPR delivered to the 30 × 30 × 30 cm³ PMMA phantom volume by the Philips Allura Xper FD20 c-arm system with linear inversely-correlating linear trend lines applied ......................................................... 95
3.10 Impact of adjusting beam collimation on the a) ESDR and b) DAPR delivered to the 30 × 30 × 30 cm³ PMMA phantom volume by the Philips Allura Xper FD20 c-arm system ................................................................. 95
3.11 Impact of adjusting wedge filtration settings on the a) ESDR and b) DAPR delivered to the 30 × 30 × 30 cm³ PMMA phantom volume by the Philips Allura Xper FD20 c-arm system. (For visual reference to wedge filtration configurations, refer to Figure 3.4) ............................................... 95
3.12 Impact of adjusting the SID on the a) ESDR and b) DAPR delivered to the 30 × 30 × 30 cm³ PMMA phantom volume by the Philips Allura Xper FD20 c-arm system with linear inversely correlating linear trend lines applied ......................................................... 98
3.13 Impact of adjusting table height on the a) ESDR and b) DAPR delivered to the 30 × 30 × 30 cm³ PMMA phantom volume by the Philips Allura Xper FD20 c-arm system with correlating trend lines applied ......................................................... 98
3.14 Impact of adjusting air gap thickness on the a) ESDR and b) DAPR delivered to the 30 × 30 × 30 cm³ PMMA phantom volume by the Philips Allura Xper FD20 c-arm system with inversely correlating trend lines applied ......................................................... 99
3.15 Impact of adjusting table height with constant air gap thickness on the a) ESDR and b) DAPR delivered to the 30 × 30 × 30 cm³ PMMA phantom volume by the Philips Allura Xper FD20 c-arm system with correlating trend lines applied ......................................................... 99
3.16 Impact of adjusting EPT on the dose delivered to the 30 × 30 × 30 cm³ PMMA phantom volume by the a) Philips Allura Xper FD20 and b) the GE Innova 2100IQ c-arm systems with trend lines applied ......................................................... 100
3.17 Impact of adjusting LAO/RAO angulation on the ESDR and DAPR delivered to the CIRS 702 Atom phantom volume by the Philips Allura Xper FD20 c-arm system .......................................................... 101
3.18 Impact of adjusting CRA/CAU angulation on the ESDR and DAPR delivered to the CIRS 702 Atom phantom volume by the Philips Allura Xper FD20 c-arm system .......................................................... 101
3.19 Comparison of experimental angulation data around the a) LAO/RAO axis and the b) CRA/CAU axis[125] .......................................................... 110
4.1 Diagram of the CIRS Model 901 phantom configured in the a) 20 cm EPT configuration and in the b) 30 cm EPT configuration .................. 119
4.2 CIRS Model 901 phantom equipped with the a) image quality target plate Configuration and equipped with the b) motion target plate .... 122
4.3 Depiction of a typical ROI through use of a) a graphical representation of a typical selection area and b) the corresponding sampling distribution of the ROI area .......................................................... 126
4.4 Example of acquisition mode image quality when using the a) Cardiac 4 protocol as compared to the b) Clarity Normal protocol .......... 146
4.5 a) An isolated post-upgrade GE Innova 2100 IQ protocol acquisition frame presenting edge enhancement effects at the boundary of high contrast image features with line profiles provided for a b) low density WTR imaging target and a c) high density WTR imaging target .......................................................... 155
4.6 a) An isolated post-upgrade GE Innova 2100 IQ protocol acquisition frame presenting radial gradient effects as shown graphically through b) line profile .......................................................... 155
5.1 Eye lens dosimeters that have been used clinically include the a) Rad- card Eye-D, b) the Landauer Vision, Hp(10) depth chest monitors (esti- mated) and the MOSkin dosimeter (with active bias applied via battery apparatus) .......................................................... 169
5.2 Photographs of a) Glasses A, b) Glasses B (with side shielding) and c) the Rando Alderson phantom equipped with Glasses type A ........ 173
5.3 Diagrammatic representation of head phantom mounted to the turn table apparatus a) with angle measured marked and b) in the 45° con- figuration .......................................................... 173
5.4 a) Annotated diagram of MOSkin measurement points with b) photo- graph of phantom fitted with MOSkin dosimeters .......................................................... 175
5.5 Hp(3) MOSkin capping material represented schematically a) in isomet- ric view, b) in isometric view (internal borders visible), c) from the side view (cross-section with MOSkin cavity visible), d) from the front view (MOSkin position marked, dimensions marked) and e) as positioned on the phantom volume during characterisation (photograph) ........ 175
5.6 Eye-D response to the a) Narrow Spectrum series and b) RQR series beam qualities available at the SCK-CEN calibration facilities as compared to data published by Bilski et al.\textsuperscript{[165]}

5.7 Eye-D angular response to the Narrow\textsubscript{80} and RQR\textsubscript{6} beam qualities

5.8 Phantom eye lens dose without any personal protective equipment (normalised to 0° angle of incidence)

5.9 H\textsubscript{p}(0.07) and H\textsubscript{p}(3) doses measured at the eye lens reference point with the Rando-Alderson phantom equipped with both a) Glasses A and b) Glasses B with c) comparative graph included

5.10 Comparison of MOSkin response for a) all measurement points and for eye lens measurement point compared to b) the under-glasses measurement point c) the over-glasses measurement point d) the side-glasses measurement point and the e) forehead measurement point for Glasses configuration A

5.11 Comparison of MOSkin response for a) all measurement points and for eye lens measurement point compared to b) the under-glasses measurement point c) the over-glasses measurement point d) the side-glasses measurement point and the e) forehead measurement point for Glasses configuration B

5.12 Angular response of H\textsubscript{p}(3) encapsulated MOSkin to Narrow\textsubscript{80} and RQR\textsubscript{6} beam qualities with respect to the a) θ rotational axis and the b) Φ rotational axis

5.13 Comparison of energy dependence between the native H\textsubscript{p}(0.07) MOSkin dosimeter and the encapsulated H\textsubscript{p}(3) MOSkin dosimeter

6.1 a) Diagram of the FG-MOSFET structure and visual representation of the Fowler-Nordheim tunnelling effect used to b) deposit and c) remove charge from the floating gate structure

6.2 Irradiation apparatus for irradiating the FG-MOSFET using a) a 6 MV linear accelerator and b) an angiographic c-arm system (dose verified with MOSkin)

6.3 a) FG-MOSFET readout stability over 24-hour period and b) the ambient temperature as measured by onboard resistors

6.4 FG-MOSFET readout stability over 1-hour period at constant temperature (T=22.7°C)

6.5 Accumulated FG-MOSFET response to 6MV linear accelerator beam quality (Threshold Voltage reset at 30, 60 and 90 Gy)

6.6 FG-MOSFET sensitivity to 6MV linear accelerator beam quality with respect to accumulated dose delivered

6.7 FG-MOSFET sensitivity to 6MV linear accelerator beam quality with respect to threshold voltage before irradiation

6.8 FG-MOSFET energy dependence with respect to clinical diagnostic x-ray beam qualities
<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>A.1</td>
<td>ESD delivery and image quality scores produced by the Philips Allura Xper and Philips Allura Clarity c-arm systems in acquisition mode using the a) 30 cm EPT, 15 cm FOV DFS configuration, b) 30 cm EPT, 20 cm DFS configuration c) 20 cm EPT, 15 cm DFS configuration and the d) 20 cm EPT, 20 cm DFS configuration.</td>
</tr>
<tr>
<td>A.2</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Cardiac 4’ and ‘Clarity Boost’ acquisition mode protocols for all phantom and imaging configurations.</td>
</tr>
<tr>
<td>A.3</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity Normal’ and ‘Cardiac 4’ acquisition mode protocols for all phantom and imaging configurations.</td>
</tr>
<tr>
<td>A.4</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity Normal’ and ‘Cardiac 4’ acquisition mode protocols for all phantom and imaging configurations.</td>
</tr>
<tr>
<td>A.5</td>
<td>Dose delivery and image quality scores produced by the Philips Allura Xper and Philips Allura Clarity c-arm systems in fluoroscopy modes 1 and 2 using the a) 30 cm phantom configuration and 15 cm DFS setting, b) 30 cm phantom configuration and 20 cm DFS setting c) 20 cm phantom configuration and 15 cm DFS setting and the d) 20 cm phantom configuration and 20 cm DFS setting.</td>
</tr>
<tr>
<td>A.6</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac 4’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.</td>
</tr>
<tr>
<td>A.7</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac 4’ fluoroscopy mode protocols in the dose level 2 setting for all phantom and imaging configurations.</td>
</tr>
<tr>
<td>A.8</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac 4’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.</td>
</tr>
<tr>
<td>A.9</td>
<td>Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac Low’ fluoroscopy mode protocols in the dose level 2 setting for all phantom and imaging configurations.</td>
</tr>
</tbody>
</table>


Thesis Presentations


Kilovoltage Range X-ray Characterisation and Diagnostic Applications of the MOSkin Dosimeter

Nathan Kenneth Thorpe

A Thesis for Doctor of Philosophy
School of Physics
University of Wollongong

ABSTRACT

The use of diagnostic imaging services has been increasing significantly over recent decades as the availability and the sophistication of diagnostic imaging equipment has expanded. This has resulted in improved patient outcome through earlier and more accurate diagnoses at cost of increasing radiation exposure to the general populace. This may result in an increased incidence of radiation-induced conditions and cancer. Monitoring the radiation exposures during these procedures is crucial to preventing radiation induced conditions and minimising radiation exposure both to patients and to operators of radiation imaging technologies where possible. This thesis investigated the applications of the MOSkin dosimeter, in clinical diagnostic procedures. The MOSkin provides advantages over traditional dosimetry solutions by providing minimal beam perturbation, the ability to measure doses in real-time, exhibiting radio-transparency in images acquired by diagnostic beam qualities and by allowing users to measure dose delivered at water-equivalent depths of 0.07 mm and 3 mm, depths representative of radiation induced damage to the patient’s skin and eyes. The aims of this thesis were to comprehensively characterise the dosimeter for use in kilovoltage range x-ray beam qualities, establish clinical applications, specifically in angiographic catheterisation laboratories, to develop dose minimisation strategies in clinic, to assess vendor implemented upgrades in clinic and to identify the specifications needed for developing future diagnostic dosimetry solutions. By addressing these aims, this thesis has produced a compelling argument for broader clinical adoption of diagnostic dosimetry solutions, has confirmed the efficacy of the MOSkin dosimeter in measuring diagnostic patient doses, has enabled the development of more sophisticated dose minimisation strategies in clinics, has improved upon existing standards for comparing c-arm system hardware performance by implementing simultaneous measurement of several metrics for both image quality and dose delivery performance and has formed a prospective study into development of an operator specific dosimetry solution.

KEYWORDS: MOSkin, MOSFET, Angiography, Dosimetry, Image Quality
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Chapter 1

Radiation Induced Risk in Radiology

Diagnostic radiology is an invaluable tool with widespread utility in the diagnosis and treatment of disease. Radiographic procedures harness the properties of radiation to render detailed structural and functional mappings of a patient’s physiology that otherwise could not be seen or completely understood. The utilisation of radiographic procedures continues to grow as medical knowledge, imaging technologies and clinical techniques are developed and improved on. This has improved patient outcomes at the cost of increasing clinical radiation exposures to patients and, in some cases, to clinicians. While radiation exposure should be reduced where possible, if a procedure is necessary for the diagnosis or treatment of a condition then the risk-benefit ratio of the procedure would favour performing the procedure. Instead of forgoing these procedures, efforts should be placed towards monitoring exposures, developing dose minimisation strategies and developing technologies and techniques that lower radiation exposures where possible. In 1977, the International Commission on Radiation Protection (ICRP) Publication 26 proposed general recommendations that included the adoption of an effective dose quantity\(^\text{[1]}\). The quantity was intended as a simplified
radiation protection quantity that would represent the biological impact of all radiation sources on all organs and biological tissues irradiated during an exposure. To calculate the effective dose, the dose delivered to each organ volume is first weighted based on the type of radiation that is incident and the type of tissue irradiated. These weighting factors were established in ICRP Publication 60 based on Monte Carlo simulations before later being updated in ICRP Publication 103 using newer simulation frameworks and clinically reported evidence\cite{2,3}. The weighted dose values are then summated into an effective dose value, as shown in Equation 1.1 The effective dose is useful for the justification, optimisation and limitation of radiation exposures to the public.

\[ E = \sum_T w_T \sum_R D_{T,R}, \text{ Sv} \quad (1.1) \]

Effective dose has become a convenient measure of the radiation exposures and of radiation-induced risk attributed to patients undertaking radiology procedures. There are several reviews in the literature comparing effective doses reported for various radiology procedures using this metric. These reviews were written with the intent to inform clinical behaviours and to open discussion on the risks presented by everyday clinical procedures. This section will discuss some of these reviews in detail, specifically the European Dose DataMed (DDM) initiatives that identified radiation exposure risks to the public, the 2008 Mettler review that compared a broad range of radiology procedures and the Vilar-Palop review which presented a modern update to the Mettler review with consideration of the outcome of the DDM initiatives, paediatric exposures and of the update in ICRP weighting factors\cite{4-7}. In 2003, the European Dose DataMed initiative was launched to assess public exposure from diagnostic and interventional imaging procedures\cite{4}. The DDM collected information on the frequency of diagnostic and interventional procedures and the effective doses experienced during these procedures as reported across ten European countries. As part of the survey, the
DDM identified the top twenty procedures that were the highest contributors to the collective public exposure. The DDM initiative laid the foundation for the broader DDM2 initiative that expanded the scope of the study to 36 European countries, continued monitoring public exposure from the top twenty contributors list and explored the implementation of diagnostic reference levels in different procedures[5]. The DDM top twenty identifies four key categories of imaging procedures: radiographs, computerised tomography (CT) scans, fluoroscopic imaging and interventional cardiology procedures. This was consistent with a report from the United Nations Scientific Committee on the Effects of Atomic Radiation (UNSCEAR) published in 2008 that stated ‘just under half of the collective effective dose due to diagnostic radiology arises from three key procedures: CT, angiographic examination and interventional radiology’[8].

In 2008, another extensive review of effective doses in common radiology procedures was published by Mettler et.al.[6]. The review summarised findings published in 157 peer reviewed articles published between 1980 and 2007, representing clinics based across the United States, Canada, Japan, Australia and Western Europe. Mettler et.al. presented the effective doses for 31 radiographic procedures with literature reported values ranging from 0.0002 mSv to 180 mSv. In publishing a review of these reported effective doses, Mettler et.al. intended to provide patients and clinicians with an indication of radiation-induced risk and detriment that could be used both in the justification and optimisation of procedures. Mettler et.al. noted that the literature reported the highest effective doses during computerised tomographic scanning and interventional procedures. He also noted high variation in effective doses reported during fluoroscopic and interventional procedures. In 2016, Vilar-Palop et.al. published a review of effective doses reported in the literature and noted changes in these results since the 2007 Mettler et.al. publication[7]. The Vilar-Palop et.al. review focuses on the top 20 procedures identified by the DDM initiative and summarises the find-
ings of 33 peer-reviewed articles representing clinics in Asia, Australia, Brazil, Europe and the United States. The review reported effective doses calculated using both the ICRP Publication 60 and ICRP Publication 103. Both standards were included where possible as many publications were reporting exposures either using the deprecated ICRP Publication 60 guidelines exclusively or in supplement to the ICRP Publication 103 guidelines. The review also used the average effective dose values reported by the Mettler et.al. review and the DDM2 initiative in comparison to the study’s reported figures.

A commonality between all four of these reviews was that interventional procedures were one of the highest dose procedures despite their comparatively lower procedural frequency. The DDM initiatives reported two interventional cardiology procedures, cardiac angiography and percutaneous transluminal coronary angioplasty (PTCA), in their top 20 highest dose procedures. In both the Mettler et.al. and Vilar-Palop et.al. literature reviews, interventional cardiology procedures reported exposures ranging across a broad effective dose range with some procedures ranking consistently amongst the highest dose procedures reviewed. Another review of the literature written by Pantos et.al. suggested that this variability in interventional cardiology procedure doses correlated closely with the complexity of the procedures, the type of fluoroscopic equipment available at the catheterisation laboratory, the differences in operator experience and the level of radiation protection training employed at individual clinics[^9]. The consensus within the literature is that exposures in interventional cardiology procedures have high potential for optimisation and that there is a real need for comprehensive dosimetry solutions informing the implementation of dose minimisation strategies.


1.1 Interventional Cardiology Procedures

Interventional cardiology is a subdiscipline of interventional radiology that focuses on the diagnosis and treatment of cardiovascular disease. Procedures are performed through the execution of minimally invasive percutaneous surgical techniques. Procedures can involve coronary catheterisation, angiographic imaging, angioplasty, delivery of thrombolytic agents, stent placement, valve replacement/repair and radiofrequency ablation. Procedures are performed under fluoroscopic image guidance provided by an angiographic c-arm system. These systems and techniques enable clinicians to administer targeted treatments to cure disorders where open surgery would have once been necessary. Cardiovascular disease is the leading cause of death in Australia and was responsible for 43,963 deaths in 2016\textsuperscript{[10]}. In the 2014/2015 financial year, approximately 4.2 million Australians reportedly suffered from cardiovascular disease, which includes 2.6 million reports of hypertension and 1.2 million reports of heart disease\textsuperscript{[11]}. This represents a 53\% increase in reports of hypertension and a 36\% increase reports of heart disease since the year 2001. In the 2015/2016 financial year, the Australian Institute of Health and Welfare reported that 577,661 procedures were performed on patient cardiovascular systems, which was an increase in procedures of 34.6\% since the 2000/2001 financial\textsuperscript{[12,13]}. This total for cardiovascular procedures in the 2015/2016 financial included:

- 1,918 cardiac catheterisation procedures, which is an increase of 77.4\% since the 2000/2001 financial year.
- 134,195 coronary angiography procedures, which is an increase of 63.8\% since the 2000/2001 financial year.
- 44,334 percutaneous transluminal coronary angioplasty procedures, which is an increase of 88.8\% since the 2000/2001 financial year.
1.1. Interventional Cardiology Procedures

Figure 1.1: Percentage Share and Total of Revascularisation Procedures Performed in Australia (2000-2014)[14]

Figure 1.2: Trends in Revascularisation Procedural Preference in OECD Countries (2000-2013)[14]
According to the Organisation for Economic Co-operation and Development (OECD) health report of 2015, utilisation of angioplasty in revascularisation procedures in Australia has gradually increased from 56.9% in the year 2000 to 76.0% in 2014, as shown in Figure 1.1\textsuperscript{[14]}. This trend was observed globally across all 31 of the OECD survey countries, as shown in Figure 1.2. The growing utilisation of angioplasty techniques can be attributed to the benefits that these techniques provide over traditional bypass surgery and is consistent with the decline in utilisation of bypass surgery, traditionally the gold standard for revascularisation procedures. This shift in preference makes sense, angioplasty offers health outcomes comparable to bypass surgery for most forms of coronary artery disease through use of minimally invasive techniques and can be performed without inducing anaesthesia. The development of new angiographic techniques and equipment, such as drug eluting stents, are constantly improving patient outcome, improving the versatility of the procedure and reducing the advantages that traditional bypass surgeries can offer. Angioplasty patients also experience less pain, reduced hospitalisation times and reduced risk of in-hospital mortality when compared to bypass patients\textsuperscript{[15-17]}. Bypass surgery is still recommended for patients over the age of 65, patients with diabetes, patients with multivessel coronary artery disease, patients with impaired ventricular function and for patients who have experienced restenosis. While angioplasty does offer many advantages over bypass surgery, the procedure does carry a higher risk of restenosis than bypass surgery and requires fluoroscopic imaging, meaning it contributes to the total public medical exposure.

Advances in angiographic techniques and equipment have also led to the development of percutaneous treatments for paediatric and adolescent heart conditions that were once either untreatable or required open heart surgery\textsuperscript{[7,18]}. Paediatric conditions such as patent ductus arteriosus and patent foramen ovale are now commonly treated using interventional techniques including atrial septostomy, aortic dilatation, aortic
1.1. *Interventional Cardiology Procedures*

valvuloplasty, pulmonary valvuloplasty and pulmonary artery angioplasty. Due to the complexity of these interventional paediatric procedures, patients may experience extended imaging times, require multiple catheterisations and require regular angiographic follow-ups after treatment. As such, these procedures, while lifesaving, can result in significant accumulated radiation exposures to children which can result in fluoroscopic injuries and increases the child’s lifetime risk of developing cancer\[^{7,18-22}\]. This is in part due to the heightened radiation sensitivity and longer remaining life spans of children as compared to adults. The increasing use of interventional cardiology procedures, the broadening range of applications for interventional techniques, the increasing exposure to paediatric patients and the growing public diagnostic collective exposure substantiate the need for thorough investigation into the radiation related risks involved with these procedures and the potential for dose minimisation in every day practice.
1.2 The Risks of Radiation-induced Injury during Interventional Procedures

1.2.1 Classifications of Radiation-Induced Injury

To understand the necessity of monitoring exposure, it is important to understand the risks associated with radiation overexposure. Radiation damage to biological material occurs when radiation is incident upon a cell and causes damage to the cell through direct or indirect action\(^\text{[23]}\). Direct action refers to when radiation causes ionisation within the biological material of interest, potentially by ionising key structural molecules within organelle structures, while indirect action refers to when radiation causes ionisation through highly reactive free radicals produced by direct action\(^\text{[23 - 25]}\). These free radicals most commonly include hydroxyl groups, nitrogen species and alkoxyl type free radicals. The damage caused by radiation exposure can cause cell death and genetic abnormalities, or mutations, in cells that survive the exposure. This can cause the tissue to develop radiation-induced conditions. These conditions can be categorised into two classifications: stochastic conditions and deterministic conditions\(^\text{[1,3]}\).

Stochastic conditions occur due to genetic mutations and abnormalities\(^\text{[25]}\). Genetic mutation happens spontaneously in nature but can occur at greater frequency due to radiation exposure. While genetic mutations can be beneficial, they more commonly cause genetic diseases and carcinogenesis and can result in hereditary effects that can be observed in the exposed individual’s subsequent offspring. Stochastic conditions occur with probabilistic incidence and therefore the probability of inducing a stochastic condition will most likely increase as radiation exposure is accumulated over time. The severity of stochastic conditions is independent of dose\(^\text{[3]}\). The ICRP recommends assessing stochastic risk using the effective dose metric.
the linear no-threshold model, the ICRP estimated in ICRP Publication 103 that the risk of developing a stochastic condition from an exposure is approximately $5.7\% /\text{Sv}$. This figure was comprised of an increased risk of carcinogenesis of $5.5\% /\text{Sv}$ and an increased risk of heritable effects of $0.2\% /\text{Sv}$. The US National Research Council on Radiation Protection and Measurements (NCRP) supported the linear no-threshold model for doses lower than 100 mSv at the conclusion of their seventh Biological Effects of Ionizing Radiation report (BEIR VII) stating that there was no observable threshold below which ionising radiation was proven to be harmless or beneficial\cite{26}. UNSCEAR, the NCRP and the ICRP have all released publications concluding that the linear no-threshold model is an important basis for radiation protection measures for low dose exposures but caution that the model may not accurately represent biological risk for low dose exposures\cite{3,27,28}. Other studies have disputed the linear no-threshold model claiming that there is insufficient epidemiological evidence. Instead, these studies suggest that thresholds for stochastic conditions may exist, that dose-response behaviour may exhibit non-linearity and that individual risk thresholds may vary greatly from population-averaged thresholds. Some studies even suggest low doses of radiation may be beneficial, as per the radiation hormesis model, and conclude that extrapolation of the linear no-threshold model is harmful both to scientific and public understandings and perspectives of radiation induced-risk\cite{29 - 34}.

Deterministic conditions occur when radiation exposure to a tissue exceeds a threshold dose value and causes sufficient levels of cell death within a tissue which results in impaired functionality of the tissue\cite{3}. Deterministic risk can be estimated by assessing the radiation dose delivered to specific tissues during a procedure. Once the threshold dose is exceeded, the severity of the condition will increase with further exposure to the tissue. Some tissues, such as the skin, can repair over time which means that, depending on the tissue, subsequent exposures may not necessarily accumulate to
cause deterministic conditions. The ICRP recognises that radiation sensitivity varies between individuals and as such in ICRP Publication 103, the ICRP set deterministic thresholds using conservative dose values that represent the absorbed dose required to induce a condition within 1% of the general population. Commonly observed deterministic conditions include cataract formation, sterility, dermatitis, hair loss, epilation, erythema, telangiectasia, ulceration of tissues and, in extreme cases, acute radiation syndrome, cellular necrosis and death[35].

1.2.2 Radiation-induced injuries in the Catheterisation Lab

Earlier it was established that interventional radiology procedures, specifically coronary interventions, result in significant contributions toward total accumulated public exposure despite comparatively lower procedural frequency. The reviews observed that the radiation exposures during these types of procedures vary considerably. Patient exposure may also be extended due to unforeseen circumstances, repeated procedures or complications during a specific procedure. As such, patient dose and the incidence of radiation-induced conditions can be difficult to anticipate or even to assess. To conservatively estimate the incidence of radiation-induced conditions resulting from diagnostic doses, the doses reported by the literature for the two highest exposure coronary interventional procedures identified in the DDM top 20 have been collated below:
• The Dose DataMed 2 initiative reported that cardiac angiography procedures resulted in an average effective dose of 7.71 mSv with reported values ranging from 3.3 - 11.3 mSv while angioplasty procedures produced an average effective dose of 15.2 mSv with reported values ranging from 4.0 - 29.0 mSv\[5\].

• The Mettler et.al. review reported that cardiac angiography procedures resulted in an average effective dose of 7.0 mSv with reported values ranging from 2.0 - 15.8 mSv while angioplasty procedures produced an average effective dose of 15.0 mSv with reported values ranging from 6.9 - 57.0 mSv\[6\].

• Vilar-Palop et.al. reported that cardiac angiography procedures produced an average effective dose of 9.3 mSv with reported values ranging from 3.3 - 22.3 mSv while angioplasty procedures produced an average effective dose of 19.5 mSv with reported values ranging from 7.4 - 48.6 mSv\[7\].

If the stochastic risk were extrapolated from the ICRP risk estimate presented earlier, a 5.7% risk of stochastic condition incidence per Sv, then these average effective doses could translate to an average of 944 stochastic conditions induced per million angioplasty procedures performed. Alternatively, using the collective doses reported during the Dose DataMed 2 initiative would predict 2,562 stochastic conditions induced per year from the exposures experienced during interventional procedures\[5\]. There are some published epidemiology studies researching the stochastic risks involved in diagnostic procedures. Mathews et.al found that in Australia, CT patients exposed to an average procedural effective dose of 4.5 mSv exhibited a 24% higher incidence of cancer\[36\]. Mathews et.al, observed risk increased was proportional to exposure and was greater for younger patients. While their study involved a different imaging modality and potentially assessed a different patient demographic, the doses in interventional cardiology are similar or higher than those seen in their study and as such it is not
1.2. The Risks of Radiation-induced Injury during Interventional Procedures

unreasonable to assume similar stochastic risks may be induced from angiographic exposures\cite{37}.

Patients are also at risk of developing deterministic conditions from angiographic procedures. During angiographic procedures the maximum dose is delivered at skin depth and as such the patient’s skin is especially vulnerable to deterministic injuries. Koenig et.al. produced an extensive two-part review of radiation-induced injuries resulting from fluoroscopic exposures\cite{38,39}. The review included 47 angiography patients who had suffered from injuries that ranged from radiodermatitis, erythema, ulceration and even necrosis of the skin tissue. Injured patients included a 48-year-old woman with severe coronary artery disease who underwent two percutaneous transluminal coronary angioplasties with stent placement performed within the same month who suffered a well-marginated focal erythema with desquamation within 2 months of the last procedure and a 17-year-old girl with a history of cardiac arrhythmia who underwent two cardiac ablation procedures within a 13-month period and developed an atrophic indurated plaque, skin telangiectasia, was placed at an increased risk of breast cancer and suffered limited mobility in her right arm. Koenig et.al. suggest that these deterministic conditions occurred when long exposure times were necessary, when large field sizes were predominantly used, where limited or overlapping beam projections were used, when extraneous body parts were imaged, when thicker patients were imaged, when high exposure imaging settings were utilised and when clinicians were not trained to practise dose minimisation strategies and were unaware of radiobiological effects. Koenig et.al. also noted that real-time direct dosimetry solutions to monitor skin dose were not implemented during any of the case studies investigated and suggested that the implementation of these dose monitoring systems would inform clinician behaviour in a more meaningful way than c-arm reported quantities such as fluoroscopic time. While the Koenig et.al. review was published in 2001, similar
1.2. The Risks of Radiation-induced Injury during Interventional Procedures

Concerns of fluoroscopic injury have since been published since then by Balter et al. and Boncher et al. that were motivated by similar concerns\cite{39, 40}. In some cases, it is necessary to perform interventional procedures to pregnant women to administer life-saving treatments. Care must be taken when imaging during these procedures as foetuses are particularly susceptible to both stochastic and deterministic conditions and exposure could lead to malformation or death of the foetus\cite{22}.

Clinicians experience a significantly lower radiation exposure than patients during procedures, however, recent studies suggest that clinicians may be at risk of radiation-induced conditions due to accumulation of exposure between procedures over the span of the clinician's career. Clinician based studies have reported that interventional operators are potentially at risk of developing stochastic conditions such as predominantly left-brain tumours and deterministic conditions such as dermatitis and hair loss to the hands and legs\cite{41 - 43}. Recent studies have also presented compelling evidence suggesting that clinicians are developing radiation-induced rear-lens cataracts\cite{44 - 48}. The initial evidence suggested that the observed radiation-induced cataract formation was deterministic in nature, however, there are studies that consider the possibility that the condition is stochastic in origin\cite{29}. Details about clinician conditions are still emerging and are further shaping our understanding of the impact of low dose exposures. Radiation protection agencies including EURADOS, ICRP, the International Atomic Energy Agency (IAEA) and the Australian Radiation Protection and Nuclear Safety Agency (ARPANSA) have responded as new evidence becomes available by releasing statements updating recommended dose limits\cite{49 - 51}. 
1.2.3 Sources of Radiation Exposure in Angiography

Interventional cardiology procedures are performed with a device known as the angiographic c-arm system. The c-arm system consists of a c-shaped gantry with a fluoroscopic x-ray tube attached to one end of the gantry and a radiation imaging detector mounted to the other. The c-arm system produces a filtered spectrum of kilovoltage range diagnostic x-rays. The c-arm x-ray beam characteristics are set automatically using vendor-specific algorithms installed to the system on commission. The c-arm system selects beam qualities based on information specific to each imaging projection and target using metrics such as the photon flux incident on the imaging panel and the signal-to-noise ratio of the acquired images. Most systems operate under a pulsed imaging modality with each image acquired within milliseconds. Some systems, known as G-Stand or biplanar systems, include a secondary imaging apparatus. During imaging, the primary beam is incident on the patient volume. Scatter radiation is produced when radiation scatters from the patient volume, patient couch, imaging plate, roof and walls of the catheterisation laboratory. Clinicians are partially protected from primary beam by radio-protective boundaries and lead shielding. The distribution of scatter radiation is dependent on the type of c-arm system used, the catheterisation laboratory geometry and the material of the roof and walls and as such the scatter distribution is specific to each individual catheterisation laboratory.
1.3 Radiation Safety and Exposure Limitations

1.3.1 Measurement and Assessment of Patient Dose

For patients, radiation exposure during angiographic procedures is routinely estimated using the estimated air Kerma delivered to the international reference point and the Kerma Area Product (KAP)\textsuperscript{[52 - 59]}. Fluoroscopic imaging time is also recorded but is not a reliable indicator of patient dose.

The air Kerma is defined as the kinetic energy released per unit mass in air by uncharged ionising radiation. The air Kerma is typically monitored using a KAP meter adjacent to the x-ray tube. In Australia, all newly commissioned c-arm systems are legally required to include an installed KAP meter\textsuperscript{[58]}. The air Kerma measured by this KAP meter is used to estimate the air Kerma at international reference point which is defined at 15 cm from the rotational isocentre of the c-arm gantry toward the x-ray tube. This point estimates the distance between the x-ray tube and the underside of the patient and is used to estimate the patient skin dose. The derivation of air Kerma is performed utilising the KAP.

\[
DAP = KAP = Kerma \times Area
\]  

(1.2)

The KAP is defined as the air Kerma multiplied by the cross-sectional area of the x-ray beam at measurement point. This is expressed in Equation 1.2\textsuperscript{[24]}. When calculating the KAP quantity, it is assumed that changes in photon fluence are driven solely by the inverse square law. The inverse square law states that emissions originating from a point source will reduce in fluence, or intensity per unit area, proportionally to the distance from the source squared. This phenomenon occurs due to geometric dilution, that is, while net intensity is constant, the area that the emissions are spread...
1.3. Radiation Safety and Exposure Limitations

Figure 1.3: Illustration of the Kerma Area Product

across increases proportionally to the distance from the source squared. By taking the product of the photon intensity at measurement point and the cross-sectional area at measurement point the dependencies on distance from the point source squared are cancelled out. This means the KAP quantity is effectively independent of distance from the point source, as demonstrated in Figure 1.3. As such, the KAP measured by the KAP meter is equal to the KAP at international measurement point which can be used to estimate the air Kerma at international measurement point.

Older c-arm systems have historically presented a dose area product (DAP) quantity. In 2008, the term DAP meter was revised and replaced with the more accurate term KAP meter\cite{59}. This has resulted in some publications using the terms Kerma and Dose interchangeably. The NCRP recommends that angiography patients should be referred for follow-up should peak skin dose exceed 3 Gy, cumulative air Kerma at reference point exceed 5 Gy or fluoroscopy imaging time exceed 60 minutes\cite{57}. The NCRP also recommends against clinics relying on fluoroscopy time alone for referral.
ARPANSA have published an initial national dose reference level (NDRL) for angiographic procedures\textsuperscript{[60]}. These reference levels are established by surveying clinical data and are used for optimisation purposes or for benchmarking of clinical performance. The NDRL is set at the 75th percentile of the survey data. At time of writing, the only published interventional NDRL was for coronary angiograms performed on adult "Patients with ‘normal’ coronaries; no or physiologically insignificant diameter stenosis (\(< 50\%\) narrowing)". The current published NDRLs for this procedure are KAP = 30 Gy · cm\(^2\) and air Kerma = 0.5 Gy.

1.3.2 Limitations on Occupational Exposures

ARPANSA is the Australian government’s primary authority on radiation protection and nuclear safety. ARPANSA requires that all radiation workers likely to receive an annual occupational exposure of greater than 1 mSv should be equipped with an appropriate personal radiation monitoring device\textsuperscript{[58]}. This device should be worn on the torso between waist and chest height under any personal protective equipment and should be worn for no longer than three months before readout of the dosimeter is performed. ARPANSA also recommends use of a secondary dosimeter to be worn outside of any personal protective equipment at collar level and approves of the use of extremity dose monitors where appropriate. ARPANSA specified occupational equivalent dose limits are consistent with recommendations published in ICRP Publication 103 except for the eye lens equivalent dose limits which were updated in ICRP Publication 118\textsuperscript{[3,32,51,59]}. These recommendations are as follows:
1.3. Radiation Safety and Exposure Limitations

- The annual effective dose resulting from occupational exposures should not exceed 20 mSv per year as averaged over a 5-year period and should not exceed 50 mSv during any single year.

- In the case of pregnant workers, effective dose should not exceed 1 mSv for the duration of pregnancy.

- The equivalent dose to the eye lens resulting from occupational exposures should not exceed 20 mSv per year as averaged over a 5-year period and should not exceed 50 mSv during any single year.

- The equivalent dose to the hands, feet and skin as averaged over one square centimetre of skin should not exceed an effective dose of 500 mSv per year.

Figure 1.4: a) Cardiologist’s Personal Protective Equipment and b) Common Operator Exposure Monitoring Points\textsuperscript{[58]}
Occupational dose is routinely monitored using personal radiation monitoring devices which often incorporate either film or luminescent dosimeters as their radiation sensitive element. These dosimeters will be discussed further in Section 1.5. ICRP Publication 139 recommends use of supplementary dosimeters to monitor doses delivered above the clinicians’ collar level and delivered to the clinicians’ extremities including to the fingers, thigh and gonad regions\cite{61}. Radiation personal protective equipment and operator dose measurement points recommended by ICRP Publication 139 are depicted in Figure 1.4.

1.4 Alternative measures of radiation exposure

Clinicians may choose to supplement mandatory exposure reporting with a range of available dosimetry solutions. This section will detail solutions most commonly presented in the literature for coronary angiographic procedures listing the advantages that each can provide to clinicians.

1.4.1 Radiochromic Film Dosimeters

Radiochromic film dosimeters are large area film dosimeters used to measure radiation dose distribution over a specified area. Radiochromic film has applications in machine quality assurance, patient dosimetry and clinician dosimetry. Radiochromic film features an active layer commonly embedded between protective polymer/polyester sheets. The active layer typically consists of a marker dye that will change colour when exposed to radiation. Additives can be mixed into this active layer by manufacturers to adjust the film characteristics and produce specialised films for specific applications. Radiochromic film dosimeters are often considered an industry gold standard due their long history of use in radiation laboratories.

Radiochromic film dosimeters provide a visualisation of the dose delivered to a pa-
1.4. Alternative measures of radiation exposure

tient volume in high level spatial resolution. As such, film dosimeters can retroactively identify areas of local overexposure and can be used to quantify the peak skin dose experienced during procedures. The most common radiochromic film used during angiographic procedures is the Gafchromic XR-RV3 film which is designed for use with interventional beam qualities. Farah et.al. estimate that the XR-RV3 type film can be used to determine skin dose delivery to within typically 20% uncertainty, however, the accuracy of the film is heavily dependent on the quality of the film, the scanning methodology and the care taken during readout\(^{[62]}\). Farah et.al. estimated that proper care can reduce uncertainty to within 5% while minimal care can cause uncertainty to increase to up to 40%. Due to the cost of buying new film, the use of film in catheterisation laboratories tends to be reserved for high dose procedures, such as chronic total occlusion cases, where there is risk of exceeding the 2 Gy deterministic threshold\(^{[63]}\).

1.4.2 Indirect Dosimetry Solutions

Indirect dosimetry methods involve estimation of patient effective dose and peak skin dose using exposure parameters such as air Kerma at reference point, Kerma area product, imaging time, x-ray tube voltage, beam current, pulse width c-arm position, c-arm angulation. These studies are often supplemented with data from direct dosimetry solutions. A study by Compagnone et.al. attempted to estimate patient effective doses through Monte Carlo simulations that used exposure parameters provided by the c-arm system\(^{[64]}\). Compagnone et.al. were able to calculate estimations of the patient effective dose and equivalent doses delivered to specific patient organs. Compagnone et.al. observed close correlation of the estimated effective dose with the machine reported dose area product using two separate simulation methodologies. Compagne et.al. also found that using simple conversion coefficients with machine reported DAP values could be used to effectively estimate equivalent dose to 10 crit-
ical organs. Compagne et.al. suggest that these equivalent dose values may be more appropriate quantities for estimating patient exposures and stochastic risk than the estimated effective patient dose. In a 2008 study, Dominiek et.al. attempted to correlate DAP to the maximum skin dose delivered to patients as monitored using large area film dosimeters to prevent deterministic injury\cite{65}. The research concluded similarly to previous studies performed by Vano et.al., Van de Putte et.al. and Morell et.al. in that dose area product alone cannot be used to predict maximum skin dose in a meaningful way\cite{66 - 69}. Despite this result, Dominiek et.al. suggested DAP could be used as a rough indicator of risk and could be used to define trigger levels to caution clinicians on patient exposure. Dominiek et.al. also stressed that correlations between maximum skin dose and system beam monitoring quantities should be established individually for each catheterisation laboratory they are deployed at.

1.4.3 Dose contouring software packages

Vendors have recently begun installing commercial dose distribution visualisation software packages\cite{70 - 72}. These software packages paint the air Kerma values as reported by the c-arm system to a standardised patient volume. Visualisation better enables clinicians to utilise a variety of beam projections during imaging which reduces the risk of local overexposure. In the validation of the GE Dose Map system, Bordier et.al. observed that the dose distributions produced by the dose tracking system compared favourably to dose distributions assessed using radiochromic film and could predict peak skin dose to within 25% of the dose measured by XR-RV3 film\cite{71}. A key advantage of visualisation software packages is that they provide an estimation of dose distribution that is functionally similar to the results available using film-based dosimeters, but they require less effort to set up and are reusable without incurring costs additional to the initial software purchase price. Furthermore, in the future,
1.4. Alternative measures of radiation exposure

visualisation software could potentially be paired with virtual and augmented reality technology which is already used in some clinics to guide catheterisation procedures.

1.4.4 Solid-State Dosimetry: Band Theory Basics

The remaining common dosimeters in angiography are solid-state dosimetry solutions. Solid-state devices take advantage of the conductive properties of metals, insulators and semiconductor materials\(^{[23]}\). This section will explain the electronic properties of solid-state dosimeters using band theory so that the mechanisms used by solid-state dosimeters to measure incident radiation can be explained more specifically in subsequent sections.

Band theory introduces two bands that are used to characterise electrons in a material, the valence band and the conduction band. The valence band consists of the electrons that exist within the outermost stable orbital electron shell of the materials atomic lattice while the conduction band consists of electrons that can move freely throughout the lattice and as such contribute as charge carriers, enabling the material to conduct charge. In metals there is an excess of electrons present in the material. The excess electrons partially fill the conduction band and electrons can transition between free and bound states without stimulus. In insulative non-metal materials the conduction band is depleted and the valence band is full. These valence band electrons are bound tightly to the lattice and do not contribute to conduction. The energy input needed to break the bond between the electron and lattice, that is, to transition electrons from the valence band to conduction band, is known as the band gap. In insulators this band gap is typically prohibitively large and as such insulators will inhibit conduction of charge carriers. The electrical properties of a material will exist on a spectrum of perfect conductor (no band gap) through to a theoretical perfect insulator (infinite band gap).
Semiconductors are characterised by their approximately half-filled outer orbital electron shells. This electron configuration enables semiconductors to exhibit both conductive and insulative properties. Due to the lower energy difference between the highest occupied molecular orbital and lowest unoccupied molecular orbital, semiconductor lattice bonds are typically weaker than the lattice bonds found in insulators. As such, the band gap in semiconductors is smaller than in insulators. The smaller bandgap means that electrons can transition between valence and conduction bands with relatively low energy input enabling conduction within the material. The electrical properties of semiconductors can be changed during fabrication in a process known as doping. The doping process involves introducing impurities to the semiconductor lattice during fabrication and can result in n-type or p-type semiconductors.

- **N-type** (negatively charged) semiconductors are produced by doping the semiconductor with group V electron donor atoms such as phosphorous or arsenic. The dopant atoms have an additional electron available. The electron will be unable to bond locally due to the geometry of the semiconductor lattice and therefore n-type semiconductors more negatively charged electrons than an intrinsic semiconductor. The excess electrons require less energy to transition to the conduction band than bound electrons. In band theory, these electrons occupy donor energy levels in the higher energy section of the band gap below the conduction band. Conduction in n-type semiconductors results from movement of electrons in the conduction band.

- **P-type** (positively charged) semiconductors are produced by doping the semiconductor with group III electron acceptor atoms such as boron or gallium. The dopant atoms have less electrons than the semiconductor atoms they replace and therefore cannot complete bonds with all neighbouring semiconductor atoms. The empty electron positions, holes, result in a lower net electron count and
therefore n-type semiconductors are more positively charged than an equivalent intrinsic semiconductor. The holes attract electrons toward the valence band. In band theory, these electrons occupy acceptor levels in the lower energy section of the band gap above the valence band. Conduction in p-type semiconductors results from movement of positively charged holes in the valence band.

Solid-state dosimeters take advantage of the electronic properties of metals, semiconductors and insulators to measure incident radiation. This is achieved by trapping charge carriers induced by ionising radiation within specific discreet energy levels that exist within the band gap. Readout of a solid-state dosimeter will measure the number of trapped charges within the dosimeter either through liberating the trapped charges or by measuring changes in the electrical properties of the dosimeter. For a dosimeter to be effective, the relationship between the number of trapped charges and the quanta of incident radiation must be consistent and well understood.

1.4.5 Luminescent Dosimeters

Luminescent dosimeter is a term used to describe several solid-state dosimeters that emit light proportional to the radiation the dosimeter has been exposed to during irradiation of the dosimeter\textsuperscript{[23]}. The main types of luminescent dosimeter include thermoluminescent dosimeters (TLD), optically stimulated luminescent dosimeters (OSLD) and radiophotoluminescent glass dosimeters (RPLD or RPLGD)\textsuperscript{[73 - 78]}. These dosimeters are commonly packaged as chips or pellets that can be used for measuring radiation exposure to a discreet point. Luminescent dosimeters can also be used to measure dose distributions across an area either by arranging the dosimeters into an array or by suspending a powder format of the dosimeter throughout a material. TLDs are luminescent dosimeters that emit light proportional to accumulated radiation exposure when heated. The most common TLDs consist of lithium
fluoride doped with either magnesium or titanium. These dopants produce charge traps at discreet energy levels distributed throughout the band gap. When exposed to ionising radiation, electrons are scattered within the TLD. Some of these charges will be captured within the dopant-induced charge traps. Readout of TLDs is performed by thermally ‘annealing’ the dosimeters. During the annealing process, the TLD is coupled to a photomultiplier tube and placed within an oven/kiln/furnace. The ambient temperature within the apparatus is raised slowly. As an example, a typical TLD-100 readout will involve heating the TLD from 50 °C to 350 °C at a rate of ∼0.2 °C/s. As ambient temperature increases, the energy distributed throughout the TLD increases. When the energy within the TLD exceeds the band gap energy of a charge trap the charge will be liberated and a photon will be emitted. These photons are collected by the photomultiplier tube. The readout results in a photon collection profile, the ‘glow curve’, that represents the data in terms of photon count with respect to either time or temperature. The glow curve will feature peaks in photon collection at specific times. These peaks correspond to the band gap energy of specific charge traps. The photons collected in specific peaks are integrated and these integrated values are proportional to the accumulated radiation dose delivered to the TLD. This readout process is destructive meaning dose cannot be verified but the TLD can be reused post-annealing. OSLDs are luminescent dosimeters that emit light proportional to accumulated radiation exposure when exposed to an ultraviolet range to visible light ranged light source. These dosimeters are typically composed of carbon doped aluminium oxide. OSLDs are functionally analogous to TLDs in that the carbon dopant creates charge traps within the lattice that trap charges scattered by ionising radiation. The primary differentiation is in the readout method utilised. Readout of an OSLD is performed by optically stimulating the OSLD with a pulsed laser light and collecting emitted photons via a coupled photomultiplier tube. As an
1.4. Alternative measures of radiation exposure

Example of stimulation sources, the aluminium oxide OSLD is stimulated using an argon laser light (λ ≈ 515 nm). The incident laser light pulse will liberate a portion of the trapped charges causing photons to be emitted by the OSLD which are then collected by the photomultiplier tube. As only a small portion of the trapped charges are liberated in the readout process, OSLD readout can be repeated allowing users to reproducibly verify readouts and to maintain physical records of exposures. OSLDs are replacing TLDs in many applications due to their high sensitivity (1 µGy minimum reported dose), large dose measurement range (readout achievable in range of 10 µGy to 10 Gy), reduced fading effects as compared to traditional TLDs and their ability to provide a non-destructive verifiable dosimetry history[73 - 76]. RPLGDs are the latest developed type of luminescent dosimetry technology[76 - 78]. Unlike TLDs and OSLDs, the glass compound is an inorganic amorphous solid. RPLGDs are composed of a silver-phosphate activated doped glass compound. On irradiation, ionised charges interact with the silver phosphate to create Ag$^0$ and Ag$^{2+}$ ions isolated within the glass. These ions act as electron and hole traps respectively and are referred to either as colour or defect centers. Readout is performed by pulsing an ultraviolet light source (λ ≈ 337 nm) onto the RPLGD. This stimulates the colour centers and transitions the trapped charges into an excited state. The trapped charges will subsequently undergo deexcitation and will emit photons in the high energy visible range (λ ≈ 600 – 700 nm). As trapped photons are not liberated, RPLGDs provide a permanent history of accumulated dose and can be reused after readout up to the maximum exposure value. The lifetime of a RPLGD is dependent on the amount of silver phosphate dopant used during fabrication. As of time of writing, the three most commonly used RPLGD lifetimes are the low-dose range RPLGD (maximum dose = 10 µGy), the high-dose range RPLGD (maximum dose = 10 Gy) and the radiotherapy dose range RPLGD (maximum dose = 500 Gy). The lifetime of these dosimeters is moderated by adjusting...
silver phosphate content. Increasing silver phosphate content will increase maximum readout value, however, it will also affect the penetrability of the dosimeter to ionising radiation and as such sensitivity is also dependent on the doping concentration. This energy dependence necessitates design of specific shielding solutions for the RPLGD to be used in low energy exposure applications. RPLGDs offer many advantages over conventional luminescent dosimeters such as their repeatable non-destructive readout process, improved intensity retention on repeated readouts (as compared to OSLDs) and broader potential measurement ranges. These advantages ensure that with further development RPLGDs will be an important part of the personal dosimetry technology landscape in the future.

1.4.6 Electronic Dosimeters

Active personal dosimeters, also known as electronic dosimeters, are solid-state qualitative radiation monitors that inform clinicians when the exposure rate delivered to the clinician increases significantly. These dosimeters can be set to trigger auditory and visual alarms when clinician exposure or exposure rate exceeds a threshold value. The active volumes for these dosimeters are typically based on Geiger-Müller tube, scintillation crystal or silicon diode technologies. A study performed by Clairand et.al. observed the energy dependence, angular dependence and dose rate dependence of several popular commercial electronic dosimeters and concluded that while the dosimeters performed with satisfactory response for continuous low dose radiation fields, not all dosimeters could respond reliably when exposed to high dose radiation fields and pulsed radiation fields. As such, Clairand et.al. stated that while electronic dosimeters are a useful tool for reducing operator dose, the dosimeters could not be recommended in place of the traditional passive dosimeters used for personal radiation monitoring.
1.4.7 MOSFET Dosimeters

MOSFET dosimeters are another specific field of solid-state dosimeter with a long history in radiation dosimetry applications. The first published use of MOFSFETs as radiation dosimeters was the advent of the RADFET as designed by Andrew Holmes-Siedel et.al. in 1974. The RADFET was a MOSFET-type dosimeter designed specifically for monitoring radiation exposures experienced by orbital satellites\textsuperscript{81}. The potential of the RADFET established interest in the technology in other industries including clinical exposure monitoring. While some clinics have had success in applying generic MOSFETs in clinical environments, there are more sophisticated solutions designed for clinical applications\textsuperscript{82}. In diagnostic radiological applications, the three most commonly published MOSFET dosimetry solutions available are the RADFET, the Thomas-Nelson MOSFET and the MO\textit{Skin} dosimeter\textsuperscript{81,83 - 95}. As the oldest MOSFET dosimetry system, the RADFET has undergone many iterations in design. Typically, the RADFET features a 1 mm epoxy encapsulation that acts as a build-up layer. Key developments in RADFET technology for clinical applications have included real-time readout capabilities and a stacked RADFET configuration that drastically enhances the device's sensitivity. The RADFET was the first MOSFET-type dosimeter to prove the viability and advantages of MOSFET-type dosimeters including small form factors, low production costs, instant non-destructive readout and easy integration with computerised systems. The Thomas-Nelson MOSFET, or also more commonly known as the Best Medical Canada mobileMOSFET, is currently the most extensively used MOSFET dosimetry solution available for diagnostic dosimetry. The mobile MOSFET is a dual bias MOSFET system encapsulated within a black epoxy bulb\textsuperscript{87, 88}. The mobile MOSFET is sometimes used with a variety of brass capping materials to improve reproducibility and angular response of the device. The mobileMOSFET system has been used extensively to measure doses delivered during
diagnostic procedures after successful validation of the dosimeter for CT applications in 2007\cite{89}. The system has also been applied in paediatric interventional procedures where it has been used to measure patient skin doses, to estimate stochastic risk to patients and to measure organ doses delivered to an anthropomorphic phantom volume\cite{90 - 93}. The final dosimeter listed, the MOSkin dosimeter, was developed by the Centre for Medical Radiation Physics for radiotherapy applications \cite{94,95}. The MOSkin design features a novel packaging solution that enables measurement at a reproducible surface depth meaning no corrections are necessary to measure the dose delivered to the measurement point and any measurement depth can be recreated by producing an appropriate build up layer. The MOSkin is capable of real-time dose monitoring and thermal compensation is built into the reader systems. While designed for radiotherapy procedures, preliminary studies have applied the MOSkin dosimeter in diagnostic applications such as CT and interventional procedures\cite{95}. The MOSkin dosimeter and the general behaviours of MOSFET-based dosimetry systems will be discussed in greater depth in Chapter 2.
1.5 Why aren’t direct dosimetry solutions broadly deployed during interventional procedures? The Potential of the MOSkin dosimetry solution

Direct dosimetry solutions are not commonly utilised in coronary angiography due to limitations of commercially available dosimeters. Several of the dosimetry solutions discussed so far lack real-time functionality and as such these dosimeters are limited to retrospectively evaluating risk. Most commercial dosimetry solutions that do feature real-time readout capabilities are unfortunately radio-opaque to the c-arm beam as shown in Figure 1.5 Radio-opaque dosimeters are acquired during imaging which can obscure the cardiologist’s view of the patient’s cardiovascular system which could potentially result in protracted examination times, increased risk of clinical errors and misdiagnosis of patient conditions[96]. Furthermore, radio-opaque dosimeters can affect the c-arm system’s selection of beam characteristics, such as the beam current, tube voltage, filtration and pulse width, which can increase the radiation exposure experienced during the procedure.

Despite the obvious challenges involved in implementing direct dosimetry solutions, the development of appropriate real-time direct dosimetry solutions for use in the interventional catheterisation laboratory is essential to understand and reduce radiation exposures during interventional procedures. The current existing solutions are not appropriate and lack real-time preventative capabilities. Simulation/projection methods are also inadequate as they do not physically measure the real doses experienced by patients and operators and therefore cannot truly represent clinical doses with absolute certainty. If an appropriate direct dosimetry solution were developed, it could potentially inform clinicians in real-time and not only assess risk, but potentially prevent injuries, assist in dose minimisation strategies and supplement existing solutions.
1.5. Why aren’t direct dosimetry solutions broadly deployed during interventional procedures? The Potential of the MOSkin dosimetry solution

The aims of this research thesis are as follows:

1. To prove the functionality and utility of the MOSkin dosimeter when deployed in the interventional catheterisation laboratory.

2. To demonstrate the behaviours of the MOSkin dosimeter in response to clinical beam qualities and international standard beam qualities through use of established and novel methodologies, developing a foundation toward comprehensive diagnostic characterisation of the dosimeter.

3. To develop clinically useful dose minimisation strategies.

4. To establish a methodology to simultaneously evaluate both the image quality and entrance dose delivered to a standard phantom for the comparison of c-arm systems and clinical imaging protocols.

5. To identify the needs for an appropriate real-time operator dose monitor.

6. To provide foundational research for a future dosimetry solution specifically tailored to address the needs of the interventional catheterisation laboratory.
Chapter 2

Development of Real-time Dosimetry Solutions for Application in the Coronary Catheterisation Laboratory

This chapter supplements the literature review regarding solid-state devices and introduces the MOSkin dosimeter, a novel radiation dosimetry solution developed by the Center for Medical Radiation Physics at the University of Wollongong. The MOSkin will be the primary dosimetry solution used throughout this research thesis and this chapter characterises the MOSkin dosimeter for clinical applications in the coronary catheterisation laboratory. The characterisation assessed the lifetime sensitivity of the MOSkin dosimeter, calibrated the MOSkin to clinical and standard radiation protection beam qualities, compared the MOSkin response to established dosimeters and by thorough simulated the MOSkin response to kilovoltage range x-ray beam qualities.
2.1 Introduction to MOSFET dosimetry solutions

2.1.1 Solid-State Dosimetry: Development of electronic dosimeters

Solid-state device is a term referring to an electronic component composed of semiconductor and metallic structures used to manipulate the flow of electric current \[^{23}\]. The fundamental functional structure of a solid-state device is the p-n junction, a semiconductor crystal that consists of p-type semiconductor and n-type semiconductor layers arranged in an alternating arrangement. Devices consisting of a single p-n junction are classified as diodes. Because of the doping of the diode layers, the p-type silicon will contain a higher concentration of holes while the n-type silicon will contain a higher concentration of electrons. This causes an electric field to form across the diode and charge carriers will accumulate at the junction. The diffusion of charge at the junction will result in the formation of ions. The Coulomb force from these ions will repel opposing charges which in turn inhibits further electron diffusion across the junction in the absence of an applied bias. As this region contains no charge carriers it is referred to as the depletion region. For charge to traverse the depletion region, the energy of the charge must surpass the potential difference between the p-type and n-type semiconductor layers. This potential difference, and therefore the conductivity of the device, can be manipulated by applying a bias. When a forward bias is applied to the diode (positive voltage to p-type side), the depletion region across the p-n junction is reduced. If the potential difference applied surpasses a threshold voltage, the diode will conduct current proportional to the bias applied. When a reverse bias is applied (negative voltage applied to the p-type side), the potential difference across the p-n junction will increase which will further inhibit conduction. Reverse biased diodes do exhibit a small current conduction across the p-n junction, however, this conduction
is a result of minority charge carrier conduction and is referred to as leakage current or reverse saturation current. The diode is the fundamental solid-state dosimeter, but more complicated devices can be developed from this foundation. One of the most important class of solid-state devices, and more pertinent to this research thesis, is the transistor. A transistor can be produced by fabricating three alternating layers of doped semiconductor, that is, in either a p-n-p or n-p-n configuration. Transistors can be utilised to amplify signals or to redirect current based on the bias applied to the device. Transistors, specifically MOSFETs, have revolutionised electronics as they are inexpensive, mass-producible, extremely customisable and have enabled electronic devices to be miniaturised to the nanometre scale.

2.1.2 MOSFET-type Radiation Dosimeters

The Metal Oxide Silicon Field Effect Transistor (MOSFET) is a specific design of field-effect transistor. The MOSFET typically feature the following five structures/terminals that characterise current flow:

- Source
- Gate
- Substrate
- Gate Oxide
- Drain

There are two basic variants of MOSFET based on the alternating doped silicon configuration used in the (source-substrate-drain), the p-channel MOSFET (p-n-p) or the n-channel MOSFET (n-p-n). An example of a p-channel MOSFET is presented in Figure 2.1. The MOSFET is differentiated from other transistor designs by the
inclusion of a thin silicon dioxide layer separating the gate and substrate materials fabricated through inducing controlled oxidation upon the surface of the silicon substrate. During operation, electrons enter MOSFET via the source terminal. These electrons travel through the substrate. These charges cannot reach the gate terminal due to the insulative oxide layer between the gate and substrate and instead exit the substrate via a channel connecting to the drain terminal. The conductivity of the channel linking the source and drain terminals is determined by the potential difference between the gate and substrate terminals which can be adjusted by applying bias to the gate. A current will be induced across the MOSFET when the gate voltage exceeds a threshold voltage, $V_{\text{Th}}$. Radiation exposure affects the functionality of MOSFET type devices\cite{94}. When ionising radiation is incident upon a MOSFET, electron-hole pairs are generated throughout the device. The total charge induced is highly proportional to the absorbed dose delivered by the incident radiation to the gate oxide material. Electrons liberated within the oxide layer are removed by biasing the gate terminal. The removal of electrons via the gate causes hole migration and accumulation near the interface between the oxide layer and the substrate, approximately 0.2 to 20 nm from the interface. This region contains high concentrations charge traps as described in Section 1.5.4. These charge traps capture holes and forms a semi-permanent charge sheet near the interface region. The charge sheet produces an electric field which impairs the conductivity of the substrate material and as such necessitates a higher gate voltage to stimulate current flow across the device. By measuring the shift in the threshold voltage needed to stimulate current flow within the MOSFET, the energy deposited within the oxide layer, or the dose, can be determined. This means that the MOSFET can be used as a radiation dosimeter. An example of how these electrical characteristics shift after irradiation for a MOSFET dosimeter is presented in Figure 2.2. Radiation sensitivity of a MOSFET-type is determined by a few factors. The
gate oxide thickness is proportional to the number of ionisation events occurring due to incident radiation and as such a thicker gate oxide will increase sensitivity of the MOSFET. The collection efficiency of ionised charges in the MOSFET can also be moderated by inducing an electric field across the gate oxide. This is achieved by biasing the gate node.

MOSFET-type devices feature several favourable qualities for application as radiation dosimeters. MOSFETs are feature a compact sensitive volume that is scalable down to the sub-millimetre scale. As such, the MOSFET can be used as both a precise point dose monitoring solution of can be used in high spatial resolution array detector solutions. MOSFETs are also quite cheap produce, are scalable to purpose and their designs are customisable for intended use. MOSFET dosimeters are typically highly sensitive to radiation and provide stable, reproducible outputs. Exposures experienced by MOSFET-type dosimeters accumulate, meaning the device provides a permanent record of exposure. The readout procedure for MOSFET dosimeters varies with design but readout solutions typically feature low power consumption, near-instantaneous dose reporting and readout can be repeated with no cooldown period required for dose verification or for providing real-time feedback to the operator. MOSFET devices are safe to apply to the skin, are waterproof, are easy to apply to the patient volume and real-time feedback can be used to prevent radiation injuries to sensitive at-risk tissues. Previous applications have seen these dosimeters applied both as skin dosimeters and in vivo dosimeters during radiotherapy procedures. There are some drawbacks to MOSFET dosimeter technologies that should be noted. MOSFET threshold voltages are sensitive to temperature dependence. An increase in temperature will cause an increase in the conductivity across the substrate material which results in a shift in the $I_D-V_G$ characteristics that can easily be mistaken for an increase in accumulated dose. This temperature effect means that for accurate dosimetric measurements MOSFETs
2.1. Introduction to MOSFET dosimetry solutions

Figure 2.1: Cross-section of a generic p-channel type MOSFET Dosimeter

Figure 2.2: $I_D-V_G$ Characteristics of a MOSFET Dosimeter before/after irradiation
need to reach temperature equilibrium with the ambient environment or readout needs to be corrected to compensate for temperature changes. Conventional MOSFETs also have a finite lifetime limited either by the maximum threshold voltage measurable by the readout equipment or by saturation of charge traps within the oxide layer. MOSFETs may also exhibit energy dependant responses, sensitivity can degrade as charges accumulate toward saturation point and may experience fading effects due to trap recombination events occurring after irradiation. Many of these disadvantages can be mitigated through the design of the electronics, the readout methodology and the packaging solution used.

2.1.3 The MO\textsuperscript{Skin} Dosimeter

The MO\textsuperscript{Skin} dosimeter, shown in Figure 2.3 with the Clinical Semiconductor Dosimetry System, is a p-channel MOSFET-type dosimeter capable of:

- providing real-time dosimetric output
- performing in-built temperature compensation
- measuring dose at a tissue equivalent depth of 0.07 mm\textsuperscript{[94]}

The MO\textsuperscript{Skin} encapsulation has been designed to measure radiation doses delivered to the 0.07 mm measurement depth, enabling the measurement of representative skin and surface dose values. The ICRP recommend monitoring dose at this depth to monitor radiation exposure to the skin, hands, feet and eye lens. This depth was first recommended in ICRU Report 39 where three important measurement depths were identified, the 0.07 mm, 3 mm and 10 mm tissue depths\textsuperscript{[3,97]}. Dose delivered to these depths denoted as the dose quantities $H_p(0.07)$, $H_p(3)$ and $H_p(10)$ respectively. The 0.07 mm measurement depth is representative of damage delivered to the first layer of
radiation sensitive skin, the basal layer, meaning the MOSkin dosimeter, by design, is ideal for skin dosimetry applications[3].

The MOSkin dosimeter has been used to measure skin and in vivo surface doses delivered during radiotherapeutic procedures including external beam radiotherapy and brachytherapy[94, 98 - 101]. During these procedures, MOSkin has been used to measure skin and surface doses, including doses delivered to the rectum wall during brachytherapy, and has been noted to induce minimal perturbation on introduction to the field-of-view (FOV) of incident x-ray beams, even less so than TLDs, and as such, causes minimal dose enhancement effect to surrounding tissue[101]. The MOSkin has been calibrated for diagnostic range exposures. Initial diagnostic calibration studies were performed by Lian et.al. who calibrated the MOSkin for CT applications using a kilovoltage orthovoltage machine and simulated the MOSkin response to monoenergetic kilovoltage range photons using Monte Carlo code[95, 102]. Further studies were performed by Safari et.al. who developed calibration methodologies for applying the MOSkin dosimeter as an eye lens dosimeter in neuro-angiographic applications. These methodologies utilised the RQR Series of standard radiation protection beam qualities[103-106].

2.1.4 The MOSkin Readout System: Clinical Semiconductor Dosimetry System

The Clinical Semiconductor Dosimetry System (CSDS) is the traditional readout system for the MOSkin dosimeter developed by CMRP. This system is shown in Figure 2.3. This reader is battery powered and can perform readout of up to 10 MOSkin dosimeters simultaneously. During irradiation, the CSDS applies a 15 V bias across the gate oxide to encourage charge migration. This increases the sensitivity of the MOSkin to ionising radiation and ensures linear response to incident radiation. Read-
2.1. Introduction to MOSFET dosimetry solutions

Figure 2.3: Photograph of the MOSkin dosimeter with Clinical Semiconductor Dosimetry System

Figure 2.4: Photograph of the MOSkin dosimeter with the wireless OneTouch System
out of the MOSkins is performed by pulsing specific current from source to drain, \( V_{S,D} \), which induces a measurable gate voltage referred to as the threshold voltage, or \( V_{Th} \). This process can either be initiated manually by pressing a button on the system with gate voltage displayed via the LCD screen or automatically at a set rate by a logging system connected via serial port. When using the CSDS it is recommended to allow the MOSkin dosimeters to reach equilibrium at ambient temperature before measurement. This was the readout system used for the duration of this research thesis.

2.1.5 The MOSkin Readout System: OneTouch Dosimetry System

The latest iteration in MOSkin readout technology is the OneTouch readout system developed by CMRP\[^{[107]}\]. This system is shown in Figure 2.4. The OneTouch system is powered by battery and can perform readout of up to six MOSkin dosimeters simultaneously via micro-USB. The OneTouch system is a wireless readout system that can connect to a nearby computer through either Bluetooth or ultra-high frequency bandwidth connection. During irradiation, the OneTouch reader applies a 9V bias across the gate oxide as opposed to the traditional CSDS reader which applied a 15V bias. This reduced bias reduces the sensitivity of the MOSkin but improves the lifetime for radiotherapeutic applications. Readout is performed similarly to the CSDS and readout values are logged at a rate specified in the user interface. The user interface can display readout both as an accumulated dose value and as a percentage of a target dose value. The OneTouch MOSkin reader records threshold voltage readout up to microvolt scale precision and provides stable readout values in the sub-millivolt range. The OneTouch system also features a temperature compensation system. The OneTouch system calibrates temperature effect by measuring the ratio between the change
in source-to-gate voltage, $\Delta V_{S-G}$, and the change in source-to-substrate voltage, $\Delta V_{S-SS}$, for two temperature points. This novel methodology was developed by CMRP.

The calibration process, expressed formulaically in Equation 2.1, is performed at ambient room temperature and at an increased temperature induced simply by touching the head of the MOSkin pigtail. This built-in temperature compensation process reduces time and setup effort when compared to conventional MOSFET dosimeters that need to stabilise to the ambient temperature before use and it reduces potential sources of readout error that can be made during use.

2.2 Methodologies for Diagnostic Characterisation of the MOSkin Dosimeter

2.2.1 Lifetime Sensitivity of the MOSkin Dosimeter

Previous studies have observed that the sensitivity of the MOSkin dosimeter and other MOSFET-type dosimeters can decrease slightly over the lifetime of the device\textsuperscript{108}. To assess the lifetime sensitivity of the MOSkin, five dosimeters were irradiated using a Varian 21EX 6MV linear accelerator based at the Illawarra Cancer Care Centre. The dosimeters were embedded within a $20 \times 20 \times 20$ cm$^3$ solid water phantom at a depth of 1.5 cm where maximum dose to the phantom volume is imparted. The MOSkins were irradiated to a total dose of 30 Gy as delivered in 1 Gy increments. Readout was performed 30 seconds after each irradiation. Sensitivity of the MOSkin was calculated for each irradiation as the response per unit of dose delivered, as formulaically represented in Equation 2.1.

$$MOSkin\ Sensitivity = \frac{\Delta V_{Th}}{\Delta D}, \frac{V}{Gy}$$ (2.1)
2.2. Methodologies for Diagnostic Characterisation of the MOSkin Dosimeter

2.2.2 MOSkin Characterisation Using Clinical C-arm Beam Qualities

The linearity of the MOSkin dosimeter response in low dose pulsed beam qualities was tested using a Toshiba Infinix c-arm system based at Royal North Shore Hospital. During these measurements, the patient volume was simulated using a $30 \times 30 \times 30\text{cm}^3$ Perspex Polymethyl Acrylic (PMMA) phantom. The phantom was positioned on the patient couch which was then positioned so that the centre of the phantom was aligned with the c-arm gantry’s rotational isocentre. All measurements were performed with the c-arm gantry set to the posterior-anterior imaging projection, that is, the system was set to the default $0^\circ$ LAO/RAO (Left Anterior Oblique/Right Anterior Oblique), $0^\circ$ CRA/CAU (Cranial/Caudal) angulation. During these measurements, the c-arm system source to image-receptor distance (SID) was set to 100 cm while the FOV was set to an orthogonal field size (OFS) of 12 cm. This apparatus setup is diagrammatically represented in Figure 2.5. Three MOSkin dosimeters were fixed to the underside of the phantom and positioned to a point central to the c-arm system’s FOV. Irradiation was performed with the c-arm system operated in service mode by Toshiba engineers. Through service mode the c-arm beam characteristics could be set manually to fixed values. While the phantom was in field and while using clinically relevant imaging protocols, the c-arm system automatically selected a peak x-ray tube voltage of 93 kVp and a pulse width of 8 ms while a frame rate of 15 frames per seconds (FPS) was specified by the imaging protocol. These values were maintained manually for each irradiation. The MOSkin dose rate linearity was assessed using x-ray tube current values of 10, 20, 30, 50, 80, 100, 125, 160 and 200 mA. The irradiations were performed using approximately 2 minutes of beam time and readout was performed 30 seconds after the irradiation. Beam times were recorded accurately by dividing the total of imaging frames acquired by frame rate of 15 FPS. The MOSkin responses were
presented as an averaged response value normalised to time. Error for all averaged MOSkin response values was estimated by generating a 95% confidence interval for each measurement point. These confidence intervals were calculated using the standard deviation of the MOSkin responses and is represented formulaically in Equation 2.2.

Preliminary MOSkin calibration factors for clinical beam lines were established using the Philips Allura Clarity FD20 and a GE Innova 2100IQ c-arm systems based at Sutherland Heart Clinic. The calibration was performed by comparing the response of the MOSkin dosimeter and EBT2 GAFChromic film after irradiation of the dosimeters using a total of sixteen clinically-used c-arm beam qualities. EBT2 film was chosen as the gold standard dosimeter for this study due to its energy independence at kilovoltage range beam qualities\(^{[109,110]}\).

\[ CI_{95\%} = 1.96 \times \sigma_x \]  

\textbf{Figure 2.5:} PMMA Phantom with MOSkin attached to underside positioned at the c-arm system rotational isocenter
For the purposes of these experiments, the film was cut into $3 \times 3 \text{ cm}^2$ portions. Each piece of film was scanned before and after an irradiation. Scanning of the film was performed at least 24-hours after each irradiation using a Microtek ScanMaker i800 scanner. This model of scanner can produce images with optical resolutions of up to $9600 \times 4800$ dots per inch. Scanning is performed using a charge coupled device scanning element and output images are produced in 48-bit colour. Scanning was performed at an optical resolution of 300 dots per inch in the reflective scanning mode after allowing for a 10-minute warmup period. To minimise scanning artefacts and variability, positioning of the film was consistent for each portion of film and scanning was repeated 3 times per portion of film. Analysis of the optical density of the film was performed using the FIJI software package (Laboratory for Optical and Computational Instrumentation (LOCI) at the University of Wisconsin-Madison, https://fiji.sc/). The red channel of the film images was isolated for optical density assessment as this channel provides the highest response for EBT2 film[110]. Optical density was assessed over a region of interest of area $1 \times 1 \text{ cm}^2$ which was positioned at a point central to the film area.

Prior to the measurements at Sutherland Heart Clinic, the EBT2 film was calibrated using a Gulmay DX3300 orthovoltage unit based at the Illawarra Cancer Care Centre. The film was placed on the surface of a solid water phantom which was used to simulated backscatter conditions. Film portions were irradiated to doses ranging from 0 cGy to 140 cGy. Readout was performed 2 days after the irradiations were completed.

As per the measurements performed to assess dose rate linearity, the MOSkin clinical calibration was performed using the $30 \times 30 \times 30 \text{ cm}^3$ PMMA phantom aligned to rotational isocentre and with the c-arm gantry set to the posterior-anterior imaging projection. The dosimeters were positioned central to the FOV during measurements.
Irradiation of the EBT2 films were performed separately to the irradiations performed using the MOSkin dosimeters. Irradiations were performed using the low dose protocols used routinely by clinicians at Sutherland Heart Clinic. Measurements were performed using two SID settings, 100 cm and 110 cm and were performed in both the high dose pulsed acquisition mode and the lower dose continuous fluoroscopy mode. The Philips machine was operated using the clinical “CardLow” protocol. This protocol was developed as a low dose clinical protocol in collaboration with Sutherland Heart Clinic and used 5 mm of aluminium equivalent (mmAl) beam filtration when imaging in acquisition mode and 2 mmAl beam filtration when imaging in fluoroscopy mode. The GE system was operated using the “RDLS” protocol. This protocol was a variant of an internationally developed low dose standard protocol and used 5 mmAl beam filtration when imaging in acquisition mode and 2 mmAl beam filtration when imaging in fluoroscopy mode. Each c-arm system was surveyed using three FOV settings. The Philips system measures field of view using the diagonal field size (DFS). The Philips system measurements were performed using the 15 cm, 20 cm and 25 cm field sizes. The GE system measures FOV using the OFS. The GE system measurements were performed using the 10 cm, 12 cm and 17 cm OFS field sizes. The differences in FOV settings between these machines may contribute to differences in x-ray tube characteristics chosen by the c-arm systems. That said, FOV settings with comparable projected beam area were chosen to provide comparable irradiation conditions and Table 2.1 provides a direct comparison of each c-arm systems FOV settings, their associated projected beam area and the ratio of beam areas. The variety of settings chosen embody a large range of c-arm beam qualities and are representative of the clinical conditions most commonly observed during procedures by the clinicians at Sutherland Heart Clinic. To compare the MOSkin and film responses, dosimeter readout was normalised to irradiation time in units of minutes. In acquisition mode,
time of irradiation was recorded by dividing the total imaging frames acquired during each irradiation by the frame rate setting in use while in fluoroscopy mode time was recorded as measured by the c-arm system. The targeted duration of irradiation for the MOSkin dosimeters was 30 seconds in acquisition mode and 2 minutes in fluoroscopy mode. The EBT2 film portions were irradiated for approximately 50 seconds in both acquisition and fluoroscopy mode. The irradiation times selected were necessary to provide greater statistical certainty. MOSkin calibration factors were calculated as the ratio of the shift in MOSkin threshold voltage per minute to dose delivered per minute as measured by the EBT2 film.

\[
CF_X = \frac{kV_p}{\sum_{n=1}^{\text{Counts}_n \times \text{Response}_n, V}} \times \text{Gy}
\]  

(2.3)

<table>
<thead>
<tr>
<th>C-arm Settings</th>
<th>Philips System (A_{Ph} (cm^2))</th>
<th>GE System (A_{GE} (cm^2))</th>
<th>Ratio of FOVs A_{Ph}/A_{GE}</th>
</tr>
</thead>
<tbody>
<tr>
<td>Field Size 1 (OFS_{Ph} = 15 cm, OFS_{GE} = 10 cm)</td>
<td>112.5</td>
<td>100</td>
<td>1.125</td>
</tr>
<tr>
<td>Field Size 2 (DFS_{Ph} = 20 cm, OFS_{GE} = 15 cm)</td>
<td>200</td>
<td>225</td>
<td>0.889</td>
</tr>
<tr>
<td>Field Size 3 (DFS_{Ph} = 25 cm, OFS_{GE} = 17 cm)</td>
<td>312.5</td>
<td>289</td>
<td>1.081</td>
</tr>
</tbody>
</table>

**Table 2.1:** Comparison of the Beam Area produced at Image Receptor by Diagonal Field Settings and Orthogonal Field Settings
<table>
<thead>
<tr>
<th>Narrow-Series Beam Quality</th>
<th>X-ray Tube Settings</th>
<th>Beam Filtration Material</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Peak Voltage (kVp)</td>
<td>Mean Energy (keV)</td>
</tr>
<tr>
<td>$N_{10}$</td>
<td>10</td>
<td>8</td>
</tr>
<tr>
<td>$N_{15}$</td>
<td>15</td>
<td>12</td>
</tr>
<tr>
<td>$N_{20}$</td>
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<td>16</td>
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<td>20</td>
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<td>24</td>
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<td>$N_{40}$</td>
<td>40</td>
<td>33</td>
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<td>$N_{60}$</td>
<td>60</td>
<td>48</td>
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<td>65</td>
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<td>250</td>
<td>208</td>
</tr>
<tr>
<td>$N_{300}$</td>
<td>300</td>
<td>250</td>
</tr>
</tbody>
</table>

Table 2.2: Narrow Spectrum Series Beam Qualities as specified by ISO 4037-1 [111]
### 2.2 Methodologies for Diagnostic Characterisation of the MOSFet Dosimeter

#### Table 2.3: RQR Spectrum Series Beam Qualities as specified by IEC 61267

<table>
<thead>
<tr>
<th>RQR-Series</th>
<th>X-ray Tube Settings</th>
<th>Beam Filtration Material</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Peak Voltage (kVp)</td>
<td>Mean Energy (keV)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Aluminium (mmAl)</td>
</tr>
<tr>
<td>RQR_2</td>
<td>40</td>
<td>27.54</td>
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<tr>
<td>RQR_3</td>
<td>50</td>
<td>31.97</td>
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<tr>
<td>RQR_4</td>
<td>60</td>
<td>36.37</td>
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<tr>
<td>RQR_5</td>
<td>70</td>
<td>40.45</td>
</tr>
<tr>
<td>RQR_6</td>
<td>80</td>
<td>44.10</td>
</tr>
<tr>
<td>RQR_7</td>
<td>90</td>
<td>47.85</td>
</tr>
<tr>
<td>RQR_8</td>
<td>100</td>
<td>50.82</td>
</tr>
<tr>
<td>RQR_9</td>
<td>120</td>
<td>56.47</td>
</tr>
<tr>
<td>RQR_10</td>
<td>150</td>
<td>63.67</td>
</tr>
</tbody>
</table>
2.2.3 Simulated MOSkin Energy Dependence for Diagnostic X-Ray Spectra

Depending on model and protocol, the x-ray beam characteristics and x-ray tube filtration may vary between c-arm systems. Through simulation, the MOSkin calibration was expanded to include a broader range of potential c-arm beam qualities. This was intended to ensure that an appropriate calibration factor would be available during measurements using any c-arm system and c-arm configuration. Clinically representative beam spectra were generated using methodology outlined by Boone and Seibert\cite{113}. This methodology was used to calculate the energy spectra based on the x-ray tube potential and beam filtration settings. Using this method, individual spectra were generated for beam potentials ranging from 60 to 120 kVp in 10 kVp intervals using four filtration settings: 0 mm, 2 mm, 5 mm and 10 mm of aluminium equivalent material. Each simulation was performed to sufficient photon count to ensure that error was within 2% of the total photon count. The spectra were normalised to the integral of the spectrum’s counts. Data detailing the MOSkin relative response to monoenergetic diagnostic range photons was acquired from a simulation study performed by Lian et.al.\cite{102}. The product of MOSkin response and the photon count was summated for each individual monoenergetic photon bin using Equation 2.3. This procedure was performed on all simulated spectra to produce a relative MOSkin response to each of the simulated x-ray tube potentials which were then scaled using the results of the clinical system calibration through comparison to film response performed in Section 2.3.2.
2.2. Methodologies for Diagnostic Characterisation of the MOSkin Dosimeter

2.2.4 MOSkin Energy and Angular Dependence Using Standard Diagnostic Beam Qualities

Further characterisation of the MOSkin dosimeters’ energy dependence was performed by observing the MOSkin response to irradiation using standard radiation protection beam qualities. The irradiations were performed at the calibration facilities of the Studiecentrum voor Kernenergie; Centre d’Étude de l’énergie Nucléaire (SCK-CEN). Using these facilities, the MOSkin response to the Narrow Spectrum Series and RQR Series of beam qualities was assessed. The Narrow Spectrum Series of beam qualities was defined by ISO in ISO report 4037-1\cite{111}. Designed to represent the scatter fields produced by clinical x-ray sources, this series of beam qualities is typically used to characterise radiation protective equipment and radiation monitoring equipment. The RQR Series of beam qualities was developed by the IEC to reproduce clinical diagnostic beam qualities\cite{112}. These beam qualities were produced using an Xstrahl dual x-ray tube system equipped with a 100 keV and a 300 keV x-ray tube using a variety of filtration settings. The x-ray tube potentials and x-ray tube filtration settings for each of the Narrow Spectrum Series beam qualities as specified by ISO are presented in Table 2.2, while the settings for the RQR Series beam qualities as specified by IEC are presented in Table 2.3.

Measurements were performed using the 20 × 20 cm² cylindrical PMMA water-filled phantom. This phantom was designed as an outcome of the ORAMED (Optimization of RAdition protection for MEDical staff) study and was intended to represent an operator or patient head as recommended by the ICRP in ICRP Publication 139 for the purposes of calibrating eye lens dosimeters\cite{61, 114}. This phantom was chosen to further assess the MOSkin dosimeter for applications in operator dosimetry and in neuro-angiographic procedures. The phantom was placed 100 cm from the irradiation source to ensure uniform irradiation of the entire phantom volume, as shown
in **Figure 2.6**. Four MOSkin dosimeters were fixed to the surface of the phantom volume along the central axis of the front phantom surface in varying orientations. This configuration is represented diagrammatically in **Figure 2.7**.

The MOSkin energy dependence was assessed using the full series of beam qualities as delivered at normal incidence. The dose delivered to the MOSkin dosimeters during irradiations was calculated using both the air Kerma at the measurement point and air Kerma conversion coefficients as specified in ISO 4037-3\[^{115}\]. The air Kerma delivered to the measurement point was measured in free air prior to each measurement using a 1 litre PTW ionization chamber. The air Kerma values provided by the ionisation chamber were converted into $H_p(0.07)$, $H_p(3)$ and $H_p(10)$ dose quantities using **Equation 2.4**. Air Kerma conversion coefficients for the cylindrical phantom volume existed within the literature for the $H_p(3)$ dose quantity, however, the conversion coefficients for the $H_p(0.07)$ measurement depth were unavailable for the cylindrical phantom\[^{116,117}\]. As such, the missing conversion coefficients were simulated using the simulation framework developed in the previous section. Spectra for the Narrow Series and RQR Series beam qualities were input as published by Physikalisch-Technische Bundesanstalt in their DOS-34 report and using similar methodology to those employed by Grosswendt and Ginjaume et.al.\[^{118 - 120}\]. The geometry and positioning of scoring volumes used for these simulations has been represented diagrammatically in **Figure 2.8**. The simulated conversion coefficients at normal incidence was compared to the existing published air Kerma conversion coefficients for the cylindrical and slab phantoms as reported by Behrens et.al., Principi et.al. and Ankerhold et.al.\[^{116 - 118}\].

$$H_p(x) = \frac{\text{Air Kerma}}{\text{Conversion Coefficient}_x}, \text{ Gy}$$  \hspace{1cm} (2.4)
2.2. Methodologies for Diagnostic Characterisation of the MOSkin Dosimeter

Figure 2.6: Irradiation apparatus for characterisation of the MOSkin dosimeter using standardised beam qualities using cylindrical PMMA water phantom positioned with measurement point 100 cm from x-ray source.

Figure 2.7: MOSkin dosimeter positioning on cylindrical phantom volume presented with a) phantom at 0° orientation and b) phantom at 90° orientation. Φ and θ angular axes marked on phantom and c) with respect to the MOSkin pigtail.
2.2. Methodologies for Diagnostic Characterisation of the MOSkin Dosimeter

Figure 2.8: Diagram representing the scoring volumes used during air Kerma simulations as positioned in a) free air and b) with the phantom volume generated (yellow represents $H_P(0.07)$, green represents $H_P(3)$, purple represents $H_P(10)$ and grey arrows represent uniform incident radiation beam)

Figure 2.9: Irradiation apparatus for measuring angular response of the MOSkin dosimeter to standardised beam qualities at a fixed source to dosimeter distance of 1 meter (blue circle represents phantom in 0° incidence configuration, yellow circle represents phantom in 45° incidence configuration and orange circle represents phantom in 90° incidence configuration)
2.3. Results of the MOSkin Characterisation Study

The response of the MOSkin dosimeter when irradiated from a range of incident angles was assessed. Angle of incidence was adjusted by rotating the cylindrical phantom about the MOSkin measurement point while maintaining a source to dosimeter distance of 1 meter for all phantom configurations. This rotation method is presented diagrammatically in Figure 2.9. The dosimeters were irradiated over 7 phantom configurations that covered the 0° to 90° angular range in 15° increments. Dose delivery was measured using four MOSkin dosimeters aligned vertically at the measurement point. Each MOSkin dosimeter pigtail was oriented in a different direction. This enabled characterisation across a 180° degree over two rotational axes denoted as the Φ axis and the θ axis. This dosimeter configuration has been presented in Figure 2.7 with an explicit definition of the Φ and θ axes relative to the MOSkin pigtail in Figure 2.7c. Dose delivered to the measurement point in terms of $H_p(0.07)$, $H_p(3)$ and $H_p(10)$ dose quantities as calculated using air Kerma conversion coefficients specified by Behrens et.al., Principi et.al. and Ankerhold et.al.\cite{116-118}. As the cylindrical phantom is relatively new some air Kerma conversion coefficients required simulation using the methodology specified earlier in this section. MOSkin response to varying incident irradiation angles was assessed using the Narrow\textsubscript{80} and RQR\textsubscript{6} beam qualities. These beam qualities were chosen as they are representative of the irradiation conditions present in catheterisation laboratories\cite{80,121}.

2.3 Results of the MOSkin Characterisation Study

2.3.1 Lifetime Linearity of MOSkin Sensitivity

The sensitivity of the MOSkin decreased as the dosimeters accumulated exposure as shown in Figure 2.10. The average reduction in MOSkin sensitivity over the 30 Gy irradiation period was 9.69% which can also be represented as 0.334% /Gy. The reduc-
tion in MOSkin sensitivity was also consequently observed to decrease linearly with threshold voltage. When considering MOSkin sensitivity as a function of threshold voltage, it was observed that all irradiated MOSkins behaved to a common linear trend which has been presented in Figure 2.11 and expressed in Equation 2.5. Regression analysis of this linear trendline yielded an R-squared value of 0.97. All measurement points were observed to be within 1.07% of the expected value based on this trendline and therefore this data can be used to correct MOSkin desensitisation by inputting the initial and measured threshold voltages.

2.3.2 Investigation of MOSkin response to clinical c-arm beam qualities

When exposed to a consistent 93 kVp beam quality, the MOSkin dosimeter response was observed to increase linearly with increasing x-ray tube current, as shown in Figure 2.12. A linear trendline was established and using regression analysis, this trendline yielded an R-squared value of 0.98. Error was assessed by generating a 95% confidence interval using the standard deviation of the MOSkin responses. The MOSkin response was highly reproducible with individual responses varying by a maximum of 5% of the averaged value for the lowest dose settings.

The EBT2 film portions darkened proportionally with the total dose imparted to the film. Linear trendlines were established for each individual beam quality assessed and for a combined dataset comprised of data from all beam qualities assessed. Regression analysis yielded R-squared values of 0.992, 0.988, 0.992, 0.992 and 0.973 for the 50 kVp, 75 kVp, 100 kVp, 150 kVp and the combined datasets respectively. The film response error was established by generating a 95% confidence interval based on the standard deviation of pixel values within the region of interest. All film portions observed individual error of within 10% and the majority of the film portions were
2.3. Results of the MOSkin Characterisation Study

Figure 2.10: Lifetime Dependence of MOSkin Sensitivity on Accumulated Absorbed Dose using 6MV Linear Accelerator to a Total Dose of 30 Gy

Figure 2.11: MOSkin Sensitivity Expressed as a Function of Threshold Voltage

\[
\text{MOSkinSensitivity} = (-3.33 \times V_{Th} + 317.48) \text{ mV/Gy}
\]
within 10% of the combined dataset trendline values. The calibration trends with calculated error values applied as presented in Figure 2.13. This error is potentially related to minor energy dependence of EBT2 film in this x-ray energy range.

MOSkin calibration factors were established for the Philips Allura Xper FD20 and GE Innova 2100IQ c-arm systems based at Sutherland Heart Clinic. The Philips Allura Xper FD20 c-arm system produced x-ray tube voltages ranging from 82 to 120 kVp. The average MOSkin calibration factors observed on this machine were 0.737 ± 0.036 mV/mGy for acquisition mode and 0.691 ± 0.014 mV/mGy for fluoroscopy mode. The GE Innova 2100IQ c-arm system produced x-ray tube voltages ranging from 97 to 120 kVp. The average MOSkin calibration factors observed on this machine were 0.738 ± 0.032 mV/mGy for acquisition mode and 0.711 ± 0.083 mV/mGy for fluoroscopy mode. The full range of calibration factors and x-ray beam potentials for both machines in all configurations is provided in Table 2.4 to Table 2.7

2.3.3 Simulated MOSkin Energy Dependence for Diagnostic X-ray Spectra

The simulated c-arm beam spectra are presented graphically in Figure 2.14. The energy dependence of the MOSkin to monoenergetic photons as simulated by Lian et.al. is presented in Figure 2.15\textsuperscript{[102]}. A logistic trendline was fit to the data. Figure 2.16 represents the expected MOSkin response to the simulated beam qualities relative to the MOSkin response to monoenergetic 300 keV photons. A quadratic trendline was applied for each filtration setting. This trend is represented in Equation 2.7 and the relevant coefficients for each filtration setting is supplied in Table 2.8. The MOSkin response values were then scaled using the clinical calibration data points measured and presented in Section 2.3.2. The data point used was the 0.737 mV/mGy data point which was measured when using the Philips Allura Xper c-arm system with an
### Results of the MO Skin Characterisation Study

#### Philips Allura Xper Acquisition Mode

<table>
<thead>
<tr>
<th>SID:100 cm DFS:15 cm</th>
<th>Peak Beam Voltage kVp</th>
<th>MOSkin Response mV/mm</th>
<th>EBT2 Response mGy/mm</th>
<th>MOSkin CF mV/mGy</th>
</tr>
</thead>
<tbody>
<tr>
<td>106</td>
<td>262.1</td>
<td>359</td>
<td>0.730</td>
<td></td>
</tr>
<tr>
<td>SID:110 cm DFS:15 cm</td>
<td>116</td>
<td>331.8</td>
<td>462</td>
<td>0.718</td>
</tr>
<tr>
<td>SID:100 cm DFS:20 cm</td>
<td>103</td>
<td>250.2</td>
<td>339.6</td>
<td>0.737</td>
</tr>
<tr>
<td>SID:110 cm DFS:20 cm</td>
<td>107</td>
<td>294</td>
<td>403.3</td>
<td>0.729</td>
</tr>
<tr>
<td>SID:100 cm DFS:25 cm</td>
<td>98</td>
<td>216.8</td>
<td>290.1</td>
<td>0.747</td>
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<td>SID:110 cm DFS:25 cm</td>
<td>93</td>
<td>267.6</td>
<td>351.2</td>
<td>0.762</td>
</tr>
</tbody>
</table>

Average CF: $0.737 \pm 0.036$

**Table 2.4:** Calibration of the MOSkin dosimeter using the Philips Allura Xper FD20 c-arm system in acquisition mode

#### Philips Allura Xper Fluoroscopy Mode

<table>
<thead>
<tr>
<th>SID:100 cm DFS:15 cm</th>
<th>Peak Beam Voltage kVp</th>
<th>MOSkin Response mV/mm</th>
<th>EBT2 Response mGy/mm</th>
<th>MOSkin CF mV/mGy</th>
</tr>
</thead>
<tbody>
<tr>
<td>109</td>
<td>22.50</td>
<td>33.25</td>
<td>0.677</td>
<td></td>
</tr>
<tr>
<td>SID:110 cm DFS:15 cm</td>
<td>120</td>
<td>28.61</td>
<td>52.60</td>
<td>0.544</td>
</tr>
<tr>
<td>SID:100 cm DFS:20 cm</td>
<td>84</td>
<td>18.29</td>
<td>25.11</td>
<td>0.728</td>
</tr>
<tr>
<td>SID:110 cm DFS:20 cm</td>
<td>91</td>
<td>26.52</td>
<td>38.19</td>
<td>0.694</td>
</tr>
<tr>
<td>SID:100 cm DFS:25 cm</td>
<td>82</td>
<td>13.98</td>
<td>18.45</td>
<td>0.757</td>
</tr>
<tr>
<td>SID:110 cm DFS:25 cm</td>
<td>86</td>
<td>18.62</td>
<td>25.03</td>
<td>0.744</td>
</tr>
</tbody>
</table>

Average CF: $0.691 \pm 0.140$

**Table 2.5:** Calibration of the MOSkin dosimeter using the Philips Allura Xper FD20 c-arm system in fluoroscopy mode
### 2.3. Results of the MOSkin Characterisation Study

#### Table 2.6: Calibration of the MOSkin dosimeter using the GE Innova 2100IQ c-arm system in acquisition mode

<table>
<thead>
<tr>
<th>GE Innova 2100IQ Acquisition Mode</th>
<th>Peak Beam Voltage kVp</th>
<th>MOSkin Response mV/mm</th>
<th>EBT2 Response mGy/min</th>
<th>MOSkin CF mV/mGy</th>
</tr>
</thead>
<tbody>
<tr>
<td>SID:100 cm OFS:12 cm</td>
<td>97</td>
<td>259.1</td>
<td>347.4</td>
<td>0.746</td>
</tr>
<tr>
<td>SID:110 cm OFS:12 cm</td>
<td>105</td>
<td>326.0</td>
<td>445.1</td>
<td>0.732</td>
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<td>SID:100 cm OFS:15 cm</td>
<td>100</td>
<td>245.7</td>
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</tr>
<tr>
<td>SID:110 cm OFS:15 cm</td>
<td>104</td>
<td>306.3</td>
<td>411.5</td>
<td>0.744</td>
</tr>
<tr>
<td>SID:100 cm OFS:17 cm</td>
<td>110</td>
<td>212.9</td>
<td>291.7</td>
<td>0.730</td>
</tr>
<tr>
<td>SID:110 cm OFS:17 cm</td>
<td>113</td>
<td>274.8</td>
<td>375.6</td>
<td>0.732</td>
</tr>
</tbody>
</table>

Average CF: 0.738 ± 0.032

#### Table 2.7: Calibration of the MOSkin dosimeter using the GE Innova 2100IQ c-arm system in fluoroscopy mode

<table>
<thead>
<tr>
<th>GE Innova 2100IQ Fluoroscopy Mode</th>
<th>Peak Beam Voltage kVp</th>
<th>MOSkin Response mV/mm</th>
<th>EBT2 Response mGy/min</th>
<th>MOSkin CF mV/mGy</th>
</tr>
</thead>
<tbody>
<tr>
<td>SID:100 cm OFS:12 cm</td>
<td>105</td>
<td>26.51</td>
<td>39.44</td>
<td>0.672</td>
</tr>
<tr>
<td>SID:110 cm OFS:12 cm</td>
<td>115</td>
<td>32.76</td>
<td>47.07</td>
<td>0.696</td>
</tr>
<tr>
<td>SID:100 cm OFS:15 cm</td>
<td>108</td>
<td>23.58</td>
<td>30.90</td>
<td>0.763</td>
</tr>
<tr>
<td>SID:110 cm OFS:15 cm</td>
<td>117</td>
<td>29.19</td>
<td>43.02</td>
<td>0.678</td>
</tr>
<tr>
<td>SID:100 cm OFS:17 cm</td>
<td>110</td>
<td>19.73</td>
<td>28.98</td>
<td>0.681</td>
</tr>
<tr>
<td>SID:110 cm OFS:17 cm</td>
<td>120</td>
<td>23.93</td>
<td>30.78</td>
<td>0.777</td>
</tr>
</tbody>
</table>

Average CF: 0.711 ± 0.083
2.3. Results of the MOSkin Characterisation Study

x-ray tube potential of 103 kVp and with 5mmAl of beam filtration. The data was then extrapolated to provide MOSkin calibration factors for beam qualities from 80 to 120 kVp for filtration values from 1 to 10 mmAl in 1 mmAl increments. This data is presented graphically in Figure 2.17 where it is also compared to the MOSkin clinical calibration datapoints.
2.3. Results of the MOSkin Characterisation Study

Figure 2.12: MOSkin Dose Rate Linearity using a Toshiba Infinix C-arm System producing a consistent beam quality at 93 kVp x-ray tube voltage.

Figure 2.13: Energy and dose rate dependence of EBT2 film using 50, 70, 100 and 150 kVp orthovoltage beam qualities for doses of up to 200 cGy.
2.3. Results of the MOSkin Characterisation Study

Figure 2.14: Photon Spectra for x-ray tube potentials ranging from 60 kVp to 120 kVp with (A) no filter, (B) 2mmAl filter, (C) 5mmAl filter and (D) 10mmAl filter settings at a fixed x-ray tube current of 1mAs.

Figure 2.15: MOSkin response to monoenergetic photons relative to the response at 300 keV as generated using Monte Carlo with logistics trendline fit to data.\textsuperscript{102}
2.3. Results of the MOSkin Characterisation Study

Figure 2.16: Relative MOSkin response to the simulated c-arm beam spectra

Figure 2.17: MOSkin calibration factors as established by scaling the relative MOSkin response data as scaled by and compared to clinically measured MOSkin Sensitivities
Simulated Calibration Factor = A + Bx + Cx^2 \tag{2.6}

<table>
<thead>
<tr>
<th>Beam Filtration</th>
<th>Coefficient A $\frac{mV}{cGy}$</th>
<th>Coefficient B $\frac{mV \cdot kV_p}{cGy}$</th>
<th>Coefficient C $\frac{mV \cdot kV_p^2}{cGy}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0 mmAl</td>
<td>5.07</td>
<td>-4.54 x 10^{-2}</td>
<td>1.79 x 10^{-4}</td>
</tr>
<tr>
<td>2 mmAl</td>
<td>4.33</td>
<td>-3.81 x 10^{-2}</td>
<td>1.55 x 10^{-4}</td>
</tr>
<tr>
<td>4 mmAl</td>
<td>3.79</td>
<td>-3.24 x 10^{-2}</td>
<td>1.36 x 10^{-4}</td>
</tr>
<tr>
<td>10 mmAl</td>
<td>3.33</td>
<td>-2.72 x 10^{-2}</td>
<td>1.16 x 10^{-4}</td>
</tr>
</tbody>
</table>

**Table 2.8:** Coefficients for quadratic trendlines that describe MOSkin behaviour during exposure to kilovoltage range beam qualities over four filtration values
2.3.4 MOSkin Energy Dependence Using Standard Diagnostic Beam Qualities

The simulated Narrow Spectrum Series and RQR Series air Kerma conversion coefficients compared favourably to previously published data. The energy dependence of the MOSkin dosimeter to the Narrow Series and RQR Series beam qualities is presented in Figure 2.18. A peak MOSkin energy response of 1.37 mV/mGy was observed for the native MOSkin measurement depth of 0.07 mm using the Narrow_{25} beam quality. The angular response of the MOSkin dosimeter for the Narrow_{80} and RQR_{6} beam qualities is presented in Figure 2.19 to Figure 2.22.

![Figure 2.18: Energy dependence of the MOSkin dosimeter at native measurement depth with respect to H_p(0.07), H_p(3) and H_p(10) dose delivered to measurement point](image)
2.3.5 MOSkin response to varying angle of irradiation incidence using standard diagnostic beam qualities

The MOSkin response to varying angle of irradiation incidence has been presented in Figure 2.19a for the Narrow\textsubscript{80} beam quality and Figure 2.19 b for the RQR\textsubscript{6} beam quality. The phantom orientations have been reiterated in Figure 2.19c for ease of interpretation. The MOSkin response was proportional to within 5\% of the calculated $H_p(0.07)$ dose for angles of up to 70\degree. Maximum deviation occurred at the 90\degree incidence angle where the MOSkin over responded to incident radiation by an average of 16\%.

*Figure 2.19:* MOSkin response to varying angle of incidence as irradiated using a) the Narrow\textsubscript{80} beam quality and b) the RQR\textsubscript{6} beam quality with angles simulated by c) rotating the cylindrical phantom volume.
2.4 MOSkin Characterisation Discussion

Over the course of this chapter and through the results obtained during these studies the MOSkin dosimeter behaviour has been comprehensively characterised for use in the catheterisation laboratory. Sensitivity of the MOSkin dosimeter was established for linear accelerator beam qualities, c-arm system generated x-ray beams and for standardised radiation protection beam qualities. In Table 2.9 and Table 2.10, the MOSkin calibration factors established during this study have been compared to and are in close agreement with relevant literature published values.

During the linear accelerator response testing, the sensitivity of the MOSkin dosimeters was observed to decrease with dose accumulation. Over the course of the 30 Gy irradiations, the sensitivity of the MOSkin dosimeters reduced by an average of 10.442%, or 0.334% per Gy. This figure is in close agreement with results published by Wei Loong Jong et.al., a study that reported that MOSkin sensitivity reduced by an average of 0.340% per Gy after a total of 30 Gy irradiation via a linear accelerator[108]. When correlating MOSkin sensitivity with the MOSkin threshold voltage recorded pre-irradiation, a common highly linear trend was identified that on regression analysis yielded an R-squared value of 0.97 and could be used to describe the combined sensitivity dataset to within 1.07% of the trendline predicted values. The average MOSkin sensitivity to these megavoltage exposures was $0.263 \pm 0.003 \text{ mV/Gy}$ which corresponded to a threshold voltage of 16.262 V. The total life span of the MOSkin dosimeter observed during this experiment was 9.640 V to 22.670 V or a total of $\sim 13$ V which represents a minimum lifespan of $\sim 17.5$ Gy when using the average angiographic sensitivity measured during the clinical characterisation, however, this lifetime may differ depending on irradiation source and on the initial threshold voltage values of the selected MOSkin batch. Desensitisation of the MOSkin dosimeter, if not corrected, could result in an underestimation in dose of at maximum 3.5% during a 2 Gy
2.4. MOSkin Characterisation Discussion

<table>
<thead>
<tr>
<th>Irradiation Source</th>
<th>MOSkin Calibration Factor</th>
<th>Source of Data</th>
</tr>
</thead>
<tbody>
<tr>
<td>6 MeV Linac</td>
<td>0.263 ± 0.001 mV/mGy</td>
<td>Harcastle et.al. (2010) [99]</td>
</tr>
<tr>
<td>6 MV TomoTherapy Unit</td>
<td>0.257 ± 0.006 mV/mGy</td>
<td>Alnaghi et.al. (2015) [122]</td>
</tr>
</tbody>
</table>

**Table 2.9:** Experimental and Published MOSkin calibration factors for radiotherapy megavoltage beam qualities

<table>
<thead>
<tr>
<th>Irradiation Source</th>
<th>MOSkin Calibration Factor</th>
<th>Source of Data</th>
</tr>
</thead>
<tbody>
<tr>
<td>103 kVp c-arm beam</td>
<td>0.737 ± 0.036 mV/mGy</td>
<td>Experimental data Section 2.3.2</td>
</tr>
<tr>
<td>150 keV Orthovoltage</td>
<td>0.670 mV/mGy</td>
<td>Lian et.al. (2013) [123]</td>
</tr>
<tr>
<td>RQR7 Beam Quality</td>
<td>1.127 ± 0.028 mV/mGy</td>
<td>Experimental data Section 2.3.4</td>
</tr>
<tr>
<td>RQR7 Beam Quality</td>
<td>1.156 ± 0.036 mV/mGy</td>
<td>Safari et.al. (2015) [103]</td>
</tr>
</tbody>
</table>

**Table 2.10:** Experimental and Published MOSkin calibration factors for diagnostic kilovoltage beam qualities

The MOskin dosimeter displayed favourable characteristics during the clinical evaluation of the dosimeter. The MOskin was observed to exhibit radiation transparency in all beam qualities observed. In Figure 2.20, the visibility, or lack thereof, of the MOskin dosimeter has been compared to two paperclips. The MOskin responses were highly reproducible to within 2% of an average measurement value and when exposed to a fixed x-ray tube current while incrementing current, the dosimeters response increased proportionally to beam current. This was observed even for lower dose rates and during both pulsed and continuous beam types.

The clinical calibration of the MOskin dosimeter was performed through comparison to EBT2 film as the study’s gold standard comparative dosimeter. The EBT2 film
2.4. **MOSkin Characterisation Discussion**

Figure 2.20: MOSkin radiation transparency and visibility as compared to paperclips on images acquired by the Phillips Allura Xper FD20 c-arm system[124]

exhibited low levels of energy and dose rate dependence during the orthovoltage calibration, which was consistent with the reference study performed by Butson et.al.[110]. There was some divergence of the calibration trendlines, all data points were within 10% value of the combined calibration trendline. These properties made EBT2 an ideal option for the initial clinical diagnostic calibration of the MOSkin dosimeter. Ultimately, the 100 kVp film calibration curve was chosen for performing readout of the film irradiated using the Sutherland Heart Clinic c-arm systems. This curve was deemed the most appropriate to use as the 100 kVp beam quality was the closest representative to the range beam qualities observed using the Sutherland Heart Clinic c-arm systems, that is, these c-arms operated within an x-ray tube energy range of 80 to 120 kVp. The energy dependence of EBT2 film was considered negligible in this range as both the 70 kVp and 150 kVp calibration curves were both within 10% of the 100 kVp calibration curve.
The MOSkin dosimeter was compared to EBT2 film over a total of sixteen beam qualities produced by the clinical c-arm systems using clinically-used protocols and settings. MOSkin calibration factors were established for each beam quality individually and were averaged by machine and imaging modality. The MOSkin sensitivity varied depending on machine, imaging modality, field size and general irradiation setup. This variation was less pronounced in acquisition mode as this mode utilised additional beam filtration and a narrower range of x-ray tube voltages than the fluoroscopy mode. The results indicate that while it would be possible to produce simple machine and protocol specific MOSkin calibration factors in acquisition mode without introducing significant levels of error, these calibration factors would not necessarily be transferable between machines or protocols and would most likely be unsuitable for fluoroscopy mode measurements. The results indicate that this form of calibration is not ideal for clinical use but nonetheless was a useful initial investigation for developing a more sophisticated methodology. Furthermore, while not relevant to the characterisation of the MOSkin dosimeter performed in this chapter, this study further illustrated the need for real-time dosimetry in catheterisation laboratories after an expanded phantom dosimetry survey with additional local c-arm systems revealed significant differences in dose delivery between different c-arm systems\cite{124}.

The MOSkin calibration was further developed through use of the simulation framework developed in Section 2.2.2. This framework allowed for MOSkin calibration factors to be estimated as a function of x-ray tube voltage and total beam filtration. The simulation results were validated through comparison to the clinically measured MOSkin calibration factors and close agreement between the datasets was established. There were limitations to this simulation, for instance, the c-arm x-ray tube filtration settings were represented in terms of equivalent thickness of aluminium when in reality the x-ray tube filtration was composed of a combination of aluminium and copper
material. This could affect the beam quality and result in some error introduced to the simulated spectra. These results effectively provide a transferable MOSkin calibration function that with proper input can be applied to any c-arm system and protocol within reason with relative ease.

When calibrating the MOSkin using standard diagnostic radiation protection beam qualities, the MOSkin was observed to experience energy dependence at low energies. At the measurement depth of 0.07 mm, the MOSkin energy dependence peaked at 1.37 mV/mGy when using the Narrow$_{25}$ beam quality. When considering the x-ray tube energy range observed at Sutherland Heart Clinic, 80 to 120 kVp, meaning the Narrow$_{80}$ to Narrow$_{120}$ and RQR$_6$ to RQR$_8$ beam qualities are potentially the closest representatives of the c-arm system beam qualities. Furthermore, previous studies in the literature have used the Narrow$_{80}$, RQR$_6$ and RQR$_7$ beam qualities to represent clinical c-arm system beam qualities$^{[80,121]}$. As such, it is reasonable to assume that the MOSkin energy dependence would result in MOSkin calibration factors ranging from 0.479 to 1.152 mV/mGy to be suitable for use in catheterisation laboratories. This range encapsulated all diagnostic range MOSkin calibration factors observed in this study and presented from the literature in Table 2.10.

The MOSkin angular response was modelled using a parabolic trendline of form $k \cdot x^2$ where $x$ is the angulation of the dosimeter and $k$ is a function of MOSkin orientation/positioning and the measurement depth, as presented in Equation 2.8. Using this formulation, $k$ was observed to increase with the simulated measurement depth used and as such an increase in the gradient of the trendline was observed.

A slight asymmetry was present across the dosimeter volume, resulting in an approximate offset of 3.5% in the parabolic trend observed between either side of the dosimeter, as presented in Figure 2.21. This difference was more pronounced for the $\Phi$ axis measurements than for the $\theta$ measurements. For the measurement depth of
2.4. MO\textit{Skin} Characterisation Discussion

Figure 2.21: MO\textit{Skin} response to standard beam qualities when varying a) $\theta$ axis angulation and b) $\Phi$ axis angulation with as simulated by c) rotating the cylindrical phantom volume.

0.07 mm, the MO\textit{Skin} dosimeter response was within 10\% of expected dose response for angles of up to 70° deviation from normal incidence. Maximum deviation was observed at 90° incidence resulting in MO\textit{Skin} over response of up to 20\%.

Finally, the MO\textit{Skin} dosimeter response was compared to dose values at different reference depths. These depths included the native $H_p(0.07)$ reference depth, the $H_p(3)$ reference depth and the $H_p(10)$ reference depth. For simplicity, the $\theta$ axis measurements were considered in isolation and each dataset was subsequently converted into a parabolic trendline of the form given in Equation 2.7. The resulting graphs have been presented in Figure 2.22 with k-values presented in Figure 2.22c. For each subsequent reference depth, the k-values increased resulting in a steeper trendline.
Despite this, values for the $H_P(3)$ reference depth were within 10% expected response for angles up to 30° deviation from normal and were within 20% expected response for angles up to 45° deviation from normal. $K$-values also did not vary significantly between the Narrow$_{80}$ and RQR$_6$ beam qualities. These results suggest that the MOSkin could be calibrated with respect to $H_P(3)$ reference depth and could reliably measure doses delivered from near-normal incidence. This option would not be practical for the $H_P(10)$ reference depth as the MOSkin angular response was too steep for this reference depth and because $k$-values deviated significantly based upon beam quality used. This was likely caused by the low energy photon contribution that the MOSkin dosimeter was subject to for each beam quality. At the $H_P(10)$ reference depth, this low energy contribution would not be present and as such extrapolating to this depth is complicated. Chapter 5 will consider the potential for MOSkin capping materials to enable measuring accurately at specifically the $H_P(3)$ reference depth.

$$MOSkin \text{ Angular Response } = kx^2 + 1 = f(p,d) \times A^2 + 1$$ (2.7)
Figure 2.22: Comparison of MOSkin response at reference depths $H_P(0.07)$, $H_P(3)$ and $H_P(10)$ during a) Narrow$_{00}$ and b) RQR$_6$ irradiation when varying the $\theta$ axis with c) degree of angular dependence for each reference depth represented in terms of parabolic k-values
2.5 Conclusions

The MOSkin dosimeter proved itself as an ideal candidate for patient dosimetry. The dosimeter exhibited dose rate independence, radio-transparency, minimal energy dependence and minimal angular dependence. The lifetime of the MOSkin was observed to span more than 30 Gy, potentially enabling each MOSkin to be used in at least 9 procedures if deterministic thresholds of 2 Gy are not exceeded during each procedure. Through simulation, characterisation to standard beam qualities and comparison to EBT2 film in a diagnostic setting, a comprehensive understanding of the dosimeter’s behaviour when irradiated using diagnostic beam qualities was established. While the MOSkin exhibited features favourable for measuring operator dose, the highest diagnostic range MOSkin sensitivity measured was 1.37 mV/mGy, meaning that the MOSkin would not be suitable for measuring operator dose. Regardless, the results will be useful in developing a dosimeter for operator specific usage.
Chapter 3

Development of Dose Minimisation Strategies via the Optimisation of Operator Modifiable Parameters Selected During Procedure

As previously established in Chapter 1, the literature reports that patient radiation exposures experienced during interventional procedures can vary considerably depending on complexity of the procedure, sophistication of the c-arm system technology deployed within the specific catheterisation laboratory and the type and level of clinical training undertaken by the staff performing the procedure\[9\]. While the developing complexity of angiographic procedures is beyond the scope of this physics-based research project, Chapter 3 and Chapter 4 will focus on evaluating the efficacy of dose reduction initiatives made through deployment of dose minimisation strategies and through technological advances and innovations respectively. In Chapter 2, the c-arm x-ray beam characteristics and entrance doses observed by the MOSkin dosimeter changed significantly when field size and SID were adjusted. As such, dose
delivery is affected by the clinical setup chosen by c-arm system operators. Chapter 3 explored a broad range of operator modifiable parameters, assessed the impact of adjusting these parameters on dose delivery in a meaningful way, produced a series of best practice recommendations for c-arm system operators based upon the findings of the study and compared to the recommendations presented by other prominent studies within the literature.

3.1 Clinical Equipment, Patient Representative Phantom Volumes, defining ‘Dose’ Delivery and Identifying Operator Modifiable Parameters

The measurements were performed using a Philips Allura Xper FD20 c-arm system and a GE Innova 2100IQ c-arm system. The c-arm systems were operated using clinician suggested imaging protocols. There were two phantoms used during this study, a custom made 30 × 30 × 30 cm$^3$ PMMA phantom which was previously described in Section 2.2.2 and a CIRS 702-C Atom phantom provided by Liverpool Hospital. Images of these phantoms are provided in Figure 3.1. The 30 × 30 × 30 cm$^3$ PMMA slab phantom was composed of several Perspex blocks. By reconfiguring and subtracting blocks from the phantom volume the phantom thickness could be adjusted in 1 cm increments. The CIRS 702 Atom phantom is an anthropomorphic head and torso phantom[125]. This phantom is composed of tissue-equivalent epoxy resins which have been shaped into 21 defined organ volumes encased within a dense skeletal frame. Design of the phantom volume was based upon recommendations issued within ICRP Publication 23 and ICRU Report 48[126,127].
3.1. Clinical Equipment, Patient Representative Phantom Volumes, defining ‘Dose’ Delivery and Identifying Operator Modifiable Parameters

3.1.1 Measuring ‘Dose’ in the Coronary Catheterisation Laboratory

Recommendations on optimal clinical setup can be found in the literature, however, the results of these studies are not always clear and can appear to contradict each other due to the use of broad or general terminology. Terms such as peak skin dose, maximum skin dose, air Kerma at reference point, entrance skin dose, dose area product, Kerma area product and the more general term ‘dose’ are easily conflated which can lead to confusion between results. To develop effective dose minimisation solutions, radiation exposure must be assessed in a clear and meaningful way.

During this study, dose delivery was assessed using two quantities: entrance skin dose rate (ESDR) and dose area product rate (DAPR). The ESDR was defined as the dose delivered to the measurement point at the surface of the phantom volume as measured by a dosimeter and normalised to time while the DAPR was defined as the product of the measured ESDR value and the cross-sectional area of the c-arm beam at the measurement point. The cross-sectional area was calculated through trigonometric interpolation and through the properties of similar triangles. The parameters needed
3.1. Clinical Equipment, Patient Representative Phantom Volumes, defining ‘Dose’ Delivery and Identifying Operator Modifiable Parameters

to calculate this cross-section are represented graphically in Figure 3.2.

3.1.2 Indexation of Operator Modifiable Parameters

This study observed the impact on dose delivery introduced by adjusting the following factors:

- Field Of View (FOV)
- Beam Collimation
- Wedge Filtration
- Source to Image receptor Distance (SID)
- Patient to image-receptor (Air gap)
- Table height
- Phantom thickness
- C-arm gantry angulation
Figure 3.2: Diagrammatic representation of variables relevant to calculating cross-sectional area of the c-arm beam
3.2 Specific Methodologies for Monitoring Each Operator Modifiable Parameter

Specific methodologies have been devised for monitoring dose delivery with respect to each of the parameters listed in Section 3.2. The results were presented in two graphs, each representing change in ESDR and DAPR respectively. Where the selected operator modifiable parameter was assessed over multiple settings or configurations, error was assessed using a 95% confidence interval about the average change in dose delivery. For each operator modifiable parameter assessed, ESDR and DAPR were labelled as correlating, inversely correlating or as uncorrelated. These behaviours were represented on each graph with use of the colours green, red and grey respectively.

3.2.1 Field of View

The impact of adjusting the FOV settings was observed with the Philips Allura Xper FD20 system using the 30 × 30 × 30 cm³ PMMA phantom. During these measurements, the phantom’s centre was aligned to the rotational isocentre of the c-arm gantry. The MOSkin dosimeter was placed on the underside of the phantom at a point central to the incident c-arm beam. All measurements were performed using the posterior-anterior c-arm angulation setting. Measurements were performed using the 15, 20 and 25 cm DFS settings.

3.2.2 Beam Collimation

During procedures, the c-arm beam can be collimated using two pairs of lead shutters. These shutters completely attenuate beam contribution in the collimated areas. The impact of adjusting the beam collimation was observed with the Philips Allura Xper FD20 c-arm system using the 30 × 30 × 30 cm³ PMMA phantom. During these
3.2. Specific Methodologies for Monitoring Each Operator Modifiable Parameter 84

measurements, the phantom’s centre was aligned to the rotational isocentre of the c-arm gantry. The MOSkin dosimeter was placed on the underside of the phantom at a point central to the incident c-arm beam. All measurements were performed using the posterior-anterior c-arm angulation setting.

\[
Collimation = \frac{\text{Collimated Beam Area}}{\text{Primary Beam Area}} \times 100\% \tag{3.1}
\]

\[
\text{Pixels}_{\text{Image}}(X,Y) = \sum_{m=0}^{X-1} \sum_{n=0}^{Y-1} \frac{\text{pixel value}_{mn}}{\text{maximum pixel value}} \tag{3.2}
\]

\[
A_{\text{Cal}} = A_{\text{CS}} \times \frac{\text{pixels}_{\text{Cal}}}{\text{pixels}_{\text{Total}}} \tag{3.3}
\]

\[
DAP_{\text{Cal}} = D_{MP} \times A_{\text{Cal}} = \frac{A_{\text{Cal}}}{A_{\text{CS}}} \times DAP_{PBE}, \text{ Gy} \cdot \text{cm}^2 \tag{3.4}
\]

The collimated beam area, that is, the beam area transmitted through the beam collimator was described as a fraction of the total uncollimated primary beam area as calculated in Equation 3.1. Raw images of each of the collimation settings acquired are presented in Figure 3.3.

![Figure 3.3](image)

**Figure 3.3:** a) No Collimation b) 73% of Primary Beam Area c) 52% of Primary Beam Area d) 22% of Primary Beam Area e) 9% of Primary Beam Area
Matlab ([https://www.mathworks.com/products/matlab.html](https://www.mathworks.com/products/matlab.html)) was used to threshold each of the images into binary values of either 0 or 255. The thresholded images were then normalised to maximum pixel value, 255, to provide a pixel count of the collimated beam area, as shown in **Equation 3.2**. The ratio of the collimated beam area pixels to the total pixels in the image was then multiplied by the area of the primary beam cross-section at measurement point which was then used to calculate the DAP at measurement point, as formulaically represented in **Equation 3.3** and **Equation 3.4**. Measurements were performed with beam area collimated to 100%, 73%, 52%, 22% and 9% of the primary beam area.

### 3.2.3 Wedge Filtration

During procedures, the c-arm beam can be partially attenuated using two tapered lead-rubber/lead-acrylic sheets commonly known as wedges. These wedge filters remove low energy beam contribution and balance image glare when imaging high contrast regions (e.g. lung material, air). The impact of adjusting the degree of wedge filtration was observed with the Philips Allura Xper FD20 c-arm system using the $30 \times 30 \times 30$ cm³ PMMA phantom. During these measurements, the phantom’s centre was aligned to the rotational isocenter of the c-arm gantry. The MOSkin dosimeters were placed on the underside of the phantom at points that corresponded to the placement of the wedge filters. All measurements were performed using the posterior-anterior c-arm angulation setting. The entrance dose delivered to three types of filtered areas was assessed: no wedge filters present, 1 wedge filter present and 2 wedge filters overlaid. DAP was assessed for five different wedge filter configurations, as shown in **Figure 3.4**. The position of each MOSkin dosimeter has been presented in **Figure 3.4F** through the superposition of MOSkin icons on an unfiltered image. The DAP for each configuration was calculated by summating the DAP delivered to three separate
regions of interest that corresponded to each filtered area type. The area for each of these regions of interest was calculated by thresholding each image using the Matlab software package using a similar methodology to that found in Section 3.2.2. Using this methodology, the area of the three different filtered regions was calculated. The DAP values for each region were weighted using the corresponding ratio of filtered beam area to total beam area, as shown in Equation 3.9.

\[
DAP_{\text{Wedge}} = \sum_{i=0}^{3} \frac{\text{pixels}_{col_i}}{\text{pixels}_{\text{Total}}} \times D_{\text{col}_i} \times A_{CS}, \text{ Gy} \cdot \text{cm}^2
\]  \hspace{1cm} (3.5)

**Figure 3.4:** a) acquisition with no additional filters b) acquisition with one filtration wedge applied c) acquisition with two filtration wedges applied separately d) acquisition with two filtration wedges overlaid e) acquisition with two filtration wedges partially overlaid f) acquisition with MOSkin dosimeters superimposed to represent the three dosimetric measurement points utilised
3.2.4 Source to Image-receptor Distance, Table Height and Air Gap

The impact of adjusting the SID settings was observed with the Philips Allura Xper FD20 c-arm system using the $30 \times 30 \times 30 \, \text{cm}^3$ PMMA phantom. The MOSkin dosimeter was placed on the underside of the phantom at a point central to the incident c-arm beam. All measurements were performed using the posterior-anterior c-arm angulation setting. Measurements were performed by adjusting the table height as measured from the focal point of the c-arm system’s x-ray tube and the SID, or, image-receptor height. Measurements were performed using SID settings that ranged from 91-120 cm and using table height settings that ranged from 56.5-80.5 cm as measured relative to the focal point of the x-ray tube. Some SID/table height configurations were physically impossible to produce and as such these configurations were not recorded. The SID and table height configurations allowed for several air gap thicknesses, that is, distances between the phantom volume and image-receptor plate. Air gap thicknesses ranging from 1-30 cm as incremented in 1 cm intervals were observed. Air gap values were calculated using Equation 3.10. The table height, SID and air gap parameters are diagrammatically represented in Figure 3.5. The cross-sectional FOV for each of these configurations was calculated using the methodology established in Section 3.2.2. These area values were used to calculate DAPR at measurement point.

\[
\text{Air Gap} = \text{SID} - (h_{MP} + h_{Phantom}), \, \text{cm}
\]  

(3.6)
3.2. Specific Methodologies for Monitoring Each Operator Modifiable Parameter

Figure 3.5: Diagram representing the operator modifiable parameters of SID, Table Height and Air Gap relative to the c-arm gantry.
3.2.5 Phantom Thickness

The impact of adjusting phantom thickness was assessed with the Philips Allura Xper FD20 and GE Innova 2100IQ c-arm systems with the $30 \times 30 \times 30$ cm$^3$ PMMA phantom. All measurements were performed using clinician preferred protocols. During these measurements, the phantom’s centre was aligned to the rotational isocentre of the c-arm gantry. The MOSkin dosimeter was placed on the underside of the phantom at a point central to the incident c-arm beam. All measurements were performed using the posterior-anterior c-arm angulation setting. Measurements were performed using the 15, 20 and 25 cm DFS settings when using the Philips Allura Xper FD20 c-arm system and the 10, 15 and 17 cm OFS settings when using the GE Innova 2100IQ c-arm system. As mentioned in Section 2.2.2, these field settings allow for effectively similar FOV projections to be used between machines. The initial measurements were performed using the full volume 30 cm phantom height configuration with each subsequent measurement performed after subtracting one 1 cm of height in PMMA blocks until reaching the 19 cm phantom height configuration. This reduction of phantom volume is diagrammatically represented in Figure 3.6. The cross-sectional FOV for this configuration was calculated using the methodology established in Section 3.2.2. Only one value was necessary in this experiment as the c-arm system gantry configuration did not change with phantom thickness. This area value was used to calculate DAPR at measurement point.
Figure 3.6: Diagrammatic representation of phantom thickness diminishing by 1 cm with each measurement for EPT ranging from 19-30 cm (PMMA phantom aligned with c-arm rotational isocentre)
3.2.6 C-arm Angulation using the CIRS 702 Atom Phantom

The impact of adjusting the c-arm gantry angulation was assessed with the Philips Allura Xper FD20 using the CIRS 702 Atom phantom. All measurements were performed using clinician preferred protocols. During these measurements, the phantom’s heart was aligned to the rotational isocentre of the c-arm gantry, as diagrammatically represented in Figure 3.7. The MOSkin dosimeter was placed on the underside of the phantom at a point central to the incident c-arm beam and was repositioned for each specific c-arm projection used. The c-arm gantry was varied around the left anterior oblique/right anterior oblique (LAO/RAO) and the cranial/caudal axis (CRA/CAU). A total of fifteen measurement points were observed. These measurement points have been identified on cross-sectional DICOM imagery of the phantom volume in Figure 3.8a (for LAO/RAO axis measurement points) and Figure 3.8b (for cranial/caudal (CRA/CAU) axis). The c-arm angulation was defined in these measurements using the angular offset of the x-ray tube from the anterior-posterior x-ray tube position with respect to the gantry axis of interest. All c-arm angulations were specified by clinicians based at Sutherland Heart Clinic to ensure clinical relevance. When measuring dose area product delivered to the CIRS ATOM phantom, the DAPR was calculated consistent with traditional DAP methodologies and as such was defined as the product of the ESDR delivered to the measurement point with the cross-sectional beam area at measurement point, as calculated using the methodology established in Section 3.2.2.
3.2. Specific Methodologies for Monitoring Each Operator Modifiable Parameter

Figure 3.7: Diagrammatic representation of the CIRS 702 phantom with heart aligned to the c-arm gantry rotational isocentre and with ESD measure by the MOSkin

Figure 3.8: CT DICOM of the CIRS 702 Atom Phantom representing the phantom heart aligned to c-arm rotational isocenter presented as a) axial cross-section in plane of measurement points with LAO/RAO measurement points identified and as b) sagittal cross-section in plane of measurement points with CRA/CAU measurement points identified
3.3 The Impact of Operator Modifiable Parameters on Dose Delivery

The results of this study have been presented in Section 3.3.1 to Section 3.3.6 with each subsection representing a specific operator modifiable parameter. Each subsection specifies a corresponding figure and notes both the maximum and average change in the dose quantities with respect to the change in operator modifiable parameter settings. Each figure depicts dose delivery in terms of ESDR and DAPR separately. A trend line, typically linear, was generated for each graph. These trend lines may not be representative of the behaviour depicted by the data but were useful in establishing a general increasing or decreasing trend with respect to the parameter being assessed. An error of ±10% of the trend line values was applied to the trend line to account for the many sources of variance. Each graph was also assigned colours to represent correlation (green), inverse-correlation (red) or no correlation (grey) between the ESDR and DAPR for each specific operator modifiable parameter.
3.3. The Impact of Operator Modifiable Parameters on Dose Delivery

3.3.1 The Impact of Field of View on Dose Delivery

The ESDR decreased with increasing FOV, as shown in Figure 3.9a. Over the full range of 15-25 cm DFS settings and for all observed phantom thicknesses, ESDR decreased by a maximum of -44% and decreased by an average of -33%. The DAPR increased with increasing field size, as shown in Figure 3.9b. Over the full range of 15-25 cm DFS settings and for all observed phantom thicknesses, DAPR increased by a maximum of +116% and increased by an average of +86%. When adjusting FOV, the ESDR and DAPR were observed to inversely correlate.

3.3.2 The Impact of Beam Collimation on Dose Delivery

The c-arm system did not adjust beam characteristics upon introduction of beam collimation. Collimating the beam to an area between 22% and 100% of the primary beam area did not significantly affect ESDR, as shown in Figure 3.10a. Collimation to 9% of the primary beam area resulted in an average increase in ESDR of +15%. The DAPR decreased with increasing beam collimation, as shown in Figure 3.10b. Collimating the beam to 9% of the primary beam area resulted in a reduction in DAPR of -86.7% for the 25 cm field size setting, -85.8% for the 20 cm field size setting and -85.5% for the 15 cm field size setting. ESDR and DAPR were not correlated.

3.3.3 The Impact of Wedge Filtration on Dose Delivery

The c-arm system did not adjust beam characteristics on introduction of wedge filtration. Adding one layer of filtration caused the ESDR to reduce by -62%, while adding two layers of filtration caused ESDR to reduce by -78%, as shown in Figure 3.11a. The total DAPR decreased as the wedge filtration area increased, as shown in Figure 3.11b. ESDR and DAPR were not correlated.
3.3. The Impact of Operator Modifiable Parameters on Dose Delivery

Figure 3.9: Impact of adjusting the FOV on the a) ESDR and b) DAPR delivered to the $30 \times 30 \times 30$ cm$^3$ PMMA phantom volume by the Philips Allura Xper FD20 c-arm system with linear inversely-correlating linear trend lines applied.

Figure 3.10: Impact of adjusting beam collimation on the a) ESDR and b) DAPR delivered to the $30 \times 30 \times 30$ cm$^3$ PMMA phantom volume by the Philips Allura Xper FD20 c-arm system.

Figure 3.11: Impact of adjusting wedge filtration settings on the a) ESDR and b) DAPR delivered to the $30 \times 30 \times 30$ cm$^3$ PMMA phantom volume by the Philips Allura Xper FD20 c-arm system. (For visual reference to wedge filtration configurations, refer to Figure 3.4)
3.3.4 The Impact of Source to Image-receptor Distance, Table Height and Air Gap on Dose Delivery

The ESDR increased with increasing SID, as shown in Figure 3.12a. Over the full range of 91-120 cm SID for all observed table height settings, the ESDR increased by a maximum of +49% and increased by an average of +1.9% per cm. The DAPR decreased with increasing SID, as shown in Figure 3.12b. Over the full range of 91-120 cm SID for all observed table height settings, DAPR decreased by a maximum of -14% but increased by an average of +0.03% per cm. When adjusting SID, ESDR and DAPR were observed to inversely correlate. The ESDR decreased with increasing table height, as shown in Figure 3.13a. Over the full range of table height from 56.5-80.5 cm for all observed SID settings, ESDR decreased by a maximum of -61% and decreased by an average of -3.6% per cm. The DAPR decreased with increasing table height, as shown in Figure 3.13b. Over the full range of table heights from 56.5-80.5 cm for all observed SID settings, DAPR decreased by a maximum of -21% and decreased by an average of -0.95% per cm. When adjusting table height, ESDR and DAPR were observed to correlate. The ESDR increased with increasing air gap, as shown in Figure 3.14a. Over the full range of 34.5-63.5 cm Air gap, ESDR increased by a maximum of +49% and increased by an average of +7.29% per cm with increasing Air gap. The DAPR on average decreased with increasing detector to imager distance, as shown in Figure 3.14b. Over the full range of 34.5-63.5 cm air gap values, DAPR decreased by a maximum of -14% and decreased by an average of -1.51% per cm. When adjusting air gap, ESDR and DAPR were observed to inversely correlate. The ESDR decreased with increasing table height when a constant air gap was maintained, as shown in Figure 3.15a. Over the full range of table heights from 56.5-80.5 cm for all observed air gap values, ESDR decreased by a maximum of -45% and decreased by an average of -31%. Over the full range of table heights from 56.5-80.5 cm for all
observed air gap values, DAPR decreased by a maximum of -30% and decreased by an average of -19%, as shown in Figure 3.15b. When adjusting table height with a constant air gap maintained, ESDR and DAPR were observed to correlate.

### 3.3.5 The Impact of Phantom Thickness on Dose Delivery

The ESDR and DAPR increased with increasing phantom thickness, as shown in Figure 3.16. When using the Philips Allura Xper FD20 c-arm system over the full range of 19-30 cm phantom thickness, the dose delivered to the phantom surface increased by a factor of 5.15 for the 15 cm DFS setting, 4.74 for the 20 cm DFS setting and 5.34 for the 25 cm DFS setting. When using the GE Innova 2100IQ c-arm system over the full range of 19-30 cm phantom thickness, the dose delivered to the phantom surface increased by a factor of 12.65 for the 12 cm OFS setting, 10.31 for the 15 cm OFS setting and 12.00 for the 17 cm OFS setting.
3.3. The Impact of Operator Modifiable Parameters on Dose Delivery

Figure 3.12: Impact of adjusting the SID on the a) ESDR and b) DAPR delivered to the 30 \times 30 \times 30 \text{cm}^3 PMMA phantom volume by the Philips Allura Xper FD20 c-arm system with linear inversely correlating linear trend lines applied.

Figure 3.13: Impact of adjusting table height on the a) ESDR and b) DAPR delivered to the 30 \times 30 \times 30 \text{cm}^3 PMMA phantom volume by the Philips Allura Xper FD20 c-arm system with correlating trend lines applied.
3.3. The Impact of Operator Modifiable Parameters on Dose Delivery

Figure 3.14: Impact of adjusting air gap thickness on the a) ESDR and b) DAPR delivered to the $30 \times 30 \times 30 \text{ cm}^3$ PMMA phantom volume by the Philips Allura Xper FD20 c-arm system with inversely correlating trend lines applied.

Figure 3.15: Impact of adjusting table height with constant air gap thickness on the a) ESDR and b) DAPR delivered to the $30 \times 30 \times 30 \text{ cm}^3$ PMMA phantom volume by the Philips Allura Xper FD20 c-arm system with correlating trend lines applied.
3.3. The Impact of Operator Modifiable Parameters on Dose Delivery

Figure 3.16: Impact of adjusting EPT on the dose delivered to the $30 \times 30 \times 30$ cm$^3$ PMMA phantom volume by the a) Philips Allura Xper FD20 and b) the GE Innova 2100IQ c-arm systems with trend lines applied

3.3.6 The Impact of c-arm Angulation using the CIRS 702 Atom Phantom

The LAO/RAO axis angulation measurements were observed over the range of RAO $90^\circ$ through to LAO $90^\circ$ while CRA/CAUD angulation was maintained at $0^\circ$, as shown in Figure 3.17. The results were normalised to the dose delivered at LAO/RAO $0^\circ$. Dose was observed to be minimised at the angles RAO $45^\circ$ and LAO $40^\circ$ with the lowest dose observed at RAO $45^\circ$. The dose recorded at RAO $45^\circ$ was $-54\%$ lower than the dose observed at LAO/RAO $0^\circ$. The highest doses were observed at three points: At RAO $90^\circ$, LAO $90^\circ$ and at LAO/RAO $0^\circ$. The highest dose was observed at $90^\circ$ LAO. The dose recorded at LAO $90^\circ$ was $+48\%$ higher than the dose observed at LAO/RAO $0^\circ$. The CRA/CAUD axis angulation measurements were observed over the range of CRA $45^\circ$ to CAUD $45^\circ$ while LAO/RAO angulation was maintained at $0^\circ$ incidence. The lowest dose was observed at $0^\circ$ CRA/CAUD, as shown in Figure 3.18. The highest dose was observed at CRA $45^\circ$ where dose increased to $+456\%$ higher than the dose recorded at CRA/CAUD $0^\circ$. Dose delivery to CAUD $45^\circ$ was observed to be $+350\%$ higher than the dose recorded at CRA/CAUD $0^\circ$
3.3. The Impact of Operator Modifiable Parameters on Dose Delivery

Figure 3.17: Impact of adjusting LAO/RAO angulation on the ESDR and DAPR delivered to the CIRS 702 Atom phantom volume by the Philips Allura Xper FD20 c-arm system.

Figure 3.18: Impact of adjusting CRA/CAU angulation on the ESDR and DAPR delivered to the CIRS 702 Atom phantom volume by the Philips Allura Xper FD20 c-arm system.
3.4 Relative Importance of Operator Modifiable Parameters

3.4.1 Summary of Experimental Data and Subsequent Clinical Recommendations

A comprehensive summary of the parameters adjusted, the changes to ESD and DAP observed on adjusting the parameters and the correlation between the changes to ESD and DAP can be found within Table 3.1. This study observed that ESDR and DAP are very different measures that often exhibit independent behaviours in the catheterisation laboratory.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>From</th>
<th>To</th>
<th>ESDR</th>
<th>DAPR</th>
<th>ESDR/DAPR</th>
<th>Relationship</th>
</tr>
</thead>
<tbody>
<tr>
<td>Field Diagonal</td>
<td>15 cm</td>
<td>25 cm</td>
<td>↓30.5%</td>
<td>↑93.0%</td>
<td></td>
<td>Inverse Correlation</td>
</tr>
<tr>
<td>Collimation</td>
<td>0%</td>
<td>91%</td>
<td>↑2%</td>
<td>↓98.3%</td>
<td></td>
<td>Uncorrelated</td>
</tr>
<tr>
<td>Wedge Filtration</td>
<td>No Filtration</td>
<td>Maximum Filtration</td>
<td>↓79%</td>
<td>↓28%</td>
<td></td>
<td>Uncorrelated</td>
</tr>
<tr>
<td>SID</td>
<td>91 cm</td>
<td>121 cm</td>
<td>↑49%</td>
<td>↓6%</td>
<td></td>
<td>Inverse Correlation</td>
</tr>
<tr>
<td>Air Gap</td>
<td>1 cm</td>
<td>30 cm</td>
<td>↑49%</td>
<td>↓7%</td>
<td></td>
<td>Inverse Correlation</td>
</tr>
<tr>
<td>Table Height</td>
<td>55 cm</td>
<td>80 cm</td>
<td>↓59%</td>
<td>↓16%</td>
<td></td>
<td>Positive Correlation</td>
</tr>
<tr>
<td>Phantom EPT</td>
<td>19 cm</td>
<td>30 cm</td>
<td>↑408%</td>
<td>↑408%</td>
<td></td>
<td>Positive Correlation</td>
</tr>
<tr>
<td>LAO/RAO Angulation</td>
<td>Minimum Dose (40°LAO)</td>
<td>Maximum Dose (90°RAO)</td>
<td>↑249%</td>
<td>↑208%</td>
<td></td>
<td>Positive Correlation</td>
</tr>
<tr>
<td>CRA/CAU Angulation</td>
<td>Minimum Dose (0°CRA/CAU)</td>
<td>Maximum Dose (45°CRA)</td>
<td>↑456%</td>
<td>↑370%</td>
<td></td>
<td>Positive Correlation</td>
</tr>
</tbody>
</table>

Table 3.1: Averaged Impact on Dose Delivery from Manipulation of Operator Modifiable Parameters
3.4. Relative Importance of Operator Modifiable Parameters

Based on the outcome of this study, the following dose minimisation strategies should be recommended in the angiographic catheterisation laboratory:

1. EPT was the largest determinant of dose delivery. ESD and DAP were observed to increase by up to a factor of 5.5 for the Philips Allura Xper FD20 c-arm system and by up to a factor of 12.65 for the GE Innova 2100IQ c-arm system when 11 cm was added to EPT. Using steep angles will substantially increase dose to the patient and, as such, shallow angles that avoid spinal structures should be used where possible. For large patients, additional shielding should be utilised and high exposure protocols should be used with caution.

2. DAP reduction should be implemented in general practice to reduce operator dose and stochastic risk to the patient. To reduce DAP, field size should be minimised, collimation and wedge filtration should be applied where possible, table height should be maximised and SID should be maximised. Reducing DAP may also reduce ESD delivery in an optimised configuration (ESD: ↓47.2%, DAP: ↓58.0%).

3. ESD reduction is important for patients at risk of developing radiation-induced skin injuries. This can occur during long procedures or when there are limited useful acquisition angles. In these cases, field size should be maximised, table height should be maximised, SID should be minimised and air gap should be minimised. Minimising ESD may significantly increase the patient volume irradiated resulting in an increase in DAP delivery (ESD: ↓74.5%, DAP: ↑56.6

4. Collimation and wedge filtration should be applied where possible to further reduce dose delivery. ESD can be reduced by increasing collimation and layering wedge filtration while DAP was most effectively reduced by increasing the area covered by collimation and wedge filtration.
5. This study outlines recommendations to guide dose minimisation strategies within the catheterisation laboratory, however, it is important that the operator performs procedures in a setting that is comfortable so that they may perform the procedure efficiently and effectively.

6. Further development of dose minimisation strategies should be developed for catheterisation laboratories using site-specific reference data provided by comprehensive dosimetry solutions.

These recommendations were compared to four established reviews into dose reduction in angiographic catheterisation laboratories. In general, the recommendations made by these authors, Mettler et.al., Stecker et.al., Chambers et.al. and Hertault et.al., compare favourably to the recommendations made above from the experimentally derived data\textsuperscript{[128 - 131]}.

### 3.4.2 Impact of Field Related Operator Modifiable Parameters

All reviewed authors recommended using larger field size settings where possible. According to Mettler et.al., reducing the field size, also referred to as increasing magnification/zoom settings, reduces the image ‘brightness’, the photon flux, incident upon the image-receptor\textsuperscript{[128]}. The c-arm system will adjust other parameters to compensate for this effect which results in a higher entrance skin dose imparted to the patient. Hertault et.al. suggest that dose delivery changes as a function of the magnification factor, that is, the ratio of the maximum field size to the selected field size\textsuperscript{[129]}. Hertault et.al. also suggested that for uncorrected image intensifier systems dose will change proportionally to the square of the magnification factor while corrected image intensifier systems and newer flat panel systems will attempt to adjust dose deliv-
ery as a linear function of magnification factor. Experimentally the linear correlation between ESDR and magnification factor assessed using regression analysis. Regression analysis established linearity to an r-squared value of greater than 0.90 for all EPT configurations and to an r-squared value of greater than 0.98 for 75% of EPT configurations. All reviewed authors recommended using collimation where possible to reduce scatter radiation, improve accuracy of iodine contrast imaging and limit exposure to surrounding tissue. Hertault et.al. cite a study by Haqqani et.al. that observed that dose delivery to operators decreases proportionally to the reduction of image size, which correlates well with the trend in DAPR values observed experimentally[129, 132]. Mettler et.al. specify that while increasing collimation does decrease total patient exposure by reducing the irradiation volume, collimation does not reduce entrance skin dose rate[128]. Mettler et.al. also note that highly collimated beams may produce higher entrance skin dose rates to compensate for the lower photon flux incident on the image-receptor and the effects this has on images. This effect was observed experimentally when ESDR was observed to increase by an average of 15% when the collimated beam area reached 9% of the primary beam area.

No reviewed author mentioned the impact of wedge filtration on dose delivery which is probably as this mechanism is primarily used to improve image quality rather than as a dose reduction tool. Despite this, implementation of wedge filtration was observed to effectively reduce dose. Introduction of a single filtration wedge caused the ESDR of the filtered area to drop by 62% while overlaying two filtration wedges caused the ESDR of the filtered area to drop by 78%. The DAPR was minimised in configurations which prioritised area coverage rather than configurations where wedge filters were overlaid. The experimental results suggest that collimation should remain the preferred method to reduce and shape the beam area but wedge filtration could be used in conjunction with collimation to reduce dose delivery to the periphery of the
field of view and retain partial visibility of the filtered region.

### 3.4.3 Impact of C-arm Gantry related Operator Modifiable Parameters

Stecker et.al. were the only authors to provide recommendations regarding SID and recommended that SID should be maximised during procedures\[^{130}\]. Mettler et.al. caution that a practical maximum source to image-receptor distance should be established as maximising SID may adversely affect image quality which may prolong imaging time and may cause the c-arm system to select higher dosage characteristics.\[^{128}\]. Mettler et.al. and Chambers et.al. suggest that table height is a critical factor to dose delivery\[^{128, 131}\]. Mettler et.al. note that optimal practice should involve positioning the x-ray tube as far from the patient volume as possible. Metter et.al. caution that sub-optimal positioning of the x-ray tube can result in ESDR delivery increasing by a factor of ten and can cause fluoroscopic injury to occur within a very short period of time. Hertault et.al. suggest that the table should be positioned so that operators’ heads and chests are not too close to the patient volume as these are the primary sources of operator scatter exposure\[^{129}\]. All reviewed authors recommended that the air gap between the patient and image-receptor should be minimised\[^{128 - 131}\]. From an image quality perspective, reducing the air gap thickness would minimise the scattering and divergence of photons transmitted through the patient volume. This would reduce the number of photons discriminated against by the anti-scatter grid and would maximise photon flux incident on the image-receptor plate. In terms of dose delivery, larger air gaps would result in reduced photon flux incident upon the image-receptor plate which would drive the c-arm system to select higher dose delivering x-ray tube characteristics. Larger air gap thicknesses may also increase operator exposure due to a higher proportion of the x-ray beam would contribute to the radiation scatter
field. The experimental results demonstrated that both the ESDR and DAPR were most effectively reduced by maximising the distance of the measurement point to the x-ray tube. As per inverse square law, the degree of dose reduction decreased with increasing table height. This loss in effectiveness was more prominent for the measured DAP as a competing effect was noted in that the cross-sectional area of the c-arm beam increased with increasing table height. As such, while maximising the distance between the patient and x-ray tube minimised dose, it was especially important to avoid low table positions. While these results concern selection of table height, this recommendation should also be taken into consideration when selecting angulation as some projections may result in the x-ray tube being placed closer to the patient volume. The ESDR could also be reduced by minimising the SID and the air gap thickness, however, on average this also increased the DAP delivery. This behaviour was not established in all observed configurations. Similarly to when table height was adjusted, the effect of adjusting image-receptor height on DAP delivery was representative of the complicated and competing interplay between the trend in ESD delivery and the geometric deformation of the cross-sectional beam area at measurement point. While the trend in ESD delivery was dominant when table height was adjusted, the effect on the cross-sectional area at measurement either balanced or outweighed the effect on ESD delivery when image-receptor height is adjusted and as such DAP delivery behaved counter to ESD delivery for these settings. As with the selection of field size, minimisation of one dose quantity will result in an increasing to the other. Despite the effects on DAP, consideration of the significantly greater impact increasing image-receptor height and air gap thickness has on ESD delivery, the effect on radiation scatter and the potential effects on image quality suggests minimisation of these parameters is the optimal practice to recommend.
3.4.4 Impact of Patient Thickness and C-arm angulation

All authors recommended caution when irradiating larger patient volumes and cross-sections \(^{[128 - 131]}\). Mettler et.al. explain that thicker patient volumes require greater radiation dose delivery to achieve adequate penetration of their bodies\(^{[128]}\). Hertault et.al. suggest that c-arm systems adjust dose production settings to provide a consistent signal to noise ratio at the lowest possible exposure rate which causes higher dose delivery at larger patient thicknesses\(^{[129]}\). Stecker specify that consideration of radiation risk should be discussed in further depth for patients weighing more than 135 kg\(^{[130]}\). Mettler et.al. specify that ESD has been observed to reduce by a factor of two for each 4.5 - 5.0 cm of tissue depth\(^{[128]}\). This figure compared well to the experimental results where ESDR was observed to halve when reducing the 30 cm phantom thickness by 5.36 cm when using the Philips Allura Xper FD20 c-arm system and by 5.70 cm when using the GE Innova 2100IQ c-arm system. All authors revised recommended that steep angles that maximise patient cross-section should be avoided or treated with caution where possible\(^{[128 - 131]}\). These angles increase the effective patient thickness driving the c-arm system to deliver higher radiation exposures, reducing image quality and adding greater contributions to scatter fields. The revised authors each recommend using tightly collimated beams, intelligent selection of beam projections. The experimental angulation results were compared to a previous study undertaken by Kuon et.al. in 2004 that attempted to comprehensively map both operator and patient doses as a function of angulation using a Rando-Alderson torso phantom and measuring dose using a PTW ionisation chamber\(^{[133]}\). Excerpts of the data published by Kuon et.al. have been compared to the DAPR values observed using the CIRS 702 phantom in combination with the MOSkin dosimeter in Figure 3.20. The comparison observes similarities in the asymmetries in the LAO/RAO axis where lower doses were observed at shallow LAO projections and in the CRA/CAU
axis where dose was observed to increase more steeply when imaging in the higher CRA projections. There were discrepancies between the datasets. Interestingly, the Kuon et.al. data does not observe the same increase in dose seen when imaging the CIRS 702 phantom in the PA projection, as shown in Figure 3.20a. This is most likely caused by differences in bone density between the CIRS 702 Atom phantom used in this study as compared with the Alderson Rando phantom used by Kuon et.al.. These differences could occur due to the differences in age, sex and size of the human model that each phantom is based on. The discrepancy could have more specifically occurred as the CIRS 702 Atom phantom design incorporates a homogenous bone substitute in contrast to Alderson Rando phantoms which are designed with cortical and trabecular bone volumes modelled separately. Homogenous bone substitutes produce consistent, reproducible and simple phantom volumes. The density of the homogenous bone substitute has been developed as an average of cortical and trabecular bone densities as weighted by the volume ratio of these bone structures. These ratios are specific for different age and gender demographic. The density of the homogenous bone substitute is calculated as a weighted average which is based upon published values of the density and volume ratio of cortical and trabecular bone content as observed for people of specific age and gender demographics\textsuperscript{[125]}. Other contributing factors to the discrepancies between data sets may include differences in phantom geometry, dosimetry methods and the c-arm systems used during the Kuon et.al. study. Varying the imaging projection is another method to reduce, or ‘spread’, the ESD during a procedure. This method involves adjusting the angulation of the c-arm system regularly to avoid causing localised points of radiation over exposure. Mettler et.al. advocate use of this technique as it can greatly reduce the risk of local overexposure, however, studies performed by Pasciak et.al. and Koenig et.al. recommend that great care is taken in ensuring the selected projections do not overlap as this would unknowingly
and greatly increase the risk of fluoroscopic injury\cite{38, 134, 134}. The risk of accidentally overlapping fields could be mitigated through the installation of dose contouring software or through deployment of real-time array type dosimetry\cite{135}. By using a real-time dosimetry system such as the MOSkin dosimeter in a 2D array arrangement peak skin dose could be monitored effectively, DAP delivery to the patient could be measured conformal to the patients’ surface and inhomogeneity of the incident imaging beam could be accounted for. Such solutions could provide dose distributions and organ-specific assessments of stochastic risks with greater accuracy than traditional methodologies in either a stand-alone or supplementary format.

Figure 3.19: Comparison of experimental angulation data around the a) LAO/RAO axis and the b) CRA/CAU axis\cite{125}
3.4. The Importance of clear terminology

As mentioned previously in Section 3.2.2, there are sources of confusion within the literature as to how to optimise and reduce ‘dose’ delivery in the angiographic catheterisation laboratory. Mettler et.al. cite one such fallacy and explain that recommendations to utilise collimation to reduce ‘exposure’ are often misunderstood to mean the use of collimation reduces the dose rate delivery to the patients’ exposed skin\textsuperscript{[128]}. While all reviewed authors did explain the difference between ESD and DAP within their publications, their recommendations used the more general term ‘dose’ to interchangeably refer to ESD and DAP. This loose use of language contributes to confusion regarding the issues of radiation safety in the catheterisation laboratory. This vague use of terminology is problematic as within this study it was observed that there is often no clear relationship between ESD and DAP delivery when adjusting operator modifiable parameters. While correlation was observed between ESD and DAP when adjusting table height, EPT and angulation, there were complex inverse relationships between ESD and DAP when adjusting field size, SID and air gap thickness. The results of this study and the state of the literature suggest that clearer exposure terminology is necessary and that clinicians may benefit from increased access to dosimetry tools in order to better develop dose minimisation strategies tailored to their own laboratories.
3.4.6 Limitations of this study

This study featured several key limitations:

1. *This study was performed using phantom volumes:* Phantom volumes are a simplified representation of a patient volume. The effects of adjusting operator modifiable parameters on dose delivery may not translate exactly to an equivalent patient volume.

2. *This study did not directly measure operator dose:* This study focused on assessing radiation exposure to angiography patients. Commentary on dose delivered to operators was limited and relied on an assumption of proportionality between the DAP, the scatter radiation produced during a procedure and the operator dose. This proportionality does not take into consideration operator position with respect to the x-ray tube, direction of the x-ray beam, effects related to positioning of the image receptor or the effect of radiation protective shielding equipment. Further commentary on operator dose was beyond the scope of this study. There are operator-specific studies that exist within the literature that can provide further commentary on operator dose.

3. *This study was performed primarily using one c-arm system:* Catheterisation laboratories vary significantly. Different laboratories may feature different c-arm systems, different room geometries and different c-arm system protocol tailored to the preferences of different clinicians. As such, individual catheterisation laboratories may observe that dose delivery is affected to differing extents than those observed during this specific study. The effect these differences can have on c-arm dose delivery was exemplified in Figure 3.16 where the Philips Allura Xper FD20 and GE Innova 2100IQ c-arm systems were observed to respond differently to a selection of phantom EPT configurations.
4. The calculation of DAP assumed that the FOV was homogeneous:

The calculation of DAP assumed that the x-ray beam was homogeneous across the FOV. In reality the x-ray spectrum is heterogeneous, that is, photon fluence and photon energies will vary depending on measurement point. The most significant contribution to heterogeneity is beam hardening effect. Beam hardening occurs because lower energy photons are attenuated at a greater rate than higher energy photons. As a consequence of x-ray beam divergence, beam hardening will affect photon fluence and photon energies differently depending on their trajectory. This typically results in a higher intensity of x-rays at the center of the FOV and a lower intensity of x-rays at the edges of the FOV. Beam hardening is highly dependent on the x-ray tube potential, field size, the filtration utilised and the attenuative material that the x-rays pass through. As such, the degree of beam hardening will vary between acquisitions/projections and would have introduced inaccuracies, specifically when introducing wedge filtration.

Fortunately, the energy dependence of the MOSkin dosimeter was thoroughly investigated in Chapter 2. In Chapter 2 it was observed that the MOSkin energy dependence was minimal for the x-ray beam spectral range of interest. This can be exemplified with two conservative examples:

- addition of 2 mmAl filtration to a 70 kVp unfiltered x-ray beam reduces the relative MOSkin response by $<10$.

- shifting the x-ray tube voltage by 20 kV (e.g. from 100 kVp to 80 kVp reduces the relative MOSkin response by $<10$.

Either of these examples would have a greater impact on the x-ray beam spectrum than beam hardening from introducing the rubber/silicone wedges to the FOV and as such the MOSkin response can be expected to be accurate within 10% of the readout
value. Furthermore there were no observed changes to the system reported DAPR, air Kerma or x-ray tube characteristics wedge filtration configurations. This feature illustrates another application where MOSkin measurements would provide a more accurate representation of skin dose than the default c-arm system exposure metrics, especially when considering future array-type MOSkin dosimetry solutions.

In some instances, the limitations of this study restrict the applicability of these results to a clinical environment. This study was never intended as a substitute for implementation of comprehensive dosimetry solutions. Instead, this study establishes general trends in dose delivery with respect to the clinical configuration used during procedures. The recommendations produced by the results of this study are a solid foundation for developing dose minimisation strategies informed by physically measured data points.

3.5 Conclusions

This study was performed to improve knowledge and awareness of the differences between the ESD and DAP. The results challenge popular held beliefs about ‘dose’ reduction in the catheterisation lab. This study exemplifies how the gantry, beam and table settings chosen during procedure can significantly affect patient and operator doses and adoption of this study’s recommendations may reduce exposures in general practice. When adjusting operator modifiable parameters, the ESDR and DAPR dose quantities either behaved independently or shared a complex relationship. Minimisation of both ESDR and DAP was not always possible and there is a need for prioritising the type of dose reductions needed on a case by case basis. The strongest contributing factor to dose delivery was the EPT value of the phantom volume. Additional care should be undertaken when choosing projection angle during imaging and when performing procedures with larger patients. The results provide a basis for developing
3.5. Conclusions

dose minimisation strategies. Improving awareness of the dose reduction that clinical setup, projection choice and patient thickness can provide may change operator behaviour during procedures, resulting in reduced risk of radiation-induced conditions to patients and operators. Implementation of comprehensive real-time dosimetry solutions are a necessity for further development of dose minimisation strategies. Future development of 2D real-time array-type dosimeters that conform to the shape of the patient’s body, are not acquired during imaging and do not increase radiation exposure during procedures will provide operators with real-time dosimetric feedback and allow for accurate measurement of the DAP exposure, ESD distribution and maximum skin dose which will assist operators in preventing overexposure to both patients and operators.
Chapter 4

Evaluating C-arm System Performance and the Efficacy of C-arm Systems by Measuring Dose and Image Quality Simultaneously

In Chapter 3, the concept of ‘dose’ delivery as an indicator of c-arm system performance was discussed in depth, the importance of dose minimisation strategies was emphasised and the role of image quality in an optimised clinical setup was alluded to. This chapter aims to define image quality in a clinical context, to discuss the relationship between image quality and dose delivery and to develop a methodology to assess both image quality and dose delivery simultaneously. This chapter will also provide commentary on the efficacy of c-arm system upgrades in improving c-arm system performance.
4.1 Introduction to Evaluating C-arm System Performance

4.1.1 Defining C-arm System Performance through Image Quality

Image quality is a concept used to evaluate and compare groups of images. The metrics used to determine image quality depend on the purpose of the acquired images. In the context of angiographic c-arm imaging, acceptable image quality should meet the following clinical criteria:

1. The acquired images must adequately depict iodine contrast as used to enhance imagery of a patient’s vascular system,

2. The acquired images must be of high spatial resolution to enable operators to examine small vascular structures of interest,

3. The acquired images must enable clear visual discrimination between different tissues, organs, and vascular structures,

4. The impact of motion-induced artefacts, temporal averaging and other imaging distortions should be minimised where possible,

5. The image noise should be minimised where possible while signal to noise ratio and contrast to noise ratio should be maximised where possible, and,

6. An acceptable standard of image quality should be maintained while minimising dose delivery during procedures.
4.1.2 Factors that Influence C-arm System Performance

C-arm system performance can be reduced to two independent metrics, dose delivery and image quality. In general, functionality of an angiographic c-arm system is determined firstly by the sophistication of the image processing chain (hardware) and secondly by both the image processing algorithms and image acquisition protocols installed on the system that determine the x-ray tube output during imaging (software). This means that c-arm system performance is limited by independent hardware and software constraints and c-arm systems can be tuned to meet the image quality standards required by clinicians at the site of commission. This also means that c-arm system performance is periodically improved through advances in c-arm technologies which are provided by vendors through the commission of newer c-arm system models and as upgrades to existing c-arm systems. This study aimed to evaluate the performance of three c-arm systems based at local angiographic catheterisation laboratories that were subject to post-processing and image acquisition protocol upgrades provided by their respective vendors. This evaluation was performed using an existing c-arm system performance standard set by the National Electrical Manufacturers Association (NEMA), the NEMA XR 21 protocol\textsuperscript{[136]}.

4.2 Simultaneous Measurement of Dose Delivery and Image Quality

4.2.1 The NEMA XR 21 Protocol

Image quality was assessed using the NEMA XR 21 protocol in combination with the CIRS Model 901 phantom, a phantom volume that was developed specifically for voluntary compliance with the NEMA XR 21 protocol\textsuperscript{[136 - 138]}. The NEMA XR 21
4.2. Simultaneous Measurement of Dose Delivery and Image Quality

Figure 4.1: Diagram of the CIRS Model 901 phantom configured in the a) 20 cm EPT configuration and in the b) 30 cm EPT configuration

protocol enables simultaneous measurement of c-arm system image quality and dose delivery using a range of patient representative phantom configurations. There are five configurations recommended in the NEMA XR 21 protocol: a 5 cm EPT configuration, a 10 cm EPT configuration, a 20 cm EPT configuration, a 30 cm EPT configuration and a 30 cm phantom configuration with an additional lead plate that ensures that the c-arm system produces its’ maximum output. For relevance to the clinical applications of the evaluated c-arm systems, testing was performed using the 20 cm EPT configuration and the 30 cm EPT configuration. These configurations represent the commonest range of adult patient sizes. These configurations are depicted in Figure 4.1.
4.2. Simultaneous Measurement of Dose Delivery and Image Quality

4.2.2 CIRS Model 901 Phantom Configurations

The c-arm performance was assessed using the two aforementioned CIRS Model 901 phantom configurations. For both configurations, the CIRS Model 901 phantom volume was positioned as per instructed by the NEMA XR 21 protocol.

1. The phantom was placed on the patient couch at a position consistent with the expected position of the patient torso during a procedure. The NEMA XR-21 protocol suggests positioning the phantom centrally to the c-arm system’s rotational isocenter.

2. The c-arm image receptor height was set to produce a 5 cm air gap between the phantom and the image receptor.

3. The phantom was positioned so that the spatial resolution plate was positioned centrally to the acquired image area.

4. The phantom was positioned so that the spatial resolution plate was at a 45° offset to the image receptor positioning. This positioning reduces the occurrence of aliasing effects and moiré patterns on acquired images. These effects can significantly affect assessments of c-arm system spatial resolution\cite{136}.

All imaging was performed with the c-arm system set to the posterior-anterior projection angle. During imaging, the MOSkin dosimeter was taped to the underside of the phantom at a point central to the incident c-arm beam. The MOSkin was used to measure the ESD delivered to the underside of the phantom during imaging. ESD was calculated using previously established calibration methods. Image quality was assessed using a range of image acquisition protocols, imaging modalities and field sizes. The settings used for each of the c-arm systems appraised are discussed in Section 4.3.1 through to Section 4.3.3. The c-arm systems were evaluated using the
five standard image quality tests recommended in the NEMA XR 21 protocols. These tests assessed iodine contrast resolution (ICR), spatial resolution, working thickness range (WTR), static motion target resolution (MTR) and dynamic MTR. During the GE c-arm system evaluation, additional digital assessments were performed to supplement these tests. Each of the image quality tests are discussed in further detail in Section 4.2.3 through to Section 4.2.8.

### 4.2.3 Quantifying Image Quality: Iodine Contrast

The c-arm system ICR was assessed using the image quality target plate, represented by the x-ray image shown in Figure 4.2a. This target plate contained a total of 32 iodinated cavities. There were 8 of these cavities in each quadrant of the target plate. These cavities were grouped into four dot pairs, each pair of successively smaller diameter than the last. The diameter of the first iodine dot pair was 4 mm. Each subsequent dot pair decreased in diameter by 1 mm with the smallest dot pair measuring 1 mm in diameter. Iodine concentration in these cavities varied with quadrant. The first quadrant cavities contained iodine concentrations of 200 mg/cm$^3$. This concentration halved with each subsequent quadrant with the lowest concentration measuring to 25 mg/cm$^3$ in quadrant four. In this test, the assessors recorded the number of iodine dots that were resolved in the acquired images.
4.2.4 Quantifying Image Quality: Spatial Resolution

The c-arm system spatial resolution was assessed using the image quality target plate, represented by the x-ray image shown in Figure 4.2a. The image quality plate was fitted with a Nuclear Associates Model 07-538 lead bar plate. The plate consisted of 0.1 mm thickness lead and featured several densities of line pair patterns cut through the plate. These line pairs varied in density from 1.6 to 5 line pairs per millimeter (lp/mm). In this test, the assessors recorded the highest density of line pairs that could be resolved in the acquired images.

Figure 4.2: CIRS Model 901 phantom equipped with the a) image quality target plate Configuration and equipped with the b) motion target plate
4.2.5 Quantifying Image Quality: Working Thickness Range

The c-arm system WTR was assessed using the image quality target plate, represented by the x-ray image shown in Figure 4.2a. The CIRS model 901 phantom incorporated several embedded objects and cavities within the phantom volume. Each quadrant featured an air column and a metal object column.

1. The air columns were composed of several large air-filled cylinders cut through the PMMA plates and represented low density imaging targets such as lung tissue or air pockets. Each air cylinder was paired with a smaller air-filled pin positioned above the larger air column.

2. The metallic columns were composed of several large aluminium-filled cylinders embedded within the PMMA plate volumes and represented high density imaging targets such as bone or iodine contrast. Each aluminium column was paired with a smaller lead pin positioned above the larger metallic column.

The thickness of the larger cylinders varied with quadrant, all of which have been recorded in Table 4.1. The upper WTR threshold of the c-arm system was determined by noting the number of air-filled pins that were resolved against the larger air-filled cylinders while the lower WTR threshold of the c-arm system was determined by noting the number of lead pins that were resolved against larger aluminium cylinders. Selection of an appropriate WTR prevents image clipping when imaging high/low density imaging targets.
4.2. Simultaneous Measurement of Dose Delivery and Image Quality

<table>
<thead>
<tr>
<th>Quadrant</th>
<th>Air Cylinder Thickness (mm)</th>
<th>Aluminium Cylinder Thickness (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Quadrant 1</td>
<td>175</td>
<td>40</td>
</tr>
<tr>
<td>Quadrant 2</td>
<td>150</td>
<td>47</td>
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<tr>
<td>Quadrant 3</td>
<td>125</td>
<td>53</td>
</tr>
<tr>
<td>Quadrant 4</td>
<td>100</td>
<td>60</td>
</tr>
</tbody>
</table>

Table 4.1: Air cylinder and aluminium cylinder thicknesses embedded within each quadrant of the CIRS model 901 phantom for c-arm WTR assessment

4.2.6 Quantifying Image Quality: Static and Dynamic

Motion Target Resolution

The c-arm system MTR was assessed using the motion target plate, as shown in Figure 4.2b. In this configuration, the motion target plate replaced the previously used image quality target plate. The motion target plate consisted of a motorised plate with five strands of piano wire embedded within the plate volume. These wires were of diameters 0.56 mm, 0.41 mm, 0.30 mm, 0.23 mm and 0.13 mm respectively. The wires extended from the center of the phantom to the diameter of the phantom and were spaced at even intervals of 72° around the face of the motion target plate. During imaging, the plate rotated at a rate of 30 revolutions per minute. In this test, the assessors noted the thinnest wire that was observable in the image. The assessors evaluated images as presented as a series of still frame images (static) and at the real-time framerate (dynamic).
4.2.7 Quantifying Image Quality: Digital Assessment

Digital assessment of image quality was performed during the GE ‘Dose Map’ upgrade evaluation only. Digital assessment was performed on images acquired using the Motion Target plate. This configuration was chosen to maximum homogeneity of the imaging area. For each DICOM image set, a region of interest (ROI) was established using the FIJI software package. The ROI area was chosen by selecting a large homogenous area in the imaging area. This area typically encompassed approximately 10% of the imaging area. The mean pixel value and standard deviation of pixel values for this region were calculated and averaged for all frames included in each image set. The standard deviation of pixel values was used as a measure of image noise, as shown in Equation 4.1 where $\chi_i$ represents each pixel value in the range of pixel 1 to pixel N. An example of a typical ROI and the corresponding sampling distribution have been provided in Figure 4.3.
4.2. Simultaneous Measurement of Dose Delivery and Image Quality

Figure 4.3: Depiction of a typical ROI through use of a) a graphical representation of a typical selection area and b) the corresponding sampling distribution of the ROI area

4.2.8 Comparison of Image Quality Scores Between Protocols

The c-arm system protocols were categorised as high detail protocols or low dose protocols. The pre-upgrade protocols were then paired with their equivalent post-upgrade protocols. Dosimetry/image quality scores were compared in a graphical format and a tabular format. In tabular format, the protocols were compared directly by taking the ratio of the post-upgrade score (Score B) to the pre-upgrade score (Score A), as shown in Equation 4.2.

\[
Score Ratio = \frac{Score_B}{Score_A} \quad (4.1)
\]

\[
\sigma = \sqrt{\frac{\sum_{i=1}^{N}(x_i - \overline{x})}{N - 1}} \propto Image\ Noise \quad (4.2)
\]
4.3 C-arm Specific Methodologies

4.3.1 The Philips ‘Clarity’ Upgrade

The Philips Allura Xper FD20 c-arm system commissioned at Sutherland Heart Clinic underwent a post-processing systems upgrade dubbed the ‘Clarity’ upgrade. Pre-upgrade, the Philips Allura Xper FD20 c-arm system had been commissioned with two common use imaging protocols; the high imaging quality ‘Cardiac 4’ protocol and the low dose rate ‘Cardiac Low’ protocol. Both of these imaging protocols included an acquisition protocol and a fluoroscopy protocol. The fluoroscopy mode could be operated using one of three different exposure levels. Post-upgrade, the new Philips Allura ‘Clarity’ FD20 c-arm system replaced these protocols with with three different acquisition protocols; the boosted high detail ‘Clarity Boost’ protocol, the high detail ‘Clarity Normal’ protocol and the low dose rate ‘Clarity Low’ protocol. The ‘Clarity’ fluoroscopy mode protocol was common to all acquisition protocols and could also be operated using one of three exposure settings. A table of the Philips protocols observed has been provided in Table 4.1. The Philips c-arm system imaging protocols were assessed using the following metrics:

1. Dose delivery,
2. Iodine contrast resolution,
3. Spatial resolution,
4. Working thickness range,
5. Static motion target resolution, and,
6. Dynamic motion target resolution.
### Acquisition Mode Protocols

<table>
<thead>
<tr>
<th>Protocol</th>
<th>Machine Type</th>
<th>Commission Details</th>
<th>Protocol Type</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Cardiac 4</td>
<td>Philips Allura Xper FD20</td>
<td>Pre-upgrade Sutherland Heart</td>
<td>High Detail Acquisition Protocol</td>
</tr>
<tr>
<td>2. Cardiac Low</td>
<td>Philips Allura Xper FD20</td>
<td>Pre-upgrade Sutherland Heart</td>
<td>Low Dose Acquisition Protocol</td>
</tr>
<tr>
<td>3. Clarity Boost</td>
<td>Philips Allura Clarity FD20</td>
<td>Post-upgrade Sutherland Heart</td>
<td>Boosted High Detail Acquisition Protocol</td>
</tr>
<tr>
<td>4. Clarity Normal</td>
<td>Philips Allura Clarity FD20</td>
<td>Post-upgrade Sutherland Heart</td>
<td>High Detail Acquisition Protocol</td>
</tr>
<tr>
<td>5. Clarity Low</td>
<td>Philips Allura Clarity FD20</td>
<td>Post-upgrade Sutherland Heart</td>
<td>Low Dose Acquisition Protocol</td>
</tr>
</tbody>
</table>

### Fluoroscopy Mode Protocols

<table>
<thead>
<tr>
<th>Protocol</th>
<th>Machine Type</th>
<th>Commission Details</th>
<th>Protocol Type</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Cardiac 4</td>
<td>Philips Allura Xper FD20</td>
<td>Pre-upgrade Sutherland Heart</td>
<td>High Detail Fluoroscopy Protocol</td>
</tr>
<tr>
<td>2. Cardiac Low</td>
<td>Philips Allura Xper FD20</td>
<td>Pre-upgrade Sutherland Heart</td>
<td>Low Dose Fluoroscopy Protocol</td>
</tr>
<tr>
<td>3. Clarity</td>
<td>Philips Allura Clarity FD20</td>
<td>Post-upgrade Sutherland Heart</td>
<td>Clarity Fluoroscopy Protocol</td>
</tr>
</tbody>
</table>

*Table 4.2: Pre- and post- upgrade protocols installed on the Philips c-arm system commissioned at Sutherland Heart Clinic*
The Philips c-arm system’s assessment was performed using the 20 cm EPT and 30 cm EPT phantom configurations and using the 15 cm and 20 cm DFS FOV settings. The c-arm systems were evaluated in acquisition mode, fluoroscopy mode dose level 1 and fluoroscopy mode dose level 2. Fluoroscopy mode dose level 3 was not assessed as this imaging modality was not in common clinical use. The image quality of the acquired image sets was assessed in the clinic at the time of acquisition. This assessment was performed using the in-clinic monitors, the control room monitors and the DICOM files exported by the laboratory picture archiving and communication system (PACS). This test was performed to establish any variation in image quality introduced by differences in the viewing monitors and to ensure no image quality was lost during the conversion or compression of image sets into DICOM files by the PACS. The DICOM files were then exported from the PACS, anonymised, renamed and randomised before further assessments were performed. This minimised any bias which could occur when assessors read the names of the image set under evaluation. The DICOM files were assessed by a cohort of 20 assessors selected from the staff and student bodies of the University of Wollongong’s School of Physics. Choosing a large selection of assessors was important to reduce dependence of the results on individual proficiency. The assessment was performed using a designated desktop computer system using the RadiANT DICOM Viewer software package (Medixant, 2011, https://www.radiantviewer.com/) viewed in evaluation mode. The assessors were instructed to view each image set in full screen mode and without using any image enhancement tools offered by the RadiANT DICOM Viewer software package. These limitations ensured that variations due to the monitor type and monitor settings were minimised, that software proficiency based variations were minimised and that viewing conditions were as similar as possible to the those experienced in the catheterisation laboratory.
4.3. C-arm Specific Methodologies

4.3.2 The GE Innova 2100IQ c-arm system and the GE ‘Dose Map’ Upgrade

A similar upgrade was performed on the GE c-arm systems commissioned at Sutherland Heart Clinic and Eastern Heart Clinic. During this upgrade, the post-processing and imaging protocols were updated and dose contouring software, the ‘Dose Map’ software package, was installed on each system. In reference to this software package, the upgrade was referred to as the ‘Dose Map’ upgrade. The primary c-arm system studied during the ‘Dose Map’ upgrade was a GE Innova 2100IQ c-arm system based at Sutherland Heart Clinic. Pre-upgrade, the GE Innova 2100IQ c-arm system had been commissioned with two common use imaging protocols; the high imaging quality ‘CoroPlus’ protocol and the low dose rate ‘RDLS’ protocol. Each of these protocols included an acquisition mode and a fluoroscopy mode which could be then operated using one of two exposure settings, the dose level 1 or dose level 2 setting. Post-upgrade, the c-arm system featured three imaging protocols; the high detail ‘Standard’ protocol, the low dose rate ‘Low’ protocol and the low dose rate ‘Very Low’ protocol. All post-upgrade protocols were observed to operate using a default frame rate of 7.5 FPS as opposed to the 15 FPS frame rate used by all pre-upgrade protocols. Each of these protocols included an acquisition mode and a fluoroscopy mode. All post-upgrade protocols could also be operated in either the dose level 1 setting or the dose level 2 setting. All GE protocols assessed have been referenced in Table 4.2. The GE ‘Dose Map’ assessment was performed using the 20 cm and 30 cm EPT phantom configurations, the 12 cm and 15 cm OFS settings, both imaging modalities and both dose level settings. The post-upgrade fluoroscopy mode protocol ‘Low’ was also assessed separately with the imaging frame rate set to 15 FPS.
4.3.3 Comparing the GE Innova 2100IQ and GE IGS 520 c-arm systems

At the request of the vendor, the GE Innova 2100IQ was compared to a GE IGS 520 c-arm system commissioned at Eastern Heart Clinic. The primary difference between these two c-arm systems was firstly that the GE IGS 520 system featured a newer image-receptor detector plate and secondly that the system was installed with a different set of imaging protocols. Post-upgrade, the GE IGS c-arm system featured two imaging protocols; the high detail ‘IQ+’ protocol and the low dose ‘RDLS’ protocol. To differentiate this protocol from the pre-upgrade ‘RDLS’ protocol on the GE Innova 2100IQ c-arm system, this second ‘RDLS’ protocol was denoted as ‘EHC-RDLS’. These EHC protocols have been included in Table 4.3. The assessment was performed using the EHC protocols in acquisition mode only. A comparison was formed between the two of the post-upgrade IGS 520 protocols and two of the post-upgrade Innova 2100IQ protocols. For all GE ‘Dose Map’ related protocols, c-arm system performance was assessed using the standard NEMA XR21 protocol tests and using the digital assessment of noise as described in Section 4.2.7. Assessment was again performed using the in-clinic monitors, the control room monitors and the DICOM files exported by the PACS. The selection of protocols observed during the ‘Dose Map’ upgrade assessment was much more extensive than the protocol selection observed during the Philips ‘Clarity’ upgrade. This rendered visual assessment by a cohort as too time and resource consuming and as such visual assessment was performed by a single assessor. This enabled timely processing of the results.
### Acquisition/Fluoroscopy Protocols

<table>
<thead>
<tr>
<th>Protocol</th>
<th>Machine Type</th>
<th>Commission Details</th>
<th>Protocol Type</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. CoroPlus</td>
<td>GE Innova 2100IQ Pre-upgrade</td>
<td>High Detail Protocol</td>
<td></td>
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<tr>
<td>2. RDLS</td>
<td>GE Innova 2100IQ Pre-upgrade</td>
<td>Low Dose Protocol</td>
<td></td>
</tr>
<tr>
<td>3. Standard</td>
<td>GE Innova 2100IQ Post-upgrade</td>
<td>High Detail Protocol</td>
<td></td>
</tr>
<tr>
<td>4. Low</td>
<td>GE Innova 2100IQ Post-upgrade</td>
<td>High Detail Protocol</td>
<td></td>
</tr>
<tr>
<td>5. Very Low</td>
<td>GE Innova 2100IQ Post-upgrade</td>
<td>Low Dose Protocol</td>
<td></td>
</tr>
<tr>
<td>6. Low (15 FPS)</td>
<td>GE Innova 2100IQ Post-upgrade</td>
<td>Low Dose Protocol</td>
<td></td>
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<tr>
<td>7. EHC IQ+</td>
<td>GE IGS 520      Eastern Heart</td>
<td>High Detail Protocol</td>
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<tr>
<td>8. EHC-RDLS</td>
<td>GE IGS 520      Eastern Heart</td>
<td>Low Dose Protocol</td>
<td></td>
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</tbody>
</table>

**Table 4.3**: Pre- and post- upgrade protocols installed on the GE c-arm systems commissioned at Sutherland Heart Clinic and Eastern Heart Clinic.
4.4 Summary of the Results

In this section the summaries of the image quality study results have been provided. These summaries presented the results in a condensed, easy to read format. The section has been divided into the following subsections:

- The Philips System Upgrade
- The GE System Upgrade
- Comparison of the Upgraded GE protocols
- GE intersystem comparison

Some of the discussion points have referenced the results in greater details than provided in these summaries.

For the complete dataset analysed, please refer to Appendix A for the extended Philips system results and Appendix B for the extended GE system results.
4.4.1 The Philips System Upgrade

The Philips Allura Clarity upgrade summary is presented in Table 4.4. In acquisition mode, the upgraded Clarity protocols produced:

- + similar or lower ESDs delivered to the phantom.
- + higher iodine contrast resolution
- + higher spatial resolution
- - A narrower working thickness range
- - Lower motion target resolution

In fluoroscopy mode, the upgraded Clarity protocols produced:

- ≈ similar ESDs to the Cardiac 4 protocol
- - higher ESDs than the Cardiac Low protocol
- + higher iodine contrast resolution
- ≈ similar spatial resolution
- ≈ similar working thickness range
- - Lower motion target resolution
Table 4.4: Summary of the change in Philips c-arm system performance after the Clarity upgrade

<table>
<thead>
<tr>
<th>Imaging Mode</th>
<th>Pre-upgrade</th>
<th>Post-upgrade</th>
<th>Clarity Boost</th>
<th>Cardiac 4</th>
<th>Clarity Normal</th>
<th>Cardiac 4</th>
<th>Clarity Low</th>
<th>Cardiac Low</th>
<th>Fluoroscopy 1</th>
<th>Clarity 1</th>
<th>Fluoroscopy 2</th>
<th>Clarity 2</th>
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<th>Clarity 1</th>
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<td>Dose</td>
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<td>1.163</td>
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<td>1.036</td>
<td>1.242</td>
<td>1.259</td>
<td>1.163</td>
<td>1.275</td>
<td>1.036</td>
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<td>1.036</td>
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<td>1.259</td>
<td>1.163</td>
<td>1.198</td>
<td>1.275</td>
<td>1.036</td>
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<tr>
<td>Dynamic MTR</td>
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<tr>
<td>Static MTR</td>
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</table>

Table 4.4: Summary of the change in Philips c-arm system performance after the Clarity upgrade.
4.4.2 The GE System Upgrade

The GE post-processing upgrade summary is presented in Table 4.5. In acquisition mode, the upgraded GE protocols produced:

- + similar or lower ESDs delivered to the phantom.
- + higher iodine contrast resolution
- - lower spatial resolution
- \approx similar or broader working thickness range
- \approx similar motion target resolution
- + lower image noise levels

In fluoroscopy mode, the upgraded GE protocols performed as follows:

- Coronary Standard protocol significantly reduced ESD delivered to the phantom when compared to the CoroPlus protocol. This protocol produced similar or better working thickness ranges and motion target resolution, however, lower iodine contrast resolution and spatial resolution were observed and image noise was higher for this protocol.

- The Coronary Low protocol produced higher ESDs delivered to the phantom than the RDLS protocol. This improved iodine contrast resolution and motion target resolution while maintaining identical working thickness ranges and similar levels of image noise. Spatial resolution was reduced when imaging with this protocol.
### Table 4.5: Summary of the change in GE Innova 2100IQ c-arm system performance after the post-processing upgrade

<table>
<thead>
<tr>
<th>Imaging Mode</th>
<th>Pre-upgrade</th>
<th>Post-upgrade</th>
<th>Dose</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>ΔWTR</th>
<th>Dynamic MTR</th>
<th>Static MTR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Acquisition</td>
<td>Clarity Boost</td>
<td>Cardiac 4</td>
<td>0.807</td>
<td>1.339</td>
<td>1.259</td>
<td>0.829</td>
<td>0.975</td>
<td>0.970</td>
</tr>
<tr>
<td>Acquisition</td>
<td>Clarity Normal</td>
<td>Cardiac 4</td>
<td>0.458</td>
<td>1.163</td>
<td>1.237</td>
<td>0.778</td>
<td>0.994</td>
<td>0.947</td>
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<td>Cardiac Low</td>
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<td>0.958</td>
<td>0.977</td>
<td>0.984</td>
</tr>
<tr>
<td>Fluoroscopy 1</td>
<td>Clarity</td>
<td>Cardiac 4</td>
<td>0.640</td>
<td>0.379</td>
<td>0.870</td>
<td>0.909</td>
<td>0.750</td>
<td>0.762</td>
</tr>
<tr>
<td>Fluoroscopy 2</td>
<td>Clarity</td>
<td>Cardiac 4</td>
<td>1.036</td>
<td>1.538</td>
<td>0.974</td>
<td>1.024</td>
<td>0.811</td>
<td>0.796</td>
</tr>
<tr>
<td>Fluoroscopy 1</td>
<td>Clarity</td>
<td>Cardiac Low</td>
<td>2.597</td>
<td>1.242</td>
<td>0.985</td>
<td>1.012</td>
<td>1.022</td>
<td>1.050</td>
</tr>
<tr>
<td>Fluoroscopy 2</td>
<td>Clarity</td>
<td>Cardiac Low</td>
<td>1.446</td>
<td>1.574</td>
<td>1.048</td>
<td>1.262</td>
<td>0.821</td>
<td>0.892</td>
</tr>
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</table>

Table 4.5: Summary of the change in GE Innova 2100IQ c-arm system performance after the post-processing upgrade
4.4.3 The Upgraded GE System

The upgraded GE Innova 2100IQ system performance summary is presented in Table 4.6. In acquisition mode:

- ESD was ranked Standard < Low < Very Low
- Iodine contrast resolution was ranked Standard > Low > Very Low
- Working Thickness range was ranked Standard > Low > Very Low
- Motion target resolution was ranked Standard > Low > Very Low
- Image noise similar between protocols

In fluoroscopy mode:

- No consistent trend was observed for ESDs in fluoroscopy mode.
- Iodine contrast resolution was ranked Standard > Low > Very Low
- Working thickness range was ranked Standard > Low = Very Low
- Motion target resolution was ranked Low ≥ Standard > Very Low
- Image noise was ranked Standard > Low > Very Low

Increasing framerate in fluoroscopy mode resulted in:

- doubled ESD values
- higher iodine contrast resolution
- ≈ similar spatial resolution
- ≈ similar working thickness range
- + higher motion target resolution
- - higher image noise levels
### Table 4.6: Summary of the upgraded GE Innova 2100IQ protocol comparison

<table>
<thead>
<tr>
<th>Imaging Mode</th>
<th>Pre-upgrade</th>
<th>Post-upgrade</th>
<th>Dose</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>WTR₁</th>
<th>WTR₂</th>
<th>ΔWTR</th>
<th>Dynamic MTR</th>
<th>Static MTR</th>
<th>Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Acquisition</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>Standard</td>
<td>Low</td>
<td>0.793</td>
<td>0.866</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.250</td>
<td>1.050</td>
<td>1</td>
<td>1.125</td>
</tr>
<tr>
<td>2</td>
<td>Standard</td>
<td>Low</td>
<td>0.827</td>
<td>0.904</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Low</td>
<td>Very Low</td>
<td>0.780</td>
<td>0.843</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.937</td>
<td>0.965</td>
<td>1</td>
<td>0.854</td>
</tr>
<tr>
<td></td>
<td>Low</td>
<td>Very Low</td>
<td>0.798</td>
<td>0.880</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.938</td>
<td>0.965</td>
<td>1</td>
<td>0.918</td>
</tr>
<tr>
<td><strong>Fluoroscopy</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>Standard</td>
<td>Low</td>
<td>1.521</td>
<td>0.958</td>
<td>0.977</td>
<td>1</td>
<td>1</td>
<td>0.938</td>
<td>0.969</td>
<td>1.042</td>
<td>0.917</td>
</tr>
<tr>
<td>2</td>
<td>Standard</td>
<td>Low</td>
<td>0.973</td>
<td>1.758</td>
<td>0.973</td>
<td>1</td>
<td>1</td>
<td>0.938</td>
<td>0.969</td>
<td>1</td>
<td>0.938</td>
</tr>
<tr>
<td></td>
<td>Low</td>
<td>Very Low</td>
<td>0.470</td>
<td>0.358</td>
<td>0.952</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.875</td>
<td>0.792</td>
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<tr>
<td></td>
<td>Low</td>
<td>Very Low</td>
<td>1.097</td>
<td>0.494</td>
<td>0.977</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.938</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td><strong>Fluoroscopy</strong></td>
<td></td>
<td>Low (15FPS)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>Low</td>
<td>Low (15FPS)</td>
<td>2.045</td>
<td>1.617</td>
<td>1.050</td>
<td>0.938</td>
<td>1</td>
<td>0.958</td>
<td>1.250</td>
<td>1.375</td>
<td>1.182</td>
</tr>
<tr>
<td>2</td>
<td>Low</td>
<td>Low (15FPS)</td>
<td>2.055</td>
<td>1.275</td>
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<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.170</td>
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</tbody>
</table>
4.4.4 The GE Intersystem Comparison

The GE intersystem comparison is presented in Table 4.7. This comparison was performed between the upgraded GE Innova 2100IQ and the GE IGS 520 c-arm systems. For the high detail acquisition protocol the IGS 520 produced:

- $\approx$ Similar ESD values
- + higher iodine contrast resolution
- $\approx$ identical spatial resolution
- - similar or narrower working thickness range
- - similar or lower motion target resolution
- + reduced image noise

For the low dose acquisition protocol the IGS 520 produced:

- - higher ESD values
- + higher iodine contrast resolution
- + higher spatial resolution
- + broader working thickness range
- - similar or lower motion target resolution
- + reduced image noise
### Table 4.7: Summary of the comparison of the GE Innova 2100IQ and GE IGS 520 c-arm systems

<table>
<thead>
<tr>
<th>Imaging Mode</th>
<th>Pre-upgrade</th>
<th>Post-upgrade</th>
<th>Dose</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>$WTR_1$</th>
<th>$WTR_2$</th>
<th>$ΔWTR$</th>
<th>Dynamic MTR</th>
<th>Static MTR</th>
<th>Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>Acquisition 1</td>
<td>IQ+</td>
<td>Standard</td>
<td>1.049</td>
<td>1.190</td>
<td>1</td>
<td>0.875</td>
<td>0.958</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.656</td>
</tr>
<tr>
<td>Acquisition 2</td>
<td>IQ+</td>
<td>Standard</td>
<td>0.833</td>
<td>1.248</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.825</td>
<td>0.938</td>
<td>0.643</td>
</tr>
<tr>
<td>Acquisition 1</td>
<td>RDLs</td>
<td>Low</td>
<td>1.185</td>
<td>1.139</td>
<td>1.034</td>
<td>1</td>
<td>1.250</td>
<td>1.050</td>
<td>1</td>
<td>0.917</td>
<td>0.586</td>
</tr>
<tr>
<td>Acquisition 2</td>
<td>RDLs</td>
<td>Low</td>
<td>1.258</td>
<td>1.153</td>
<td>1.035</td>
<td>1</td>
<td>1.250</td>
<td>1.050</td>
<td>0.938</td>
<td>1</td>
<td>0.628</td>
</tr>
</tbody>
</table>
4.5 Discussion

4.5.1 Clinical impact of the Philips upgrade results

The ‘Clarity’ upgrade acquisition mode protocols were observed to produce similar or lower ESD deliveries than the previous Xper series protocols; the ‘Clarity Low’ protocol observed an average ESD comparable to the ‘Cardiac Low’ protocol, the ‘Clarity Normal’ protocol observed an average ESD reduction of 53.8% and the ‘Clarity Boost’ protocol observed an average ESD reduction of 22% relative to the relevant Xper series protocols. Despite this ESD reduction, the ‘Clarity’ protocols were observed to produce similar or better standards of image quality through improved iodine contrast resolution and spatial resolution. Static MTR was in general similar between the Clarity and Xper series protocols, however, the Clarity series protocols did on average score slightly lower for the dynamic MTR category and did observe a narrower WTR range resulting from both reduced upper and lower WTR threshold limits. The ‘Clarity Boost’ acquisition mode protocol was observed to produce similar x-ray tube characteristics to the ‘Clarity Normal’ acquisition mode protocol but with reduced x-ray tube filtration utilised during imaging. Overall, image quality was quite similar between protocols. As expected from a lower filtration x-ray beam, the ‘Clarity Boost’ protocol did increase ESD delivery to the surface by an average of 66.8% but also provide enhanced iodine contrast resolution by an average of 10.6%. Despite the reduced filtration, the ‘Clarity Boost’ protocol still delivered significantly less ESD delivery than the ‘Cardiac 4’ protocol, averaging ESD delivery values 22.9% lower than the Xper series protocol. The ‘Clarity’ upgrade fluoroscopy mode protocol were observed to produce higher ESD deliveries than the ‘Cardiac Low’ fluoroscopy mode protocols and lower ESD deliveries than the ‘Cardiac 4’ fluoroscopy mode protocols with exception to the ‘Cardiac 4’ protocol operated in fluoroscopy mode dose level 2 setting
when imaging the 20 cm EPT configuration which appeared to produce similar ESD deliveries as the ‘Cardiac 4’ protocol operated in fluoroscopy mode dose level 1 setting. This anomalous behaviour may have resulted from an algorithm imposed dose limit triggered when performing fluoroscopic imaging of smaller patient volumes. The Clarity series fluoroscopy mode protocol appeared to produce higher ESD delivery than the Xper series specifically when imaging the 20 cm EPT configurations which suggests the Xper series protocols were better optimised for imaging smaller patient volumes than the Clarity series fluoroscopy mode protocol. The Clarity series fluoroscopy mode protocol produced mixed image quality results relative to the Xper series fluoroscopy mode protocols with no consistent trends in comparative performance for the image quality tests. The Clarity series fluoroscopy protocols did on average perform better relative to the Xper protocols imaging the 20 cm EPT configuration which may have been related to the higher relative ESD deliveries observed when imaging 20 cm EPT configuration. Overall, the ‘Clarity’ upgrade successfully maintained or reduced the ESD delivery of the c-arm system while maintaining or improving upon the standard of image quality resolved in the acquired images. The published literature suggests that the Clarity upgrade has resulted in similar dose reduction capabilities in other angiographic catheterisation laboratories, however, the level of dose reduction achieved appears to vary depending both on patient and procedure type. The reviewed literature suggests that the ‘Clarity’ upgrade technology has reduced dose delivery in clinics by between 43 – 83% as based on clinician sampling of the air Kerma and dose area product metrics during procedures\textsuperscript{[139 - 146]}. 
Highlight conclusions from these collected studies have been compiled as follows:

- Cate et.al. observed the ‘Clarity’ upgrade reduced cine-acquisition DAP exposure by an average of 53%\(^{[141]}\).

- Spink et.al. observed the ‘Clarity’ upgrade reduced mean DAP by 57% and mean air Kerma by 58% during TIPS procedures\(^{[142]}\).

- de Ruiter et.al. observed 61% reduction in KAP per digital subtraction angiography frame when upgrading from a fixed Allura series c-arm system\(^{[143]}\).

- de Ruiter et.al. also observed that in fluoroscopy mode the Clarity series c-arm system produced comparable KAP values as their previous fixed Allura series c-arm system but observed KAP to triple when compared to their previous mobile c-arm system\(^{[143]}\).

- The Cate et.al. study also assessed image quality through blind review using a subjective semi-quantitative method. The panel of six experienced independent reviewers observed 85% of the ‘Clarity’ c-arm system image sets had either maintained or improved their standard of image quality when compared to similar image sets acquired by the contemporary c-arm system\(^{[141]}\).

Due to the use of different dosimetric metrics, which ranged from air Kerma, KAP, DAP and ESD, and due to the differences in imaging targets, procedure types and imaging projections, the level of dose reduction presented by these studies should not be conflated. That said, some degree of proportionality between these results can be assumed. The literature compares well to the results of the ‘Clarity’ upgrade c-arm system study, especially when comparing the Xper series ‘Cardiac 4’ protocol and the Clarity series ‘Clarity Normal’ protocol where an average decrease ESD of 54.2±6% was observed (error calculated using the 95% confidence interval method). The de
Ruiter et al. study suggests that the ‘Clarity’ upgrade featured a greater emphasis on improving acquisition mode performance while fluoroscopy mode performed comparably or even worse than previously utilised imaging protocols, suggesting that there is further room to optimise the Clarity series imaging protocols before hardware limits are reached\[143\]. The Cate et al. study potentially provided the most extensive commentary on image quality performance of the ‘Clarity’ upgrade and concluded that 85% of the ‘Clarity’ image sets were better or comparable to comparable image sets acquired using their previous c-arm system\[141\]. The referenced studies also noted that images were visually cleared on the upgraded Allura Clarity c-arm system. While this was not evidenced in our study through any specific test, subjective visual evaluation of the Xper series and Clarity series image sets were consistent with the published claims of reduced image noise. The reduction in noise was may have improved overall image quality as test features were observed to be easier to distinguish in the ‘Clarity’ image sets. A comparison of the Xper series images and Clarity series images has been provided in Figure 4.4 to visually present this difference in image quality between the two image sets. The algorithms used to produce this image noise reduction effect are discussed in detail by Söderman et al.\[144, 145\]. The contribution of image noise to image quality was not quantifiable using the established methodology suggesting a limitation in the NEMA XR 21 protocol assessment method. This resulted in a new digital-based test being implemented during the GE upgrade assessment that attempted to quantify image noise through sampling homogenous regions of the GE image sets.
4.5. Discussion

Figure 4.4: Example of acquisition mode image quality when using the a) Cardiac 4 protocol as compared to the b) Clarity Normal protocol

4.5.2 Clinical impact of the GE upgrade results

The GE ‘Dose map’ upgrade introduced three new imaging protocols, the ‘Standard’, ‘Low’ and ‘Very Low’ protocols. As the primary clinical protocols, the ‘Standard’ protocol was compared directly to the pre-upgrade ‘CoroPlus’ protocol as high detail imaging protocols and the ‘Low’ protocol was compared directly to the pre-upgrade ‘RDLS’ protocol as low dose protocols. As a secondary low dose clinical protocol, the ‘Very Low’ protocol was only discussed through the context of the ‘Low’ protocol. The post-upgrade ‘Standard’ protocol provided significant reduction in ESD delivery when compared to the pre-upgrade ‘CoroPlus’ protocol, averaging an ESD reduction of 26% when operated in the acquisition modes and 61.9% when operated in fluoroscopy mode. In acquisition imaging modalities, iodine contrast resolution improved by an average of 13.3% and image noise decreased by an average of 41.8%, however, in fluoroscopy mode, these aspects of image quality were observed to depreciate by averages of 39.3% and 17.9% respectively, likely resulting from photon deprivation at the image receptor. Spatial resolution was observed to reduce consistently for the ‘Standard’
4.5. Discussion

protocol across all EPT/FOV configurations and imaging modalities by an average of 14.6%. Working thickness range was similar for most EPT/FOV configurations and imaging modalities, however, in some configurations the ‘Standard’ protocol exhibited improved capabilities in imaging high density WTR targets. Motion target resolution was similar between protocols with the exception of when imaging using the ‘Standard’ protocol operated in the acquisition mode dose level 2 setting where an average of 22.5% less motion targets were acquired when imaging the 12 cm OFS configurations. The GE ‘Dose map’ upgrade ‘Low’ protocol ESD deliveries were comparable to the ‘RDLS’ protocol in acquisition mode, averaging to within 1% of the ‘RDLS’ dose values. When imaging using fluoroscopy mode, the ‘Low’ protocol dose values were also similar to the ‘RDLS’ protocol dose values when imaging the 30 cm EPT configurations, scoring an average of 4.4% higher than the ‘RDLS’ protocol, but was significantly higher when imaging the 20 cm EPT configurations, scoring an average of 249.4% higher than the ‘RDLS’ protocol dose values. Iodine contrast resolution improved for all ‘Low’ protocol images by an average of 34.4% with exception to when imaging the 20 cm EPT configurations in fluoroscopy mode where iodine contrast resolution was observed to depreciate by 26.8%. Spatial resolution was consistently lower by an average of 12.6% with exception to when imaging the 30 cm EPT configurations using the fluoroscopy mode where spatial resolution was observed to improve by an average of 5.6%. WTR/MTR was generally similar or improved upon when using the ‘Low’ protocol. Image noise decreased by an average of 24.8% in the acquisition modes, decreased by an average of 8.9% when imaging in fluoroscopy mode using the 12 cm OFS configurations and increased by an average of 25.4% when imaging in fluoroscopy mode using the 15 cm OFS configurations. When using fluoroscopy mode to image the 20 cm EPT configurations, the ‘Low’ protocol results appeared to perform worse than the ‘RDLS’ protocol despite outperforming the ‘Low’ protocol in every other
imaging modality, EPT configuration and FOV configurations which suggests that the ‘Low’ protocol may not be suitably optimised for imaging smaller patient volumes. The GE ‘Dose map’ upgrade ‘Very Low’ protocol ESD deliveries were lower than the ‘Low’ protocol ESD deliveries by an average of 34.1% when imaging the 20 cm EPT configuration in acquisition mode, lower than the ‘Low’ protocol ESD deliveries by an average of 8.1% when imaging the 30 cm EPT configuration in acquisition mode, lower than the ‘Low’ protocol ESD deliveries by an average of 53.0% when imaging with the fluoroscopy mode dose setting 1 and higher than the ‘Low’ protocol ESD deliveries by an average of 9.7% when imaging with the fluoroscopy mode dose setting 2. When imaging in the acquisition modes, the ‘Very Low’ protocol produced similar or worse standards of iodine contrast resolution and dynamic MTR compared to the ‘Low’ protocol by averages of 13.8% and 11.4% respectively. The ‘Very Low’ acquisition mode protocol also produced a similar standard of spatial resolution, static MTR, image noise and upper WTR threshold, however, the ‘Very Low’ acquisition mode protocol was capable of acquiring an average of 17.5% more high density imaging targets than the ‘Low’ protocol when imaging the 20 cm EPT configuration and an average of 30% less high density imaging targets than the ‘Low’ protocol when imaging the 30 cm EPT configurations. In the fluoroscopy modes, the ‘Very Low’ protocol produced a similar or lower standard of iodine contrast resolution, spatial resolution and MTR and acquired less image noise than the ‘Low’ protocol image sets by an average of 4.1%. It was noted that the ‘Very Low’ protocol performed significantly worse when imaging the 30 cm EPT configurations using the fluoroscopy modes. Three out of four of these configurations produced images that failed to acquire even a single iodine dot. This constituted extremely poor image quality, prevented an accurate comparison to other protocols using the established ratio method and further adds to evidence that certain post-upgrade protocols could be optimised to specific EPT ranges. Some general be-
havioural trends were observed between the post-upgrade protocols. The x-ray tube kVp and x-ray tube current were observed to increase with phantom EPT. Increasing kVp would increase the penetration of the x-ray beam while increasing the x-ray tube current would increase the number of photons incident on the image receptor plate, both of which would be necessary to maintain a standard of image quality when imaging larger, more attenuative, patient volumes at the cost of increasing exposure to the patient, which was reflected in the results. When changing from the dose level 1 to the dose level 2, the x-ray tube kVp was observed to be invariable for each protocol, suggesting kVp was primarily selected based on EPT, while x-ray tube current was observed to increase and x-ray tube filtration either reduced or was maintained at the previous filtration level. The change in these settings would increase number of photons incident to the patient volume, increase the photons transmitted to the image-receptor and increase the low energy photon contribution of the beam which would increase image contrast. These effects increase image quality at the expense of increasing the ESD delivered to the patient volume, which was reflected in the results.

In all instances, the Very Low protocol was observed to utilise a similar level of beam filtration as the ‘Standard’ protocol for all EPT/FOV configurations and an increased level of beam filtration as the ‘Low’ protocol with exception to when imaging the 30 cm EPT configurations in acquisition mode where all protocols were observed to utilise no additional beam filtration. The ‘Very Low’ protocol was also observed to generate similar kVp values to the ‘Standard’ and ‘Low’ protocols when imaging the 20 cm EPT configurations but the ‘Very Low’ protocol kVp values were lower when imaging the 30 cm EPT configuration, suggesting that the relationship between kVp values and EPT changed differently depending on the protocol used. When considered with the fact that the ‘Very Low’ protocol performed better than the ‘Low’ protocol when imaging the 20 cm EPT configuration but worse when imaging the 30 cm
EPT configuration and that the ‘Low’ protocol performed better than the ‘Standard’ protocol when imaging the 20 cm EPT configuration but worse when imaging the 30 cm EPT configuration, the results could suggest that rather than the protocol naming convention outlining a specific dose delivery hierarchy, the protocol naming convention could instead indicate that the ‘Standard’ protocol is in fact used for ‘standard’ sized patients while the ‘Low’ and ‘Very Low’ protocols are instead used for smaller sized patients which as previously established, would constitute lower dosed procedures just by being less attenuative than larger patients, however, this line of thinking is purely speculative based upon generalisations of the observations. In fluoroscopy mode, the post-upgrade fluoroscopy protocols were observed to perform at a lower framerate of 7.5 FPS as opposed to the pre-upgrade default setting of 15 FPS. This change was only observed in the fluoroscopy modes. The impact of decreasing the framerate of the ‘Low’ protocol was evaluated at the request of the local clinicians. Increasing the framerate by a factor of two more than doubled the ESD delivery but also resulted in small improvements to both ICR and spatial resolution. Increasing frame rate also produced a similar or better standard of MTR and resulted in reduced temporal artefacts, specifically imaging after-shadows. Increasing framerate did not improve image noise and instead resulted in an average increase in image noise of 17.6%.

4.5.3 Direct system comparison: Innova 2100IQ vs IGS 520

The EHC IGS520 c-arm system had undergone a similar post-processing upgrade as the SHC Innova 2100IQ c-arm system. As mentioned previously, the primary difference between these two c-arm systems was that the IGS520 c-arm system included an upgraded image-receptor plate which featured a greater number of detector elements than included in the Innova 2100IQ image-receptor plate. As such, this c-arm system comparison provides interesting insights both into the differences in c-arm system
commissioning between different clinical environments and into the impact of software upgrades versus incremental hardware upgrades. The ‘Standard’ protocol produced higher ESD deliveries than the ‘IQ+’ protocol when imaging the 20 cm EPT configurations using the acquisition mode 1 setting by an average of 22.5%, produced lower ESD deliveries than the ‘IQ+’ protocol when imaging the 30 cm EPT configurations using the acquisition mode 1 setting by an average of 12.4% and produced lower ESD deliveries than the ‘IQ+’ protocol when imaging using the acquisition mode 2 setting by an average of 16.8%. The ‘Standard’ protocol also produced an improved standard of both iodine contrast resolution and image noise by averages of 21.9% and 35.1% respectively. In general, when imaging using the acquisition mode dose level 1 setting, both protocols produced comparable spatial resolutions, working thickness ranges and MTR values. In general, when imaging using the acquisition mode dose level 2 setting, both protocols produced comparable spatial resolutions, working thickness ranges and dynamic MTRs, however, the ‘Standard’ protocol resolved 17.5% less static motion targets than the ‘IQ+’ protocol image sets. Image noise was lower when imaging using the ‘Standard’ protocol by an average of 35.1%. The ‘Low’ protocol produced higher ESD deliveries than the ‘EHC-RDLS’ protocol when imaging the 20 cm EPT configurations by an average of 41.8% when imaging in the acquisition mode dose level 1 setting and by an average of 55.6% in the acquisition mode 2 dose level 2 setting. When imaging the 30 cm EPT configurations, the ‘Low’ protocol produced similar or lower ESD deliveries than were on average 4.3% lower than the ‘EHC-RDLS’ protocol ESD deliveries. Iodine contrast resolution was in general higher when using the ‘Low’ protocol, averaging 14.6% higher than the ‘EHC-RDLS’ protocol. Spatial resolution and WTR scores were similar between protocols with exception to the 20 cm EPT, 12 cm OFS configuration where the ‘Low’ protocol resolved more line pairs and more high density imaging targets than the ‘EHC-RDLS’ protocol. MTR was similar
between protocols when imaging the 20 cm EPT configurations and was similar or reduced when using the ‘Low’ protocol to image the 30 cm EPT configurations. Image noise was lower when using the ‘Low’ protocol for all configurations by an average of 39.3%. Overall, the SHC Innova 2100IQ c-arm system produced image sets with better iodine contrast resolution and image noise scores than the IGS-520 c-arm system but otherwise the two c-arm systems produced comparable image quality scores. This was despite the IGS520 c-arm system being a newer hardware revision that included an improved image-receptor plate with a greater number of detector elements. The results suggest that current improvements in post-processing techniques may impact image quality more significantly than incremental hardware releases and may suggest that c-arm system upgrades may be a more efficient route to updating catheterisation laboratories than replacing existing units, although, further c-arm system image quality studies would be necessary to prove this to be the case.

4.5.4 Developing a better definition of ‘Image Quality’

The results of this study enable reflection on what defines good image quality. When comparing the high detail ‘Standard’ protocol and the low dose ‘Low’ protocol, the primary differences between the two protocols were dose delivery, iodine contrast resolution and image noise. The results suggest that iodine contrast resolution and image noise are the most apparent indicators of image quality and that these factors are inversely proportional to ESD delivery. Other image quality factors, such as spatial resolution, imaging working thickness range and motion target resolution were largely consistent between the ‘Standard’ and ‘Low’ protocols, suggesting that these parameters are a product of hardware and algorithmic limitations. The results have proved that c-arm system efficiency can be improved upon, however, to some extent the ESD delivery and image quality produced by the c-arm system is largely dependent on the
preferences of the operators performing the procedures and there is a trade-off presented when selecting the level of c-arm system performance. Image noise reduction was an important aspect for both the ‘Clarity’ and ‘Dose Map’ upgrades. The ‘Clarity’ upgrade attempted to achieve this through an improved reconstruction methodology while the ‘Dose Map’ upgrade seemed to achieve this through application of imaging smoothing filters. If smoothing filters were implemented in the GE ‘Dose Map’ upgrade, it could also explain the consistent reduction in spatial resolution observed in the majority of post-upgrade image sets. These techniques resulted in images appearing less granular than the pre-upgrade image set, as shown in Figure 4.4. In addition, several other post-processing effects were noted post-upgrade on the GE Innova 2100 IQ c-arm system. The boundaries of high contrast regions, such as the WTR test features, were observed to exhibit edge enhancement effects. These effects were observable as a visual glow or shadow around the test features. This effect accentuated these features creating a sharp visual distinction between mediums. This edge enhancement effect has been presented in Figure 4.5. This figure includes two plot profiles generated in the FIJI image analysis software package that graphically present the edge enhancement effects as seen applied to the high and low density WTR targets. Visually homogenous regions of the post-upgrade protocol image sets were also observed to feature a slight radial gradient that extended centrally to the edges of the image. This inhomogeneity effect has been presented in Figure 4.6. This effect could have been a result of beam hardening as previously discussed or potentially resulted from the presence of the contrast objects in field. As such, sampling noise could potentially be improved through use of an equivalent water phantom. That said, the ROI was selected centrally to minimise the impact of this gradation of the field and the sampling distribution was of the expected Gaussian trend as presented in Figure 4.3b. This fostered confidence in the testing methodology which will be improved upon further
in future work through measurement of the signal to noise ratio, the contrast to noise ratio and the modulation transfer function.

4.5.5 The efficacy of this study’s methodology

This study compared performance of c-arm system protocols using a ratio method. The ratio method was highly effective for comparing similar protocols. Unfortunately, the magnitudes of the ratio method results were, in part, dependent on the scale of the protocol, imaging modality, phantom and c-arm configurations used. For low scoring protocols or configurations, small variations in score affected the magnitude of the ratios more significantly than when comparing high scoring protocols. This was a purely mathematical consequence of comparing such a broad range of score values. The ratio method was also not capable of providing a sense of relative performance ranking when comparing individual protocols to the full range of protocols evaluated. The graphical representation provided in Appendix B, successfully depicted performance ranking between protocols, however, the graphs were complicated to read due to the number of variables displayed. This complexity of the graphical representation would only compound with an increasing number of comparative protocols, hence why the graphical representation was not used to compare the GE c-arm system results. A more effective scoring methodology could involve developing scoring thresholds, or scoring brackets, based upon images that were considered to be, for example, of good, acceptable or bad quality by a consensus of clinicians. Such a method would require sampling of a broader c-arm system performance range than was observed in this study and would require collaboration with a range of clinical experts, both of which were conditions beyond the scope of this study but may be revisited for future measurements. According to a study by Balter et.al in 2001, a study that introduced the NEMA XR21 methodology, the Society for Cardiac Angiography and Interventions
4.5. Discussion

Figure 4.5: a) An isolated post-upgrade GE Innova 2100IQ protocol acquisition frame presenting edge enhancement effects at the boundary of high contrast image features with line profiles provided for a b) low density WTR imaging target and a c) high density WTR imaging target.

Figure 4.6: a) An isolated post-upgrade GE Innova 2100IQ protocol acquisition frame presenting radial gradient effects as shown graphically through b) line profile.
initially intended to establish a system performance registry based on standardised NEMA XR21 protocol testing\textsuperscript{[138]}. To date, this performance registry project does not appear to have been established. From the findings of this study, the most significant hurdle to developing a performance registry based on the NEMA XR21 protocol is the methodologies reliance on the ability and availability of trained human observers. The number of machines, protocols, and standardised imaging configurations that are necessary to evaluate performance generates a considerable quantity of data to assess. Widespread implementation of the existing methodology would be time consuming, resource intensive and open to assessor proficiency biases. The methodology could be streamlined through development of an image assessment software package that deploys sophisticated feature recognition algorithms and calculates digital indications of image quality such as image noise, signal to noise ratio and contrast to noise ratio. Such a software package would enable operators to processes image set batches significantly faster than could be performed by trained human observers. The NEMA XR 21 protocol design could also improved upon. During the c-arm upgrade evaluations, the ESD delivery, image quality and the general c-arm system performance varied greatly between the 20 cm EPT and 30 cm EPT phantom configurations. C-arm systems and imaging protocols may be optimised to specific expected patient thicknesses. An expanded range of EPT configurations using the NEMA XR21 image test specifications would provide better resolution to these results and improve clinical understanding of the strengths of specific protocols. The evaluation methodology could be further improved by increasing the range of existing tests. A study by Simon et.al. compared two of most popular c-arm system evaluation phantoms, the American NEMA XR21 protocol test object and the European Leeds TOR 18FG test object\textsuperscript{[146]}. Simon et.al. considers that there are more sophisticated test objects, but asserts that these test objects are simple enough to enable routine quality assurance image quality evalu-
4.5. Discussion

The study used a digital sampling methodology to establish image quality and measured dose delivery using the incident air Kerma metric as measured by an Unfors Xi detector. In their discussion of the results, Simon et al. commented that the iodine circled embedded in the NEMA XR21 protocol test object were smaller than the Leeds test object circles and contained less iodine content which resulted in greater sources of error in the NEMA XR21 results, especially when assessing larger field sizes. The Leeds test object also included both a larger range and resolution of iodine contrast samples and, as iodine contrast resolution was a primary indicator of image quality, may provide greater details regarding this aspect of c-arm system performance than the currently existing NEMA XR21 protocol specified tests. Other studies yet have attempted to adapt the specified phantom volume to new testing criteria or have used the phantom volume as a teaching tool in clinic, a notion that represents the latent value in the NEMA XR21 protocol [147, 148].

4.5.6 Air Kerma and the importance of implementing real-time surface dosimetry solutions

For all post-upgrade GE Innova 2100 IQ and GE IGS 520 measurements, the machine calculated air Kerma at reference point was recorded and compared to the ESD delivery as measured by the MOSkin. air Kerma and ESD are by no means equivalent dosimetric quantities, however, dose contouring software, such as the ‘Dose Map’ software package, rely on the air Kerma quantity to advise clinicians of deterministic risk and as such direct comparison of these quantities was of interest to this study. For the purposes of this comparison, the quantities will be compared using the ratio described in Equation 4.3. The results of the comparison have been presented in Table 4.8.
for the acquisition mode results and in Table 4.9 for the fluoroscopy mode results.

\[
\frac{K_{\text{ReferencePoint}}}{E_{\text{SD} \text{MOSkin}}} \quad (4.3)
\]

The results observed that in general, the machine estimated air Kerma underestimates the surface ESD. In acquisition mode, the air Kerma estimation values were an average of 28% lower than the MOSkin measured ESD values. This falls within the expected error range for dose map of 40% as calculated in a verification study performed by Bordier et.al. in 2015 and was similar in magnitude to their verification that found dose map values were within 24.9% of their EBT3 based measurement of the patient ESD\textsuperscript{[149]}. Errors were much larger during fluoroscopy mode measurements and overestimated the surface ESD delivered to the phantom volume in some of the 20 cm EPT configuration measurements. This increase in error may have resulted from the reduced x-ray beam filtration used when imaging in fluoroscopy mode affecting beam quality and scatter. The 20 cm EPT configurations may also have been affected more significantly due to geometric differences in the measurement and reference point when imaging the 20 cm EPT phantom configuration.
### 4.5. Discussion

Table 4.8: Air Kerma to ESD ratio as recorded for each acquisition mode measurement performed during the c-arm system performance study.

<table>
<thead>
<tr>
<th>Acquisition Mode</th>
<th>Protocol</th>
<th>C-arm Parameters</th>
<th>Settings</th>
<th>$K_{\text{ReferencePoint}}^{\text{ESD MOSkin}}$</th>
<th>Protocol Average</th>
<th>C-arm System Specific Averages</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Standard</td>
<td>FOV = 12 cm</td>
<td>EPT = 20 cm</td>
<td>0.79</td>
<td>0.69</td>
<td>0.88</td>
</tr>
<tr>
<td></td>
<td></td>
<td>FOV = 12 cm</td>
<td>EPT = 30 cm</td>
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<td>0.69</td>
<td>0.88</td>
</tr>
<tr>
<td></td>
<td></td>
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<td></td>
<td></td>
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<td>Dose Level 2</td>
<td>0.94</td>
<td>0.75</td>
<td>0.88</td>
</tr>
<tr>
<td></td>
<td>Low</td>
<td>FOV = 12 cm</td>
<td>EPT = 20 cm</td>
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<td>0.67</td>
<td>0.81 SHC Innova2100IQ</td>
</tr>
<tr>
<td></td>
<td></td>
<td>FOV = 12 cm</td>
<td>EPT = 30 cm</td>
<td>0.77</td>
<td>0.67</td>
<td>0.81 SHC Innova2100IQ</td>
</tr>
<tr>
<td></td>
<td></td>
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<td>0.82</td>
<td>0.69</td>
<td>0.81 SHC Innova2100IQ</td>
</tr>
<tr>
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<td></td>
<td>FOV = 15 cm</td>
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<td>0.69</td>
<td>0.81 SHC Innova2100IQ</td>
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</tr>
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<td></td>
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<td>0.74</td>
<td>0.63 EHC IGS520</td>
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Table 4.8: Air Kerma to ESD ratio as recorded for each acquisition mode measurement performed during the c-arm system performance study.
### Table 4.9: Air Kerma to ESD ratio as recorded for each fluoroscopy mode measurement performed during the c-arm system performance study

<table>
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<th>Fluoroscopy Mode</th>
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<td>0.47</td>
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<tr>
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<td></td>
<td></td>
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<td></td>
<td></td>
<td>FOV = 15 cm</td>
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<td>0.53</td>
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</tr>
<tr>
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<td>0.61</td>
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<td>Dose Level 1</td>
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<td></td>
<td></td>
<td>FOV = 12 cm</td>
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<tr>
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<td>2.95</td>
<td>0.72</td>
<td>1.14</td>
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<tr>
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<td>0.60</td>
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<tr>
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<tr>
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<td>FOV = 12 cm</td>
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<td>FOV = 15 cm</td>
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<tr>
<td></td>
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<tr>
<td></td>
<td></td>
<td>FOV = 12 cm</td>
<td>1.30</td>
<td>0.57</td>
<td>0.93</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Dose Level 2</td>
<td></td>
<td></td>
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</tr>
</tbody>
</table>

**EPT Specific Average Values**

- 1.30
- 0.57
- 0.93
4.6 Conclusions

This study established that significant improvements to c-arm system performance can be achieved through post-processing imaging chain upgrades, refinement of image reconstruction algorithms and optimisation of clinical imaging protocols. This study also concluded that c-arm system upgrades enable older machines to perform comparably to newer hardware revisions which suggests that upgrading c-arm systems may in some cases be more efficient than replacing older c-arm systems with incremental hardware upgrades. The NEMA XR21 protocol was an effective measure for c-arm system performance and was capable of providing a clear comparison between individual c-arm systems and protocols, however, the protocol lacked any established simple method to clearly rank multiple c-arm systems and their imaging protocols. The NEMA XR21 protocol could be improved upon through the inclusion of digital image quality assessment software, expanded testing parameter ranges and additional phantom thickness configurations. Future modifications or revisions of the NEMA XR21 protocol would be useful for benchmarking and optimising c-arm system performance and would benefit catheterisation laboratories both in informing prospective purchase decisions and in enabling routine quality assurance testing. The MOSkin results were comparable to similar literature based reviews of the Philips issued ‘Clarity’ upgrade. The c-arm system calculated air Kerma value, which is often used for estimating deterministic risk to patients, was observed to overestimate ESD deliveries in acquisition mode. The observed discrepancy of air Kerma to ESD ranged within established error limits, however, fluoroscopy mode error values observed a much broader range, potentially due to the lower filtration values used in acquisition mode protocols. air Kerma values were also in general higher relative to the measured ESD when using the 20 cm EPT phantom configuration as compared to the 30 cm EPT phantom configuration results with some protocols observed to overestimate the ESD deliveries to the phantom. As
such, methods for assessment and prevention of deterministic risk would clearly benefit from the incorporation of supplementary data provided through direct dosimetry solutions.
Chapter 5

Development of Operator Dosimetry Solutions and Methodologies to Address Emerging Health Concerns

In previous chapters there has been a focus on patient risk during interventional angiographic procedures, however, recent studies have reported that experienced clinicians could be developing their own radiation-induced symptoms. Over the course of a decades long career clinicians may be at even greater risk of radiation-induced conditions negatively impacting their quality of life than their patients are. This chapter will discuss radiation-induced conditions reported by interventional clinicians, established and developing methods to measure clinician exposures and considerations that are necessary for developing better operator dosimetry solutions.
5.1 Operator Specific Dosimetry Solutions

5.1.1 Radiation-induced risks to interventional clinicians

Throughout decades of ICRP publications, the ICRP has chronicled the ever-developing understanding of radiation-induced risks to the general population\(^2\). Radiation-induced risk modelling was first developed on observation of uranium/radon exposures primarily in mining industry. These models have been improved upon as new information has been presented. When classifying radiation-induced conditions, the ICRP cite epidemiological studies that detail the health outcomes of atomic weapon survivors, radiotherapy patients, patients undertaking therapeutic/diagnostic procedures and occupationally exposed workers. As an example, Shimizu et.al. reported that low dose exposures increased incidence of stroke and heart disease bomb survivors from radiation exposures exceeding 500 mSv\(^{[150]}\). Unfortunately, the ICRP lists the following limitations to this approach\(^2\):

- epidemiological studies are open to interpretation
- stochastic conditions can take decades to manifest
- confounding factors may contribute to or cause the condition under study
- stochastic conditions occur with statistical distribution
- individuals may exhibit susceptibility or resistance to radiation-induced conditions which could affect the conclusions of the study.

These limitations have meant that the understanding of radiation-induced conditions, especially for low dose exposures, has been a developing area of study under constant revision. As an example, ICRP Publication 103 proposed significant changes to the ICRP Publication 60 tissue weighting factors due to the publication of further
epidemiological studies\textsuperscript{2,3}. Innovations in imaging technology and techniques have also affected the working conditions and workloads of clinicians significantly.

There are many reports of potentially radiation-induced conditions affecting interventional clinicians. Publicised conditions have included erythema, loss of hair on legs, hands and arms, heart disease, neurological disorders including brain tumours or stroke and rear eye lens cataract formation\textsuperscript{[151 - 159]}. Of these, studies on cataract formation have lead to the most significant revision to ICRP guidance since ICRP Publication 103. This chapter will focus specifically on the issue of eye lens dose in catheterisation laboratories and potential solutions to protect clinician health.

### 5.1.2 Radiation-induced injury to Clinician Eye Lens

Of all radiation-induced conditions currently under scrutiny, there is a specific interest in monitoring eye lens exposures. Clinical case studies have observed that retiring interventional operators have been developing rear eye lens cataracts at a rate four to five times than the general public control group\textsuperscript{[61]}. These observations suggest that the eye lens is considerably more sensitive to radiation than previously anticipated. These reports triggered a worldwide revision of eye lens exposure limits with the ICRP recommending in ICRP Publication 118 (2012) that eye lens exposures should be reduced to beneath 20 mSv/year as averaged over 5 years with no yearly exposures exceeding 50 mSv/year as compared to the previous standard of 150 mSv/year as averaged over 5 years\textsuperscript{[32]}.

Eye lens doses came under scrutiny on the publication of several multi-centre studies including the ORAMED and RELID studies\textsuperscript{[160, 161]}. The retrospective evaluation of eye lens injuries and dose study (RELID) was an international study implemented by the IAEA in 2008. The aims of the RELID study were to retrospectively evaluate the history of clinician eye lens exposure and radiation induced injury. This was
achieved by reviewing dosimetric records, by assessing the past behaviours of clinicians and by assessing clinician eye lens health using a variety of tests. Interventional cardiologists of ages 30-60 years old with clinical careers up to 40 years in length were assessed. The RELID study discovered 45% of interventional cardiology respondents exhibited damage to the posterior subcapsular lens in one or both eyes with evidence suggesting radiation-induced cause, 57% reported never using eye protection and 61% reported never using lead shielded screens. The induced condition rate compared to a control group incident rate of 12%. The conclusions of this study demonstrated the prevalence of eye lens issues in cardiologists, exemplified the history of behaviours in interventional fields and exemplified the necessity to change behaviours through improving awareness of radiation-induced risks and occupational exposures. Currently the understanding of how radiation-induced cataracts form is still developing. Competing perspectives have debated whether this is a stochastic or deterministic effect[61]. Conversely, the ORAMED project studied staff eye lens exposures, developed a phantom for calibrating future eye lens dosimeters and developed the specifications for an eye lens dosimeter enclosure that could measure at the $H_p(3)$ measurement depth.

To further bolster understanding, eye lens dose needs to be measured regularly in clinics and common methodologies need to be adopted on a broader scale so that the information collected is useful to researchers.

5.1.3 Requirements for eye lens specific dose monitoring

Despite published concerns, information regarding clinician exposures is limited. The primary measures of operator exposure are currently through correlation with c-arm system reported air Kerma or $H_p(10)$ depth personal radiation monitors worn at chest or collar height. Studies performed by Estathopoulos et.al. and Nikodemová et.al. observed that air Kerma was a poor indicator of the extremity dose[161, 162]. For eye
lens dose, Sanchez et al. observed that eye lens dose could be conservatively estimated as approximately a factor 0.8 of the chest level $H_p(10)$ monitor readout but such estimations can introduce large variations and errors ($>40\%$) due to geometry related factors\textsuperscript{163}. From their findings, Sanchez et al. concluded that with proper radiation protective equipment and appropriate in-room shielding, most clinicians should not exceed current occupational dose limits, however, clinics that do not utilise appropriate radiation protection strategies, especially clinics that specialise in more complicated procedures, could easily exceed the dose limits unintentionally during their standard workloads\textsuperscript{163}.

Eye lens dose can be monitored in the clinic in a variety of ways. Landauer include provide $H_p(3)$ dose estimates based on $H_p(10)$ dosimeter readouts for occupationally exposed workers. The ICRU recommends eye lens dose should be measured using the $H_p(3)$ measurement depth however the ICRP have also suggested that $H_p(0.07)$ dosimeters could be used as appropriate substitutes for $H_p(3)$ dosimeters\textsuperscript{61, 97}. Commercially there are only two $H_p(3)$ dosimeter enclosures available to clinics at this time, the Radcard Eye-D and the Landauer Vision TLD enclosures\textsuperscript{164 - 166}. Both enclosures are pictured in Figure 5.1 alongside the MOSkin dosimeter. The dosimeters are implemented by some clinics but are not a required measure of dose.

Instructions on placement of these dosimeters varies with Radcard recommending that the Eye-D dosimeter should be worn on the side of the operators’ radioprotective glasses and Landaur recommending that the Vision should be worn on the underside of the operators’ radioprotective glasses\textsuperscript{164, 166}. ICRP Publication 139 also specified studies that measured eye lens exposures from the operator’s left eyebrow and the operator’s forehead\textsuperscript{61}. While there is general consensus that eye lens dosimeters should measure exposure to the eye closest to the c-arm system, typically the left eye, the measurement point itself is yet to be standardised\textsuperscript{160}. The ICRP concluded that
the appropriate position may vary depending on the procedure [160].

To further protect clinicians, operator eye lens doses must be monitored to ensure compliance to the new dose limits. Barnard et al. have suggested that while compliance with the new dose limits are achievable on a clinical basis, many clinics in 2016 were at risk of non-compliance [168]. The authors of this study recommended that renewed radiation safety training courses, exposure monitors and appropriate personal protective equipment must be deployed to ensure clinician safety. They also recommended that cataract reporting methodologies needed to be standardised. The aims of the studies performed in this chapter include:

1. To compare different eye lens dose reference points using clinically representative beam qualities and phantom configurations.

2. To compare the efficacy of the MOSkin dosimeter in comparison to existing gold standard TLD dosimetry solutions.

3. To test a prospective \( H_P(3) \) MOSkin encapsulation intended for operator dosimetry specific applications.

4. To assess the suitability of MOSkin technologies in operator dosimetry applications.

5. To investigate considerations for development of future operator dosimetry solutions.
Figure 5.1: Eye lens dosimeters that have been used clinically include the a) Radcard Eye-D, b) the Landauer Vision, Hp(10) depth chest monitors (estimated) and the MOSkin dosimeter (with active bias applied via battery apparatus)
5.2 Methodology

The MOSkin dosimeter was the primary dosimeter used during measurements. Dose delivery was assessed at native depth $H_p(0.07)$ using calibration data reported on in Chapter 2. Dose delivery was verified through comparison to an established operator dosimeter, the TLD-100 dosimeter encapsulated within a Radcard Eye-D enclosure. The TLD-100 dosimeter was an ideal comparative measure of the MOSkin dosimeter performance when using diagnostic standard beam qualities. While OSL and RPL dosimeters have been popular in operator dosimetry studies, the traditional TLD-100 (LiF:Mg, Ti) is still the most commonly used dosimeter in the clinical environment due to how comprehensively established and understood the TLD-100 dosimeters have become over the last several decades. Measurements were performed at the SCK-CEN calibration facilities and readout of the TLD-100 was performed with assistance from SCK-CEN personnel.

5.2.1 Calibration of the TLD-100 Eye-D Dosimeter

Similarly to the MOSkin in Section 2.2.4, the Eye-D dosimeters were calibrated for energy dependence using the full range of Narrow Spectrum Series and RQR Series beam qualities available at the SCK-CEN calibration facilities. Angular dependence was assessed using the Narrow$_{80}$ and RQR$_6$ beam qualities. These qualities were chosen for their reported similarity to interventional beam qualities $^{[121]}$. Irradiation was performed using an Xstrahl dual x-ray tube system equipped with a 100 keV and a 300 keV x-ray tube using a variety of filtration settings and with a collimated Cesium-137 (Cs-137) point source at distance of 1 meter from the irradiation source. Before irradiation, air Kerma delivery to the measurement point in free air was measured using a 1 litre PTW ionization chamber for each beam quality used. After measuring the air Kerma in free air, the Eye-D was attached to the $20 \times 20$ cm cylindrical
phantom volume and irradiated for an appropriate exposure time. For the Narrow Spectrum Series beam qualities, this exposure time varied depending on dose rate for each specific beam quality due to time constraints. On average the Eye-D dosimeters were irradiated to an exposure of \( \sim 56 \text{ mGy in air Kerma} \). The RQR Series beam qualities provided much higher dose rates and, as such, the Eye-D dosimeters were irradiated to a threshold of 120 mGy for all RQR Series beam qualities. Angular response was measured with the Eye-D dosimeter positioned at a fixed distance from the irradiation source. During these measurements, the phantom was positioned on a rotatable turntable. The angle of the Eye-D dosimeter with respect to normal incidence of the x-ray beam was manipulated by rotating the turntable. For further description and schematics of this apparatus, please refer to Section 2.2.4. TLD Readout was performed by the SCK-CEN calibration facility with results provided as counts contributing to glow peak 5. Error was calculated using the standard deviation of the TLD Eye-D dosimeters irradiated for each beam quality. \( H_p(3) \) equivalent dose to the measurement point was calculated using the ISO 4037-3 methodology outlined in Section 2.3.4. Air Kerma conversion coefficients for the cylindrical phantom volume were primarily sourced from Behrans et.al. for Narrow Spectrum Series beam qualities and from Principi et.al. for RQR Series beam qualities\textsuperscript{[116, 117]}. Missing conversion coefficients were simulated using Monte Carlo methodology. The energy dependence datasets were presented normalised with respect to the response to the Cs-137 source so as to compare to similar published calibration studies.
5.2. Methodology

5.2.2 Measuring Eye Lens Dose from Several Recommended Reference Points

The clinicians’ head was simulated using a Rando-Alderson head phantom. Photographs of the phantom have been presented in Figure 5.2. The phantom was irradiated in three equipment configurations; without radioprotective glasses, wearing standard radioprotective glasses (Glasses A) and wearing side shielded radioprotective glasses (Glasses B). Photographs of the Rando-Alderson phantom and the two types of radiation protective glasses have been provided in Figure 5.2. The phantom was mounted on the rotatable turntable, as represented diagrammatically in Figure 5.3, and positioned with the right eye lens at a distance of 1 meter from the irradiation source. The phantom was then irradiated in each equipment configuration for an angular range of -90° to 90° degrees relative to the incident x-ray beam as incremented in 15° degree steps. For the purposes of these measurements, 90° was defined as when the phantom was facing right relative to the incident x-ray beam. All irradiations were performed using the higher dose rate RQR₆ beam quality rather than the Narrow₈₀ beam quality due to time constraints. Each irradiation delivered an air Kerma of 120 mGy in free air to the measurement point. \( H_P(0.07) \) and \( H_P(3) \) doses were calculated using the aforementioned ISO 4037-3 protocol.
5.2. Methodology

Figure 5.2: Photographs of a) Glasses A, b) Glasses B (with side shielding) and c) the Rando Alderson phantom equipped with Glasses type A

Figure 5.3: Diagrammatic representation of head phantom mounted to the turn table apparatus a) with angle measured marked and b) in the 45° configuration
During irradiation, the Eye-D dosimeter was used to measure $H_p(3)$ dose delivered to the left eye lens for all phantom equipment configurations and all angular configurations. Conversely, the MOSkin dosimeter was used to measure $H_p(0.07)$ dose delivered to five measurement points; the eye lens, under-glasses, over-glasses, forehead and side of glasses measurement points. These measurement points are identified in Figure 5.4. The MOSkin measurement points were monitored for both of the glasses configurations for all angular configurations specified previously. The eye lens dose measurement point dose values were compared to and verified using the Eye-D response. Each of the other MOSkin measurement points were compared directly to the eye lens dose MOSkin measurement point. Each measurement point was assessed using qualitative metrics that considered how representative the response at the measurement point was of the eye lens measurement point response.

### 5.2.3 Development of an $H_p(3)$ equivalent MOSkin enclosure

The MOSkin dosimeter measures dose at a native $H_p(0.07)$ measurement depth. As mentioned previously, the most appropriate depth for measuring eye lens dose is the 3 mm measurement depth. To more accurately measure eye lens dose delivery, an appropriate encapsulation solution was designed for the MOSkin dosimeter. This encapsulation cap was produced from solid water material. The capping geometry resembled a half dome/half oblong shape and was designed to provide a consistent 3 mm thickness of solid water around the sensitive volume of the MOSkin dosimeter. The design of the capping material is presented in Figure 5.5.

The capping material was tested at ARPANSA calibration facilities based in Yallambie Melbourne. The cylindrical phantom methodology used at SCK-CEN was reproduced as faithfully as possible. Narrow Spectrum Series beam qualities were limited to the Narrow$_{40}$ to 250 beam qualities due to limitations of the x-ray apparatus. A
singular $H_P(0.07) \text{ MOSkin}$ was irradiated in addition to the $H_P(3) \text{ MOSkins}$ to ensure that MOSkin response to the ARPANSA beam qualities was consistent with MOSkin response to the SCK-CEN beam qualities. Energy dependence and angular response of the $H_P(3) \text{ MOSkin}$ was compared to the performance of the native $H_P(0.07) \text{ MOSkin}$ results presented in Section 2.3.4.

Figure 5.4: a) Annotated diagram of MOSkin measurement points with b) photograph of phantom fitted with MOSkin dosimeters

Figure 5.5: $H_P(3) \text{ MOSkin}$ capping material represented schematically a) in isometric view, b) in isometric view (internal borders visible), c) from the side view (cross-section with MOSkin cavity visible), d) from the front view (MOSkin position marked, dimensions marked) and e) as positioned on the phantom volume during characterisation (photograph)
5.3 Results

5.3.1 TLD Eye-D calibration values

The TLD Eye-D, reference depth $\sim 3$ mm, response to the Narrow Spectrum Series and RQR Series beam qualities are presented in Figure 5.6a and Figure 5.6b respectively using red diamond points. The results were validated through comparison to simulation data published by Bilski et.al.$^{[165]}$. The Bilski et.al. study included simulated TLD Eye-D response values generated via a custom MCNP-$\chi$ code, represented by the dotted line and six measured data points, denoted by crosses, obtained through irradiation with Narrow Spectrum Series and RQR Series beam qualities. All data points were presented on a logarithmic scale of the mean beam energy and normalised to Cs-137 response. Peak response was observed when using the Narrow$_{30}$ (Mean energy = 24 keV) and RQR$_2$ (Mean energy = 27.54 keV) beam qualities. Angular response of the Eye-D dosimeter to the Narrow$_{80}$ and RQR$_6$ beam qualities has been presented in Figure 5.7. All values were normalised to the $0^\circ$ incidence response values. Eye-D response per unit of the calculated $H_P(3)$ doses, as calculated for each angle of incidence, increased with angulation and peaked at $90^\circ$ where response was 23.7% higher than the expected value for the Narrow$_{80}$ beam quality and 67.8% higher than expected for the RQR$_6$ beam quality.
Figure 5.6: Eye-D response to the a) Narrow Spectrum series and b) RQR series beam qualities available at the SCK-CEN calibration facilities as compared to data published by Bilski et.al.\textsuperscript{[165]}
Figure 5.7: Eye-D angular response to the Narrow$_{80}$ and RQR$_6$ beam qualities
5.3. Results

5.3.2 Comparison of Eye Lens Dose Measurement Points

Eye lens dose to the phantom was measured using the Eye-D dosimeter for all phantom configurations and was considered the gold standard for eye lens dosimetry in this experiment. The eye lens dose delivered to the phantom without radioprotective equipment equipped has been presented in Figure 5.8. Eye-D response peaked when irradiating the -30° configuration where doses were 20.9% higher than observed at normal incidence. Eye lens dose to the phantom configurations when equipped with radioprotective glasses as measured by the MO Skin and by the Eye-D has been presented in Figure 5.9. For the angular range of -60° to 60°, the MO Skin dosimeter measured doses that were on average 3% lower when using Glasses A and 12% lower when using Glasses B. For the angular range of -60° to 60°, eye lens dose was on average ∼90% lower when wearing the radioprotective glasses. The efficacy of Glasses A as compared to Glasses B has been presented in Figure 5.9c. Glasses B provided slightly more radiation protection than Glasses A did. The MO Skin response at all measurement points for the two phantom equipment configurations have been presented in Figure 5.10 and Figure 5.11. Each MO Skin measurement point was compared individually to the eye lens measurement point using a dual-Y axis graph format to better visualise how closely each measurement point represented the eye lens dose delivery. The ratio of dose delivered to the measurement point with respect to eye lens dose was calculated and averaged over an angular range of -60° to 60°. The exception to this general rule was the side lens measurement point in which the 0° to 60° range of measurement points were not included in the calculation as dosimeter response trended towards negligible for these angles. The formula for this ‘Ratio of Response’ variable has been presented as Equation 5.1.

\[
\text{Ratio of Response} = \frac{\sum_{-60^\circ}^{60^\circ} R_{MP}}{\sum_{-60^\circ}^{60^\circ} R_{MP}}
\]  

Equation 5.1
Figure 5.8: Phantom eye lens dose without any personal protective equipment (normalised to 0° angle of incidence)

Figure 5.9: $H_p(0.07)$ and $H_p(3)$ doses measured at the eye lens reference point with the Rando-Alderson phantom equipped with both a) Glasses A and b) Glasses B with c) comparative graph included.
5.3. Results

**Glasses A**

Figure 5.10: Comparison of MOSkin response for a) all measurement points and for eye lens measurement point compared to b) the under-glasses measurement point c) the over-glasses measurement point d) the side-glasses measurement point and the e) forehead measurement point for Glasses configuration A

**Glasses B**

Figure 5.11: Comparison of MOSkin response for a) all measurement points and for eye lens measurement point compared to b) the under-glasses measurement point c) the over-glasses measurement point d) the side-glasses measurement point and the e) forehead measurement point for Glasses configuration B
5.3.3 H_<sub>P</sub>(3) MOSkin Calibrations

The H_<sub>P</sub>(3) MOSkin encapsulation energy dependence has been presented in Figure 5.12. H_<sub>P</sub>(3) MOSkin response was observed to peak for the RQR<sub>4</sub> beam quality where the sensitivity of the dosimeter was 1.25 mV/mSv. Angular response of the H_<sub>P</sub>(3) MOSkin with respect to the θ rotational axis has been presented in Figure 5.12a and angular response of the H_<sub>P</sub>(3) MOSkin with respect to the Φ rotational axis has been presented in Figure 5.12b. The MOSkin response was within 5% of expected values for incident angles of up to 60°.

![Figure 5.12: Angular response of H_<sub>P</sub>(3) encapsulated MOSkin to Narrow<sub>80</sub> and RQR<sub>6</sub> beam qualities with respect to the a) θ rotational axis and the b) Φ rotational axis.](image)
5.4 Discussion

5.4.1 The Accuracy of the TLD Eye-D Calibration

The Eye-D dosimeters performed comparably to data published by Bilski et.al.\textsuperscript{[165]}. The peak response of the TLD- Eye-D dosimeters occurred somewhere between the Narrow\textsubscript{30} and RQR\textsubscript{2} beam qualities, that is, within a mean energy range of 24 - 27.5 keV. This peak in response occurred at lower energies than the Bilski et.al. simulated energy response which peaked when using the Narrow\textsubscript{40} beam quality (33 keV). This was a result of each study using different composition TLD dosimeters. The calibration was performed using the more traditional LiF:Mg, Ti type TLD while Bilski et.al. used newer high sensitivity LiF:Mg,Cu,P type TLDs. The difference between these two types of TLD is well understood and the lower energy response peak is consistent with results published previously by Bilski et.al.\textsuperscript{[172]}. The angular response of the Eye-D dosimeter was observed to increase with increasing angulation from normal incidence. TLD response remained within ±15% of response at normal incidence for angles up to the 75° configuration for the Narrow\textsubscript{80} beam quality and for angles up to the 60° configuration for the RQR\textsubscript{6} beam quality which was consistent with results published by Bilski et.al. for the RQR\textsubscript{7} beam quality\textsuperscript{[165]}.

5.4.2 Operator Phantom Eye Lens Doses

The eye lens dose delivered to the Rando-Alderson head phantom peaked when irradiating the -30° incidence configuration. This angle maximises the profile of the Eye-D dosimeter to the incident beam and minimises obstructions between the incident beam and the dosimeter. Equipping the phantom with radioprotective glasses reduced dose delivered to the ±60° configurations significantly by an average of 91.2%. When irradiating the phantom in angular configurations beyond ±60° range, the radioprotective
glasses only provided partial protection to the eye lens measurement point. When
irradiating the phantom in the Glasses A configuration from the 75° angular configu-
ration, the dosimeters experienced a significant spike in response which represented an
increase in dose be a factor of 4.65, or alternatively, 52.1% of the no-glasses phantom
configuration value. The Glasses B configuration also recorded a spike in eye lens dose
at the -90° angle. These increases in dose delivery demonstrate radiation incident from
broad angles can potentially bypass the coverage of the radiation protective glasses
and irradiate the eye lens directly. Radioprotective glasses do not provide protection
for all angles and the results have demonstrated that gaps in protection differ based on
the geometry of glasses. The results emphasise the need for eye lens dose monitoring
to ensure compliance with operator dose limits.

Of the two styles of glasses reviewed in this study, the Glasses B, the side shielded
glasses, were observed to provide a more homogenous protective coverage than Glasses
A. The MOSkin and Eye-D dosimeters produced very similar dosimetric responses.
The results suggested that the MOSkin dosimeter was able to accurately represent the
eye lens dose and that the $H_P(0.07)$ and $H_P(3)$ measurements depths were comparable
measures of eye lens doses for interventional range beam qualities which was consistent
with the guidance provided by ICRP Publication 139\textsuperscript{61}.

When assessing the utility of different eye lens dose measurement points, each
measurement point exhibited different benefits and limitations. The most difficult
challenge in monitoring the eye lens dose effectively was in finding a measurement
point that provided representative response of the clinician head and glasses geomet-
tries. The highest doses were measured at the over-lens and forehead measurement
points. As such, these measurement points would enable use of less sensitive dosime-
ters to measure operator eye lens dose. These measurement points would require a
scaling factor to properly estimate eye lens dose. In this study, that scaling factor was
5.4. Discussion

a factor of roughly 10-12 times the phantom eye lens dose. Conversely, the under-
glasses measurement point accounted for the attenuation of incident radiation by the
radioprotective glasses. Due to how close this measurement point was to the radioprotective glasses, this measurement point was not able to detect variance in eye lens
dose resulting from the geometry of the glasses used and, as such, this measurement
point may potentially underestimate eye lens dose. The side lens measurement point
was a good compromise in measurement points as this measurement point received the
higher dose from being on the outside of the glasses and was closely representative of
the eye lens dose for angles for -90° - 0°. Unfortunately, the side glasses was completely
incapable of accurately measuring eye lens dose when exposed from angles in the 0
- 90° range. For Glasses A, the side glasses measurement point observed decreasing
response with increasing angulation from normal while for Glasses B, the side glasses
measurement point observed negligible response for all angulations >0°. This result
suggests that passive dosimeters positioned on the side of an operator’s glasses may
underestimate eye lens dose depending on type of glasses and on the directions an
operator faces during the procedure. The impact of this effect on accuracy could be
reduced through comparison to a second complimentary measurement point, however,
this would require a more sophisticated measurement. The results indicate that there
is no clearly superior measurement point and the most appropriate measurement point
depends on variables such as operator behaviour, procedure undertaken, type of glasses
worn and dosimetry solution implemented.

5.4.3 Performance of the $H_{P}(3)$ MOSkin enclosure

The $H_{P}(3)$ MOSkin dosimeter performed comparably to the native $H_{P}(0.07)$ mea-
surement depth MOSkin. Comparison of the $H_{P}(3)$ and $H_{P}(0.07)$ MOSkin energy
dependence has been provided in Figure 5.13. The primary differences observed
Figure 5.13: Comparison of energy dependence between the native H\textsubscript{P}(0.07) MOSkin dosimeter and the encapsulated H\textsubscript{P}(3) MOSkin dosimeter were in the energy response of the H\textsubscript{P}(3) MOSkin. Peak energy response was observed for the RQR\textsubscript{4} beam quality which represented a shift in peak response to higher beam energies (\(\Delta E = 12\) keV). The H\textsubscript{P}(3) MOSkin also observed reduced sensitivity at peak of \(~9\%\). Low energy dependence of the H\textsubscript{P}(3) MOSkin could not be assessed as low energy beam qualities were unavailable at the ARPANSA calibration facilities. The H\textsubscript{P}(3) MOSkin did provide greater angular homogeneity than the H\textsubscript{P}(0.07) MOSkin. Asymmetry was observed when rotating with respect to the \(\Phi\) axis due to radiation entering the dosimeter MOSkin cavity directly and therefore bypassing the capping material. This should be taken into consideration when positioning the H\textsubscript{P}(3) MOSkin.
5.4.4 Consideration for Future Operator Dosimetry MOSkin Design Iterations

5.5 Conclusions

The results of this operator dosimetry study provided insights to the requirements of an operator specific dosimeter. Investigation regarding eye lens measurement points observed unique benefits and drawbacks to use of each measurement point. Glasses geometry had a significant effect eye lens dose and should be considered when choosing an eye lens dose measurement point. $H_p(3)$ capping reduced angular dependence of the MOSkin dosimeter at cost of $\sim 9\%$ sensitivity. Peak response shifted to higher beam energies which may not be beneficial for operator specific applications but would benefit neuro-interventional or CT applications. The sensitivity of the MOSkin dosimeter remains the primary challenge to operator dosimetry specific implementations.
Chapter 6

Floating Gate MOSFET Dosimeters: A Prospective Study

The final measurement focused chapter of this research thesis will focus on the future of diagnostic dosimetry solutions. This chapter will focus on characterisation of a novel prospective dosimetry solution, the Floating Gate MOSFET dosimeter. This chapter will assess stability, dose response and reproducibility of the new dosimeter and will showcase detector stability, dose response to a range of sources and identify potential applications going forward from this research thesis.

6.1 Overview of Floating Gate MOSFET devices

A Floating Gate MOSFET, referred to hereafter as FG-MOSFET, is a MOSFET type device differentiated from conventional MOSFET designs by a ‘floating gate’ isolated within the oxide material. A diagram of the FG-MOSFET structure has been provided in Figure 6.1. Under normal operation, charges cannot conduct between the floating gate and other terminals, however, there are methods for manipulating charge within the floating gate structure. Depending on context, adjusting charge within the floating
gate intentionally is referred to as a ‘write’ operation while readout is referred to as a ‘read’ operation. By accumulating charge carriers within the floating gate, the electric field generated by floating gate will change, affecting the conductivity of the channel between the source and drain terminals. Charge can be manipulated using the following methodologies:

- **Fowler-Nordheim tunnelling**
  A quantum mechanical effect that occurs when valence electrons near a sufficiently thin insulative barrier are subjected to a sufficiently strong electric field. These conditions enable charge carriers to tunnel through the insulator in the direction dictated by the electric field. In FG-MOSFET devices, this interaction is used to accumulate or remove charge from the floating gate via the tunnelling oxide. Excess charges are supplied and removed via the source and drain terminals [173].

- **Hot-carrier injection**
  In hot-carrier injections, charge carriers are accumulated within the floating gate by attracting excited charge carriers toward the control gate. If the charge carriers are sufficiently energised, there is a chance of the charge carrier tunnelling through the tunnelling oxide [174].

- **Annealing**
  FG-MOSFETs can be annealed through applying temperature or, more commonly, with ultraviolet light. This process imparts energy to charge carriers isolated within the floating gate structure. If these charge carriers exceed an energy threshold the annealing process will enable conduction through the tunnelling oxide [175].
6.1. Overview of Floating Gate MOSFET devices

Figure 6.1: a) Diagram of the FG-MOSFET structure and visual representation of the Fowler-Nordheim tunnelling effect used to b) deposit and c) remove charge from the floating gate structure.

Typically a write operation will cause degradation to the gate oxide as such there is a limit to the number of write operations possible in the lifetime of the device. Read operations do not cause significant damage and can be performed ad infinitum. Fowler-Nordheim tunnelling is the least damaging methodology to adjust floating gate charges followed by UV annealing and as such these are more commonly used in dosimetric applications. For further reference, the roles of Fowler-Nordheim tunnelling, annealing and hot-carrier injection in FG-MOSFET dosimetry applications are discussed comprehensively in publications by Peters et. al., Martin et.al. and McNulty et.al. respectively.\cite{173 - 175}

An FG-MOSFET based dosimeter functions similarly to a standard MOSFET dosimetry solution; charge accumulates within the gate oxide due to ionisation events from incident radiation, the charges are trapped within the gate oxide and contribute to the conduction characteristics of the device. FG-MOSFETs enable two distinct advantages over traditional MOSFET devices in that an FG-MOSFET is able to accumulate charges within the isolated floating gate structure and that the charge on
the floating gate structure can be manipulated using the above methods. This means a FG-MOSFET is resettable and reusable to a greater and much more accessible extent than traditional MOSFET dosimetry solutions. The FG-MOSFET, by design, is a highly stable and customisable device. In the absence of stimulation, the isolated floating gate retains net charge indefinitely. This functional stability has resulted in widespread use of FG-MOSFETs in EPROM/EEPROM flash memory applications. Through adjusting the design of the gate geometry, fabrication process, size and biasing voltage, the FG-MOSFET sensitivity, linearity and noise characteristics can be customised for specific applications.

6.2 FG-MOSFET Characterisation methodologies

This chapter will focus on characterising several prospective Application-specific integrated circuit (ASIC) designs each featuring unique FG-MOSFET active sensitive volumes. A total of 12 devices underwent the characterisation and these ASICS have been designated in the format ASIC###. The specific of each design are beyond the scope of this research thesis. The characterisation was performed to assess the following performance metrics:

1. Temperature dependence of FG-MOSFET readout

2. FG-MOSFET Output Stability at thermostable ambient temperature

3. FG-MOSFET response to radiotherapeutic $\gamma$ irradiation

4. FG-MOSFET response to diagnostic x-ray irradiation

5. FG-MOSFET reproducibility and desensitisation effects

During irradiation, the dosimeters were socketed into a National Instruments daughterboard. Threshold voltage and readout was performed using proprietary software
developed in Lab View (https://www.ni.com/en-au/shop/labview.html). An alternative configuration featuring wireless readout was also available, however, the daughterboard configuration was intentionally used for all tests to provide best results. Ambient temperature was monitored at all times via four on-daughterboard temperature sensors. Before and after each experiment the FG-MOSFETs were pre-charged to produce a threshold voltage of 1.300 V.

6.2.1 FG-MOSFET Characterisation: Output Stability and Temperature Dependence

The output stability and temperature dependence of the FG-MOSFET devices were assessed over long acquisition periods in the absence of ionising stimulus. The measurement was performed over a 24-hour period. At commencement of the monitoring period, threshold voltage was initialised at 1.300 V. During the monitoring period, threshold voltage was sampled once every two seconds. Ambient temperature was recorded using the on-board temperature sensors which was used to establish the influence of temperature on dosimeter readout. Readout stability was assessed over a 1-hour monitoring period at a thermostable ambient temperature. Readout stability was by assessed by measuring the variance of the readout from average readout value.
6.2.2 FG-MOSFET Characterisation: Mega-Voltage range

Gamma Characterisation

The FG-MOSFETs were irradiated using the clinical 6 MV Varian Truebeam Linear Accelerator commissioned at the Illawarra Cancer Care Centre used in Chapter 2. Ten of the FG-MOSFETs were positioned on the patient couch within a Solidwater slab phantom. This phantom was composed of several $30 \times 30 \text{cm}^2$ Solidwater plates arranged to provide a 10 cm thickness build-up layer and a 20 cm thickness backscatter layer. A $20 \times 20 \text{cm}^2$ irradiation field was selected to ensure all dosimeters were within the irradiation field. The phantom configuration has been represented graphically in Figure 6.2a.

The FG-MOSFETs were irradiated over the course of four 35-minute cycles. These cycles consisted of three 10-minute irradiation periods with five minutes of rest allocated before and after each irradiation. Each irradiation delivered a dose of 10 Gy to the FG-MOSFETs at a rate of 2 Gy per minute. At the end of each cycle, the FG-MOSFETs had been exposed to a total dose of 30 Gy. This irradiation cycle was performed four times and as such a total cumulative dose of 120 Gy was delivered to the FG-MOSFETs. At the start of each cycle, the FG-MOSFETs were reinitialised at a threshold voltage of 1.300 volts. Readout operations were performed every second continuously throughout each cycle. Output stability was assessed before and after each irradiation to assess stability and identify potential dose fading effects. Sensitivity of the FG-MOSFETs was assessed by averaging readout values within 6-second interval blocks. Sensitivity was monitored for any potential desensitisation caused by oxide degradation or charge trap saturation effects. The results were compared directly to the MOSkin sensitivity results reported in Chapter 2 from a similar characterisation methodology.
6.2. FG-MOSFET Characterisation methodologies

Figure 6.2: Irradiation apparatus for irradiating the FG-MOSFET using a) a 6 MV linear accelerator and b) an angiographic c-arm system (dose verified with MOSkin)

### 6.2.3 FG-MOSFET Characterisation: Kilovoltage Range

#### X-ray Characterisation

The FG-MOSFETs were irradiated using a Philips Allura Clarity angiographic c-arm system commissioned at Southern Heart Clinic. Ten FG-MOSFETs were positioned on a daughterboard fixed centrally beneath a $30 \times 30 \times 30$ cm$^3$ Perspex slab phantom. The c-arm x-ray tube and couch were positioned to align the center of the phantom to the c-arm system’s rotational isocenter. X-ray tube parameters including x-ray tube voltage, x-ray tube current and fluoroscopic pulse width were selected automatically by the c-arm system and, as such, the phantom EPT was adjusted throughout the measurements to produce a range of irradiation conditions. Phantom EPT was reduced in 2 cm increments for the 30 to 18 cm EPT range and was reduced in 4 cm increments for the 18 10 cm range. The irradiations were repeated using three different imaging protocols, ‘Low’, ‘Normal’ and ‘Boost’. These protocols affected the x-ray tube voltage, x-ray tube current, pulse width and filtration selected the c-arm x-ray tube. The most
consistent impact of changing protocol was the beam filtration utilised: ‘Low’ protocol using 5 mmAl filtration, the ‘Normal’ protocol using 2 mmAl filtration and the ‘Boost’ protocol using 0 mmAl filtration. The c-arm system was positioned at the posterior-anterior angle for all phantom configurations. An FOV of 25 cm DFS was selected to ensure all FG-MOSFETs were irradiated uniformly. All irradiations were performed in cine-acquisition mode at an imaging frame rate of 15 FPS. The FG-MOSFETs were irradiated for a minimum of 30 seconds per phantom configuration. Irradiation was continued after the 30 second period for low dose exposures until a minimum air Kerma at reference point (as estimated by the c-arm system) of 50 mGy was achieved. This ensured better statistics for the dosimeter. Dose delivered to the FG-MOSFET devices was verified using the MOSkin dosimeter using calibration data reported in Section 2.3.4. Irradiation time was calculated by dividing the number of cine-acquisition frames by the imaging frame rate. The MOSkin was positioned centrally to the FG-MOSFET daughter board during the irradiations.

6.3 Results of the FG-MOSFET Characterisation

6.3.1 Output Stability and Temperature Dependence

The FG-MOSFET readout and ambient temperature over the 24-hour monitoring period has been presented graphically in Figure 6.3. In this figure, the FG-MOSFET outputs have been each been offset by 5 mV to clearly represent the variation in each device. A relationship between ambient temperature and FG-MOSFET readout was observed through comparing the maximum and minimum values for each quantity. The 1-hour stability monitoring period has been presented in Figure 6.4. During this 1-hour period, temperature was measured constant at 22.65°C. The 95% confidence interval has been denoted using dashed lines.
6.3. Results of the FG-MOSFET Characterisation

Figure 6.3: a) FG-MOSFET readout stability over 24-hour period and b) the ambient temperature as measured by onboard resistors.

Figure 6.4: FG-MOSFET readout stability over 1-hour period at constant temperature (T=22.7°C)
6.3.2 Megavoltage Range Gamma-ray Sensitivity

The increase in FG-MOSFET threshold voltage with increasing dose deposition to the device from the 6MV linear accelerator has been presented graphically in Figure 6.5. It should be noted that the FG-MOSFETs were reinitialised to a threshold voltage of 1.3 V on finalisation of each 30 Gy cycle, a milestone marked by a partitioning line in figure. The FG-MOSFET sensitivity from the real-time data is presented graphically in Figure 6.6. FG-MOSFET sensitivity was also calculated in terms of each 10 Gy irradiation step. This interpretation of the results is presented in Table 4.1. Similar to the MOSkin, a linear trend was observed between desensitisation and FG-MOSFET threshold voltage. This linear trend has been presented graphically in Figure 6.7 using data from irradiation cycles 3 and 4. FG-MOSFET desensitisation with increasing charge accumulation was calculated to be 22.99 mV per volt of increasing threshold voltage. Regression analysis of this trend yielded an R-squared value of 0.81.

Figure 6.5: Accumulated FG-MOSFET response to 6MV linear accelerator beam quality (Threshold Voltage reset at 30, 60 and 90 Gy)
6.3. Results of the FG-MOSFET Characterisation

Figure 6.6: FG-MOSFET sensitivity to 6MV linear accelerator beam quality with respect to accumulated dose delivered

Figure 6.7: FG-MOSFET sensitivity to 6MV linear accelerator beam quality with respect to threshold voltage before irradiation
### Results of the FG-MOSFET Characterisation

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All FG-MOSFET Sensitivities Presented in mV/Gy

**Table 6.1:** FG-MOSFET Sensitivities for each 10 Gy Irradiation Period
6.3.3 Kilovoltage Range X-ray Sensitivity

The FG-MOSFET response to diagnostic range x-ray beam qualities has been presented in Figure 6.8. The FG-MOSFET sensitivity was calculated by measuring the change in FG-MOSFET threshold voltage over each minute-long irradiation and dividing by dose delivered as measured by the MOSkin dosimeter. The FG-MOSFET has been presented with respect to peak beam energy as reported by the c-arm system for each phantom configuration.

![Figure 6.8: FG-MOSFET energy dependence with respect to clinical diagnostic x-ray beam qualities](image)
6.4 Discussion of FG-MOSFET Performance

The FG-MOSFET performance was acceptable for dosimetric applications. The results produced during this prospective study cover three distinct discussion points: Output stability, radiation sensitivity and lifetime.

6.4.1 FG-MOSFET Output Stability

The FG-MOSFET was observed to retain threshold value over the 1-hr thermostable monitoring period. The maximum deviation from average was observed at 630 µV above the average value and a 95% confidence intervals was established at ±310 µV from average value. Over the extended monitoring period the FG-MOSFET did vary more significantly. Threshold voltage varied in a range of ∼12 - 25 mV over the 24-hour monitoring period. This corresponded closely with changes in the ambient temperature, as shown in Figure 6.3b. From this data, a temperature shift in threshold voltage of ∼2 - 4 mV/°C was established. As this correlation was consistent throughout the 24-hour monitoring period, this data may in future be used to develop a temperature correction method to compensate for this effect.

6.4.2 FG-MOSFET Sensitivity

On irradiation the following average FG-MOSFET sensitivities were observed:

- γ-ray (E_6 MV): 73 mV/Gy.
- X-ray (E_{65-100 kVp}): 300 - 600 mV/Gy.

The FG-MOSFET response during kilovoltage x-ray irradiation resulted in minimal energy dependence observed in the 70-100 kVp range of filtered beam qualities. Low energy photons appeared to produce higher FG-MOSFET responses and as such low
energy photon sensitivity should definitely be explored further, potentially with the Narrow Spectrum Series and RQR beam qualities used to characterise the MOSkin dosimeter in Chapter 2

The most comprehensive display of the FG-MOSFET capabilities was the gamma. Initial sensitivity @ $V_{Th} = 1.300$ V in Irradiation Cycle 1 was observed to be $75.1 \text{ mV/Gy}$ as averaged over all devices over the initial 10 Gy irradiation period. Sensitivity compared well to the lower limit literature reported sensitivity values. Peters et.al. observed their FG-MOSFET designs were capable of sensitivities between 80-280 mV/Gy $^{173}$. Peters et.al. also suggested that sensitivity in the higher region of this range was constrained by collection efficiency of FG-MOSFET devices. Live readout of the FG-MOSFET was possible and accumulative dose delivery is presented in Figure 6.7.

6.4.3 FG-MOSFET Desensitisation

The following observations regarding the FG-MOSFET mega-voltage gamma-ray sensitivity were noted:

- FG-MOSFET sensitivity decreased with increasing cumulative dose delivered to the devices. Over each 30 Gy irradiation cycle this drop in sensitivity averaged to $\sim 36.6\%$ of the response recorded @ $V_{Th} = 1.300$ V.

- FG-MOSFET sensitivity was mostly restored on reinitialising the FG-MOSFET to $V_{Th} = 1.300$ V. FG-MOSFET sensitivity did decrease during each subsequent irradiation cycle with Irradiation Cycle 2, 3 and 4 observing initial sensitivities 5.6%, 10.5% and 10.6% lower than Irradiation Cycle 1 initial sensitivities respectively.

- There was significantly less impact on sensitivity when reinitialising the FG-
MOSFETs for Irradiation Cycle 4, suggesting that this desensitisation effect could reach saturation and as such a pre-irradiation of $\sim$100 Gy may improve reproducibility of the FG-MOSFET on subsequent reinitialisation.

- FG-MOSFET sensitivity was observed to reduce linearly with per volt increase in threshold voltage at a rate of 22.99 mV/V. This conclusion was made from data collected during Irradiation Cycles 3 and 4 to minimise the impact of desensitisation effects.

- Readout stability was greater in the threshold voltage range of 1.500 V to 2.500 V.

The results suggest that desensitisation could potentially be minimised by performing pre-irradiation to 100 Gy cumulative dose, that dose reporting accuracy could be increased by implementing sensitivity correction based on threshold voltage and reinitialising the FG-MOSFET after exposure to minimise variations in sensitivity.

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**Table 6.2:** Comparison of FG-MOSFET characteristics to MOSkin Dosimeter
6.5 FG-MOSFET Conclusions

The FG-MOSFETs were observed to effectively measure exposures produced from both megavoltage therapeutic range and kV diagnostic range x-ray sources. The FG-MOSFETs were observed to produce stable threshold voltage responses to within 400 \( \mu V \) of the average value as measured over an hour period at static ambient temperature conditions. The FG-MOSFET threshold voltages were observed to be affected by temperature at a rate of 2 - 4 mV/°C depending on chip design. As such it is advised to operate the chip in static ambient temperature conditions and at a time after initialising readout to ensure that variations in response due to temperature are avoided. Future readout software could incorporate readout from onboard temperature monitors similar to the ones used during these experiments to compensate for temperature effects. Some ASICs produced anomalous responses during measurements. These anomalies were considered outside the scope of this report and can be discussed in a separate meeting but may indicate refinement of the chip design or production is necessary going forward.

The FG-MOSFETs exhibited reduced observed lifetime (30 Gy vs 50 Gy), reduced radiation sensitivity (\(~25\% - 35\%)\) and increased rates of desensitisation (\(~1\%/Gy vs 0.33\%/Gy\)) when compared to the MOSkin dosimeter, however, the FG-MOSFET can be reinitialised to the initial threshold voltage multiple times. This expands the lifetime of the FG-MOSFET significantly enabling the FG-MOSFET to easily eclipse the MOSkin lifetime. On reinitialisation there was some initial desensitisation observed, however, on irradiating to a threshold of 100 Gy, desensitisation on reinitialisation was insignificant and FG-MOSFET results were reproducible to within 5% on subsequent reinitialisation. Further testing is required to confirm this effect is permanent. Desensitisation of the FG-MOSFET was significantly (3x) higher than on standard MOSkin devices, however, this desensitisation was observed to correlate linearly with threshold
voltage at time of readout and is therefore predictable and can be compensated for in readout software. For high dose applications, such as multifractionate radiotherapy regimes, it is suggested that the FG-MOSFET is reinitialised regularly for maximum consistency and efficacy.

Kilovoltage range calibration of the FG-MOSFET indicate that the FG-MOSFET can measure 4 mGy per mV of threshold voltage. This low sensitivity may mean the current FG-MOSFET is unsuitable for low and very low dose diagnostic applications. The device may however be suitable for high dose diagnostic applications such as CT and angiography. The FG-MOSFET was observed to exhibit energy dependence at x-ray tube voltages lower than $\sim 80$ kVp. Sensitivity of the FG-MOSFET was observed to increase at these low voltage tube settings at a steeper rate than the traditional MOSkin dosimeters. The number of measurements/phantom and x-ray tube configurations utilised enabled a preliminary characterisation of the FG-MOSFET devices. This data could be used in future modelling of FG-MOSFET behaviour and could be used to compared to standardised beam qualities similar to how diagnostic data was utilised in Chapter 2.
Chapter 7

Conclusions and the Future of Diagnostic Dosimetry Solutions

In 2017, Rivera and Uruchurtu identified radiation monitoring for patients in interventional cardiology procedures as a requirement for the field going forward, stating that "radiation exposure is a significant concern for interventional cardiologists and patients due to the increasing workloads and the complexity of procedures"\cite{176}. Until now, all dosimetry solutions have had limitations. Passive dosimetry solutions do not provide real-time feedback and as such cannot identify or prevent accidental exposures. Active personal dosimeters have poor response to low energy x-ray spectrum which prevents the APD from being an effective quantitative exposure tool\cite{80, 121}. Dose contouring software solutions currently estimate dose to a reference point rather than report on a measured dose value which can result in significant inaccuracies \cite{149}. Traditional MOSFET dosimeters are visible in radiographs, are temperature dependent and often involve a non-reproducible packaging solution\cite{82}. The MO.Skin dosimeter does not experience these functional limitations and as such is the ideal interventional patient dosimetry solution.

Throughout this thesis, MO.Skin based technologies have proven to be the ideal
solution to diagnostic dosimetry requirements and can be used either as. By design, the MOSkin measures in real-time at a native 0.07 mm depth without being acquired in images or producing significant perturbation of the x-ray field. This thesis has comprehensively characterised the device for the kilovoltage range in Chapter 2, has been clinically implemented as an aid for developing dose minimisation strategies in Chapter 3 and has been used to effectively compare c-arm systems in Chapter 4. During the Chapter 5 operator dose study it was observed that the sensitivity of the MOSkin dosimeter was too low to effectively measure operator dose in real-time, however, the requirements of an operator specific dosimetry solution were identified and these requirements can be used during future development of prospective operator dosimetry solutions such as the FG-MOSFET technology that was prospectively assessed in Chapter 6.

The true value of the MOSkin dosimeter in the catheterisation laboratory was realised through the clinical applications of the dosimeter firstly as a tool to produce dose minimisation strategies and secondly to compare imaging equipment between catheterisation laboratories. In the context of dose minimisation, the MOSkin was used to identify the impact of adjusting imaging and irradiation configurations in Chapter 3. It was observed that these factors significantly affect ESD and DAP delivery, that the distinction between these parameters is not always clear or understood both clinically and within the literature and that the estimated air Kerma and DAP values may not even accurately represent the exposure the patient experiences. Future work could easily result in an array type MOSkin dosimetry system that simultaneously measures instantaneous entrance skin doses and dose area product delivery during clinical cases or further phantom studies. Furthermore, such a system could be used complementarily with dose contouring software packages to improve the accuracy of these systems. Alternatively, in Chapter 4, the MOSkin was used in conjunction with
existing protocols to assess c-arm system performance. The methodology requires further refinement to be deployed on a larger scale, however, the potential for this method to aid commercial decisions and spur research into improving the irradiation apparatus could revolutionise the industry.

In conclusion, this research thesis has provided a foundation for using the MOSkin dosimeter as a diagnostic dosimetry tool. This technology has great potential within clinical catheterisation laboratories. Future research will broaden that potential with new applications, further solidifying the necessity of using this dosimeter in clinical catheterisation laboratories.
Appendix A

Philips ‘Clarity’ Upgrade Evaluation
A.0.1 Overview of the Clarity Upgrade: Acquisition Mode

The Philips c-arm system assessment results for acquisition mode have been presented in both graphical and tabular formats. All acquisition mode results have been collated into the graphic seen in Figure A.1. In this graph, the ESD delivery to the phantom has been normalised to the ‘Cardiac Low’ ESD delivery value. In subsequent sections, comparable pre-upgrade and post-upgrade protocols have been paired and compared directly. Figure A.2 to Figure A.4 and Table A.1 to Table A.3 present these individual pairings using the ‘Score Ratio’ method described in Equation 4.2. Tables are labelled in the format ‘Protocol B’ : ‘Protocol A’.

Figure A.1: ESD delivery and image quality scores produced by the Philips Allura Xper and Philips Allura Clarity c-arm systems in acquisition mode using the a) 30 cm EPT, 15 cm FOV DFS configuration, b) 30 cm EPT, 20 cm DFS configuration c) 20 cm EPT, 15 cm DFS configuration and the d) 20 cm EPT, 20 cm DFS configuration
A.0.2 Comparison of the high detail acquisition mode protocols: ‘Clarity Boost’ and ‘Cardiac 4’

The ratio of ESD delivery and image quality scores recorded for the ‘Clarity Boost’ and ‘Cardiac 4’ acquisition mode protocols have been presented graphically in Figure A.2 and tabularly in Table A.1. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Clarity Boost’ protocol by an average of 19.3%.

- ICR scores were higher when using the ‘Clarity Boost’ protocol by an average of 10.3% when imaging the 20 cm EPT configurations and higher when using the ‘Clarity Boost’ protocol by an average of 57.4% when imaging the 30 cm EPT configurations.

- Spatial resolution scores were higher when using the ‘Clarity Boost’ protocol by an average of 25.9%.

- Total WTR scores were lower when using the ‘Clarity Boost’ protocol by an average 24.3% with exception to the 30 cm EPT, 20 cm DFS configuration in which the ‘Clarity Boost’ scored higher than the ‘Cardiac 4’ protocol by 4.2%.

- Static MTR scores were lower when using the ‘Clarity Boost’ protocol by an average 3.4% with exception to the 20 cm EPT, 15 cm DFS configuration where both protocols scored comparably.

- Dynamic MTR scores were lower when using the ‘Clarity Boost’ protocol by an average 4.1% with exception to the 20 cm EPT, 20 cm DFS configuration where both protocols scored comparably.
Figure A.2: Ratios of the dose delivery and image quality scores recorded when using the ‘Cardiac 4’ and ‘Clarity Boost’ acquisition mode protocols for all phantom and imaging configurations

Table A.1: Ratios of the dose delivery and image quality scores recorded when using the ‘Cardiac 4’ and ‘Clarity Boost’ acquisition mode protocols for all phantom and imaging configurations
A.0.3 Comparison of the high detail acquisition mode
protocols: ‘Clarity Normal’ and ‘Cardiac 4’

The ratio of ESD delivery and image quality scores recorded for the ‘Clarity Normal’ and ‘Cardiac 4’ acquisition mode protocols have been presented graphically in Figure A.3 and tabularly in Table A.2. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Clarity Normal’ protocol by an average of 54.3%.

- ICR scores were higher when using the ‘Clarity Normal’ protocol by an average of 22.3% with exception to the 15 cm EPT, 15 cm DFS configuration in which the ‘Clarity Normal’ protocol scored lower than the ‘Cardiac 4’ protocol score by 1.7%.

- Spatial resolution scores were higher when using the ‘Clarity Normal’ protocol by an average of 23.7%.

- Total WTR scores were lower when using the ‘Clarity Normal’ protocol to image the 20 cm EPT configurations by an average of 40.9% and were lower when using the ‘Clarity Normal’ protocol to image the 30 cm EPT configurations by an average of 3.5%.

- Static MTR scores were comparable between protocols with exception to the 30 cm EPT, 15 cm DFS configurations in which the ‘Clarity Normal’ protocol scored lower than the ‘Clarity Normal’ protocol by 2.6%.

- Dynamic MTR scores were lower when using the ‘Clarity Normal’ protocol by an average of 5.3%.
**Figure A.3:** Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity Normal’ and ‘Cardiac 4’ acquisition mode protocols for all phantom and imaging configurations.

**Table A.2:** Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity Normal’ and ‘Cardiac 4’ acquisition mode protocols for all phantom and imaging configurations.
A.0.4 Comparison of low dose acquisition mode protocols: ‘Cardiac Low’ and ‘Clarity Low’

The ratio of ESD delivery and image quality scores recorded for the ‘Clarity Low’ and ‘Cardiac Low’ acquisition mode protocols have been presented graphically in Figure A.4 and tabularly in Table A.3. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Clarity Low’ protocol to image the 20 cm EPT configurations by an average of 3.7% and higher when using the ‘Clarity Low’ protocol to image the 30 cm EPT configurations by an average of 5.7%.

- ICR scores were higher when using the ‘Clarity Low’ protocol by an average of 27.5%.

- Spatial resolution scores were higher when using the ‘Clarity Low’ protocol by an average of 19.8%.

- Total WTR scores were lower when using the ‘Clarity Low’ protocol by an average 6.8% with exception to the 30 cm EPT, 15 cm DFS configuration in which the ‘Cardiac Low’ protocol scored higher than the ‘Cardiac Low’ protocol scores by 3.8%.

- Static MTR scores were lower when using the ‘Clarity Low’ protocol by an average of 6.4% with exception to the 30 cm EPT, 15 cm DFS configuration in which the ‘Clarity Low’ protocol scored higher than the ‘Cardiac Low’ protocol scores by 10%.

- Dynamic MTR scores were lower when using the ‘Clarity Low’ protocol to image the 20 cm EPT configurations by an average of 3.2% and was comparable between protocols when imaging the 30 cm EPT configurations.
Figure A.4: Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity Normal’ and ‘Cardiac 4’ acquisition mode protocols for all phantom and imaging configurations

Table A.3: Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity Low’ and ‘Cardiac Low’ acquisition mode protocols for all phantom and imaging configurations
A.0.5 Overview of the Clarity Upgrade: Fluoroscopy mode protocols

The fluoroscopy mode results of the ‘Clarity’ upgrade assessment have been presented graphically and in tabular form. An overview of the acquisition mode results has been provided in Figure A.5. In this graph the ESD values recorded by the MOSkin were normalised to the ‘Cardiac Low’ fluoroscopy mode dose level 1 values for each respective EPT/FOV configuration. Figure A.6 to Figure A.9 and Table A.4 to Table A.7 represent each individual pairing of low dose and high detail protocols in further detail using the ‘Score Ratio’ method described in Equation 4.2. Tables are labelled in the format ‘Protocol B’ : ‘Protocol A’.

Figure A.5: Dose delivery and image quality scores produced by the Philips Allura Xper and Philips Allura Clarity c-arm systems in fluoroscopy modes 1 and 2 using the a) 30 cm phantom configuration and 15 cm DFS setting, b) 30 cm phantom configuration and 20 cm DFS setting c) 20 cm phantom configuration and 15 cm DFS setting and the d) 20 cm phantom configuration and 20 cm DFS setting
A.0.6 Comparison of the fluoroscopy mode dose level 1 protocols: ‘Clarity’ and ‘Cardiac 4’

The ratio of ESD delivery and image quality scores recorded for the ‘Clarity’ and ‘Cardiac 4’ fluoroscopy mode dose level 1 protocols have been presented graphically in Figure A.6 and tabularly in Table A.4. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Clarity’ protocol by an average of 36%.
- ICR scores were lower when using the ‘Clarity’ protocol by an average of 82.8% with exception to the 30 cm EPT, 20 cm DFS configuration where both protocols scored comparably.
- Spatial resolution scores were lower when using the ‘Clarity’ protocol by an average of 17.5% with exception to the 20 cm EPT, 20 cm DFS configuration in which the ‘Clarity’ protocol scored higher than the ‘Cardiac 4’ protocol by 0.4%.
- Total WTR scores were lower when using the ‘Clarity’ protocol by an average of 9.1%.
- Static MTR scores were higher when using the ‘Clarity’ protocol by an average of 25.1%
- Dynamic MTR scores were higher when using the ‘Clarity’ protocol by an average of 23.8%.
Figure A.6: Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac 4’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.

Table A.4: Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac 4’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.
A.0.7 Comparison of the fluoroscopy mode dose level 2 protocols: The ‘Clarity’ and ‘Cardiac 4’ protocols

The ratio of ESD delivery and image quality scores recorded for the ‘Clarity’ and ‘Cardiac 4’ fluoroscopy mode dose level 2 protocols have been presented graphically in Figure A.7 and tabularly in Table A.5. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was higher when using the ‘Clarity’ protocol to image the 20 cm EPT configurations by an average of 55.4% and lower when using the ‘Clarity’ protocol to image the 30 cm EPT configurations by an average of 48.3%.

- ICR scores were lower when using the ‘Clarity’ protocol by an average of 45% with exception to the 20 cm EPT, 20 cm DFS configuration in which the ‘Clarity’ protocol scored higher than the ‘Cardiac 4’ protocol by 350%.

- Spatial resolution scores were higher when using the ‘Clarity’ protocol to image the 20 cm EPT configurations by an average of 9.5% and were lower when using the ‘Clarity Low’ protocol to image the 30 cm EPT configurations by an average of 14.7%.
• Total WTR scores were higher when using the ‘Clarity’ protocol by an average of 5.5% with exception to the 20 cm EPT, 15 cm DFS configuration in which the ‘Clarity’ protocol scored lower than the ‘Cardiac 4’ protocol by 6.9%.

• Static MTR scores were lower when using the ‘Clarity’ protocol by an average of 28.2% with exception to the 20 cm EPT, 20 cm DFS configuration in which the ‘Clarity’ protocol scored higher than the ‘Cardiac 4’ protocol by 9.1%.

• Dynamic MTR scores were lower when using the ‘Clarity’ protocol by an average of 29.3% with exception to the 20 cm EPT, 20 cm DFS configuration in which the ‘Clarity’ protocol scored higher than the ‘Cardiac 4’ protocol by 6.5%.
Figure A.7: Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac 4’ fluoroscopy mode protocols in the dose level 2 setting for all phantom and imaging configurations

Table A.5: Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac 4’ fluoroscopy mode protocols in the dose level 2 setting for all phantom and imaging configurations
A.0.8 Comparison of the fluoroscopy mode dose level 1 protocols: The ‘Clarity’ and ‘Cardiac Low’ protocols

The ratio of ESD delivery and image quality scores recorded for the ‘Clarity’ and ‘Cardiac Low’ fluoroscopy mode dose level 1 protocols have been presented graphically in Figure A.8 and tabularly in Table A.6. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was higher when using the ‘Clarity’ protocol to image the 20 cm EPT configurations by an average of 309.4%, was lower when using the ‘Clarity’ protocol to image the 30 cm EPT, 15 cm DFS configuration by 10.2% and was higher when using the ‘Clarity’ protocol to image the 30 cm EPT, 20 cm DFS configuration by 30.3%.

- ICR scores were lower when using the ‘Clarity’ protocol to image the 20 cm EPT, 15 cm DFS and 30 cm EPT, 20 cm DFS configurations by an average of 68.4%, and were higher when using the ‘Clarity’ protocol to image the 20 cm EPT, 20 cm DFS and 30 cm EPT, 15 cm DFS configurations by an average of 116.7%.

- Spatial resolution scores were higher when using the ‘Clarity’ protocol to image the 20 cm EPT configurations by an average of 15.9% and lower when using the ‘Clarity’ protocol to image the 30 cm EPT configurations by an average of 18.9%.
• Total WTR scores were lower when using the ‘Clarity’ protocol by an average of 3.9% with exception to the 30 cm EPT, 20 cm DFS configuration in which the ‘Clarity’ protocol scored higher than the ‘Cardiac 4’ protocol by 16.6%.

• Static MTR scores were higher when using the ‘Clarity’ protocol to image the 20 cm EPT configurations by an average 22.1% and were lower when using the ‘Clarity’ protocol to image the 30 cm EPT configurations by an average of 17.7%.

• Dynamic MTR scores were higher when using the ‘Clarity’ protocol by an average of 18.6% with exception to the 30 cm EPT, 15 cm DFS configuration in which the ‘Clarity’ protocol scored higher than the ‘Cardiac 4’ protocol by 35.7%.
**Figure A.8:** Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac 4’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.

**Table A.6:** Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac Low’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.
A.0.9 Comparison of the low fluoroscopy mode dose level 2 protocols: The ‘Clarity’ and ‘Cardiac Low’ protocols

The ratio of ESD delivery and image quality scores recorded for the ‘Clarity’ and ‘Cardiac Low’ fluoroscopy mode dose level 2 protocols have been presented graphically in Figure A.9 and tabularly in Table A.7. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was higher when using the ‘Clarity’ protocol to image the 20 cm EPT configurations by an average of 92%, was lower when using the ‘Clarity’ protocol to image the 30 cm EPT, 15 cm DFS configuration by 18.3% and was higher when using the ‘Clarity’ protocol to image the 30 cm EPT, 20 cm DFS configuration by 12.7%.

- ICR scores were lower when using the ‘Clarity’ protocol by an average of 23.4% with exception to the 30 cm EPT, 15 cm DFS configuration in which the ‘Clarity’ protocol scored higher than the ‘Cardiac Low’ protocol by 300%.

- Spatial resolution scores were higher when using the ‘Clarity’ protocol to image the 20 cm EPT configurations by an average of 16.7% and lower when using the ‘Clarity’ protocol to image the 30 cm EPT configurations by an average of 7%.
• Total WTR scores were higher when using the ‘Clarity’ protocol by an average of 26.2%.

• Static MTR scores were lower when using the ‘Clarity’ protocol to image the 20 cm EPT, 15 cm DFS configuration by 14%, were higher when using the ‘Clarity’ protocol to image the 20 cm EPT, 20 cm DFS configuration by 5.9% and were lower when using the ‘Clarity’ protocol to image the 30 cm EPT configurations by an average of 32.9%.

• Dynamic MTR scores were lower when using the ‘Clarity’ protocol by an average of 13.1%.
Figure A.9: Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac Low’ fluoroscopy mode protocols in the dose level 2 setting for all phantom and imaging configurations

<table>
<thead>
<tr>
<th>Fluoroscopy Mode 2</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Clarity : Cardiac Low</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EPT: 20 cm DFS: 15 cm</td>
<td>1.826</td>
<td>0.767</td>
<td>1.155</td>
<td>1.234</td>
<td>0.86</td>
<td>0.895</td>
</tr>
<tr>
<td>EPT: 20 cm DFS: 20 cm</td>
<td>2.013</td>
<td>0.863</td>
<td>1.178</td>
<td>1.361</td>
<td>1.059</td>
<td>0.971</td>
</tr>
<tr>
<td>EPT: 30 cm DFS: 15 cm</td>
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<td>4</td>
<td>0.9</td>
<td>1.094</td>
<td>0.696</td>
<td>0.762</td>
</tr>
<tr>
<td>EPT: 30 cm DFS: 15 cm</td>
<td>1.127</td>
<td>0.667</td>
<td>0.96</td>
<td>1.358</td>
<td>0.667</td>
<td>0.85</td>
</tr>
</tbody>
</table>

Table A.7: Ratios of the dose delivery and image quality scores recorded when using the ‘Clarity’ and ‘Cardiac Low’ fluoroscopy mode protocols in the dose level 2 setting for all phantom and imaging configurations
Appendix B

GE ‘Dose Map’ Upgrade Evaluation
B.0.1 Overview of the GE Upgrade Evaluation

The results of the GE c-arm systems have been presented in this appendix in a tabular format only. The acquisition mode dose level 1 results have been presented in Table B.1 to Table B.6 while the acquisition mode dose level 2 results have been presented in Table B.7 and Table B.12. These tables compare the dose and image quality score results of protocols used on the pre-upgrade Innova 2100IQ c-arm system, the post-upgrade Innova 21000IQ c-arm system and the post-upgrade IGS 520 c-arm system commissioned at Eastern Heart Clinic. The fluoroscopy mode dose level 1 results have been presented in Table B.13 and Table B.17 while the fluoroscopy mode dose level 2 results have been presented in Table B.18 and Table B.20. The fluoroscopy mode tables compare the dose and image quality score results of protocols used on the pre-upgrade Innova 2100IQ c-arm system, the post-upgrade Innova 21000IQ c-arm system.

The results have then been categorised and averaged using the following classifications:

- Protocol
- Phantom EPT
- Field size
- General imaging modality
- Imaging Dose level
B.0.2 Comparison of the high detail acquisition mode dose level 1 protocols: ‘Standard’ and ‘CoroPlus’

The ratio of ESD delivery and image quality scores recorded for the ‘Standard’ and ‘CoroPlus’ acquisition mode dose level 1 protocols have been presented tabularly in Table B.1. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Standard’ protocol by an average of 21.0%.

- ICR scores were higher when using the ‘Standard’ protocol by an average of 11% with exception to the 20 cm EPT, 12 cm OFS configuration in which the ‘Standard’ protocol scored lower than the ‘CoroPlus’ protocol by 10%.

- Spatial resolution scores were lower when using the ‘Standard’ protocol by an average of 13.3%.

- Total WTR scores were comparable between protocols.

- Static MTR scores were comparable between protocols.

- Dynamic MTR scores were comparable between protocols.

- Image noise was lower when using the ‘Standard’ protocol by an average of 34% when imaging the 20 cm EPT configurations and lower when using the ‘Standard’ protocol and by an average of 62.6% when imaging the 30 cm EPT configurations.
<table>
<thead>
<tr>
<th>Acquisition Mode 1</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>EPT : 20 cm</td>
<td>0.789</td>
<td>1.125</td>
<td>0.893</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
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<td></td>
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<tr>
<td>EPT : 20 cm</td>
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<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.615</td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 30 cm</td>
<td>0.622</td>
<td>0.900</td>
<td>0.880</td>
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<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
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<td></td>
</tr>
<tr>
<td>EPT : 30 cm</td>
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<td></td>
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<tr>
<td>Average</td>
<td>0.790</td>
<td>1.058</td>
<td>0.867</td>
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<td>0.517</td>
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<tr>
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<td>1.143</td>
<td>0.909</td>
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<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.705</td>
</tr>
<tr>
<td>Minimum Value</td>
<td>0.622</td>
<td>0.900</td>
<td>0.786</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.311</td>
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<td>1</td>
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<tr>
<td>Average for OFS : 15 cm</td>
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<td>0.374</td>
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<tr>
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<td>0.705</td>
<td>1.013</td>
<td>0.886</td>
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<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.572</td>
</tr>
<tr>
<td>Average for EPT : 30 cm</td>
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<td>1.102</td>
<td>0.847</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.462</td>
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</table>

Table B.1: Ratios of the dose delivery and image quality scores recorded when using the ‘Standard’ and ‘CoroPlus’ acquisition mode protocols in the dose level 1 setting for all phantom and imaging configurations.
The ratio of ESD delivery and image quality scores recorded for the ‘Low’ and ‘RDLS’ acquisition mode dose level 1 protocols have been presented tabularly in Table B.2. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Low’ protocol by an average of 6% with exception to the 30 cm EPT, 12 cm OFS configuration in which the ‘Low’ protocol delivered higher doses than the ‘RDLS’ protocol by 4.5%.

- ICR scores were higher when using the ‘Low’ protocol by an average of 32% when imaging the 12 cm OFS configurations, were higher when using the ‘Low’ protocol to image the 20 cm EPT, 15 cm OFS configuration by 7.1% and were comparable between protocols when imaging the 30 cm EPT, 15 cm OFS configuration.

- Spatial resolution scores were lower when using the ‘Low’ protocol by an average of 10.6% with exception to the 30 cm EPT, 12 cm OFS configuration where both protocols scored comparably.
• Total WTR scores were comparable between protocols with exception to the 20 cm EPT, 12 cm OFS configuration in which the 'Low' protocol acquired twice as many high density WTR targets as the 'RDLS' protocol.

• Static MTR scores were comparable between protocols.

• Dynamic MTR scores were higher when using the ‘Low’ protocol to image the 20 cm ETP, 12 cm OFS configuration by 33.3%, were comparable between protocols when imaging the 20 cm EPT, 15 cm OFS and 30 cm EPT, 12 cm OFS configurations and were lower when using the ‘Low’ protocol to image the 30 cm EPT, 15 cm OFS configuration by 33.3%.

• Image noise was lower when using the ‘Low’ protocol by an average of 39.4% when imaging the 20 cm EPT configurations and lower by an average of 65.4% when imaging the 30 cm EPT configurations.
<table>
<thead>
<tr>
<th>Acquisition Mode 1</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>Low : RDLS</td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
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<td>EPT : 20 cm</td>
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</tr>
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<td>OFS : 12 cm</td>
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<td>1</td>
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<td>1</td>
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</tr>
<tr>
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<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 30 cm</td>
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<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.666</td>
<td>0.321</td>
</tr>
<tr>
<td>OFS : 15 cm</td>
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<td>0.920</td>
<td>1</td>
<td>1.250</td>
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<td>1</td>
<td>1.333</td>
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</tr>
<tr>
<td>Maximum Value</td>
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<td>1</td>
<td>1</td>
<td>0.666</td>
<td>0.321</td>
</tr>
<tr>
<td>Minimum Value</td>
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<td>0.886</td>
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<td>1.500</td>
<td>1.100</td>
<td>1</td>
<td>1.167</td>
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</tr>
<tr>
<td>Average for OFS : 12 cm</td>
<td>0.963</td>
<td>1.167</td>
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<td>1</td>
<td>0.833</td>
<td>0.346</td>
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<tr>
<td>Average for OFS : 15 cm</td>
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<td>1.320</td>
<td>0.946</td>
<td>1</td>
<td>1.500</td>
<td>1.100</td>
<td>1</td>
<td>1.167</td>
<td>0.537</td>
</tr>
<tr>
<td>Average for EPT : 20 cm</td>
<td>1.017</td>
<td>1.036</td>
<td>0.895</td>
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<td>1</td>
<td>1</td>
<td>1</td>
<td>0.833</td>
<td>0.415</td>
</tr>
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</table>

Table B.2: Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘RDLS’ acquisition mode protocols in the dose level 1 setting for all phantom and imaging configurations.
B.0.4 Comparison of the post-upgrade acquisition mode dose level 1 protocols: ‘Low’ and ‘Standard’

The ratio of ESD delivery and image quality scores recorded for the ‘Low’ and ‘Standard’ acquisition mode dose level 1 protocols have been presented tabularly in Table B.3. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Low’ protocol to image the 20 cm EPT configurations by an average of 40.6%, was lower when using the ‘Low’ protocol to image the 30 cm EPT, 12 cm OFS configuration by 4% and was higher when using the ‘Low’ protocol to image the 30 cm EPT, 15 cm OFS configuration by 2.1%.

- ICR scores were lower when using the ‘Low’ protocol to image the 12 cm OFS configurations by an average of 8.4% and were lower when using the ‘Low’ protocol to image the 15 cm OFS configurations by an average of 18.4%.

- Spatial resolution scores were comparable between protocols.
• Total WTR scores were comparable between protocols with exception to the 20 cm EPT, 12 cm OFS configuration in which the ‘Low’ protocol acquired twice as many high density WTR targets as the ‘RDLS’ protocol.

• Static MTR scores were comparable between protocols.

• Dynamic MTR scores were comparable between protocols with exception to the 30 cm EPT, 15 cm OFS configuration in which the ‘Standard’ protocol scored higher than the ‘Low’ protocol by 50%.

• Image noise was lower when using the ‘Low’ protocol by an average of 3.4% with exception to the 30 cm EPT, 15 cm OFS configuration in which the ‘Low’ protocol scored higher than the ‘Standard’ protocol by 2.4%.
<table>
<thead>
<tr>
<th>Acquisition Mode 1</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Low : Standard</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 20 cm OFS : 12 cm</td>
<td>0.596</td>
<td>0.944</td>
<td>1</td>
<td>1</td>
<td>2</td>
<td>1.200</td>
<td>1</td>
<td>1</td>
<td>0.990</td>
</tr>
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<td>0.882</td>
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<td>1</td>
<td>0.963</td>
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<td>EPT : 30 cm OFS : 12 cm</td>
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<td>0.889</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.944</td>
</tr>
<tr>
<td>EPT : 30 cm OFS : 15 cm</td>
<td>1.021</td>
<td>0.750</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.500</td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td>0.793</td>
<td>0.866</td>
<td>1</td>
<td>1</td>
<td>1.250</td>
<td>1.050</td>
<td>1</td>
<td>1.125</td>
<td>0.980</td>
</tr>
<tr>
<td><strong>Maximum Value</strong></td>
<td>1.021</td>
<td>0.944</td>
<td>1</td>
<td>2</td>
<td>1.200</td>
<td>1</td>
<td>1</td>
<td>1.500</td>
<td></td>
</tr>
<tr>
<td><strong>Minimum Value</strong></td>
<td>0.593</td>
<td>0.750</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.944</td>
</tr>
<tr>
<td><strong>Average for</strong></td>
<td>0.595</td>
<td>0.913</td>
<td>1</td>
<td>1</td>
<td>1.500</td>
<td>1.100</td>
<td>1</td>
<td>1</td>
<td>0.977</td>
</tr>
<tr>
<td>OFS : 12 cm</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Average for OFS : 15 cm</td>
<td>0.991</td>
<td>0.820</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.250</td>
</tr>
<tr>
<td><strong>Average for</strong></td>
<td>0.778</td>
<td>0.917</td>
<td>1</td>
<td>1</td>
<td>1.500</td>
<td>1.100</td>
<td>1</td>
<td>1</td>
<td>0.967</td>
</tr>
<tr>
<td>EPT : 20 cm</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Average for EPT : 30 cm</td>
<td>0.807</td>
<td>0.816</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.250</td>
</tr>
</tbody>
</table>

Table B.3: Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘Standard’ acquisition mode protocols in the dose level 1 setting for all phantom and imaging configurations.
B.0.5 Comparison of the post-upgrade acquisition mode dose level 1 protocols: ‘Very Low’ and ‘Low’

The ratio of ESD delivery and image quality scores recorded for the ‘Very Low’ and ‘Low’ acquisition mode dose level 1 protocols have been presented tabularly in Table B.4. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Very Low’ protocol to image the 20 cm EPT configurations by an average of 36.2% and lower when using the ‘Very Low’ protocol to image the 30 cm EPT configurations by an average of 7.8%.

- ICR scores were lower when using the ‘Very Low’ protocol by an average of 20.9% with exception to the 30 cm EPT, 15 cm OFS configuration where both protocols scored comparably.

- Spatial resolution scores were comparable between protocols.

- Total WTR scores were higher when using the ‘Very Low’ protocol to image the 20 cm EPT configurations where the ‘Very Low’ protocol acquired 17.5% more high density WTR targets than the ‘Low’ protocol and were when using the ‘Very Low’ protocol to image the 30 cm EPT configurations where the ‘Very Low’ protocol acquired 30% high low density WTR targets than the ‘Low’ protocol.
- Static MTR scores were comparable between protocols.

- Dynamic MTR scores were comparable between protocols when imaging the 20 cm EPT, 12 cm OFS and 30 cm EPT, 15 cm OFS configurations, and lower when using the ‘Very Low’ protocol to image the 20 cm EPT, 15 cm OFS and 30 cm EPT, 12 cm OFS configurations by an average of 29.2%.

- Image noise was lower when using the ‘Low’ protocol by an average of 3.4% with exception to the 30 cm EPT, 15 cm OFS configuration in which the ‘Very Low’ protocol scored higher than the ‘Standard’ protocol by 2.4%. 
<table>
<thead>
<tr>
<th>Acquisition Mode 1 Very Low : Low</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>EPT : 20 cm, OFS : 12 cm</td>
<td>0.608</td>
<td>0.823</td>
<td>1</td>
<td>1</td>
<td>1.250</td>
<td>1.083</td>
<td>1</td>
<td>1</td>
<td>1.041</td>
</tr>
<tr>
<td>EPT : 20 cm, OFS : 15 cm</td>
<td>0.668</td>
<td>0.800</td>
<td>1</td>
<td>1</td>
<td>1.100</td>
<td>1.033</td>
<td>1</td>
<td>0.750</td>
<td>1.009</td>
</tr>
<tr>
<td>EPT : 30 cm, OFS : 12 cm</td>
<td>0.947</td>
<td>0.750</td>
<td>1</td>
<td>1</td>
<td>0.733</td>
<td>0.885</td>
<td>1</td>
<td>0.666</td>
<td>1.130</td>
</tr>
<tr>
<td>EPT : 30 cm, OFS : 15 cm</td>
<td>0.897</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.666</td>
<td>0.857</td>
<td>1</td>
<td>1</td>
<td>1.070</td>
</tr>
<tr>
<td>Average</td>
<td>0.780</td>
<td>0.843</td>
<td>1</td>
<td>1</td>
<td>0.937</td>
<td>0.965</td>
<td>1</td>
<td>0.854</td>
<td>1.063</td>
</tr>
<tr>
<td>Maximum Value</td>
<td>0.947</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.250</td>
<td>1.083</td>
<td>1</td>
<td>1</td>
<td>1.130</td>
</tr>
<tr>
<td>Minimum Value</td>
<td>0.608</td>
<td>0.750</td>
<td>1</td>
<td>1</td>
<td>0.666</td>
<td>0.857</td>
<td>1</td>
<td>0.666</td>
<td>1.009</td>
</tr>
<tr>
<td>Average for OFS : 12 cm</td>
<td>0.638</td>
<td>0.812</td>
<td>1</td>
<td>1</td>
<td>1.175</td>
<td>1.058</td>
<td>1</td>
<td>0.875</td>
<td>1.025</td>
</tr>
<tr>
<td>Average for OFS : 15 cm</td>
<td>0.922</td>
<td>0.875</td>
<td>1</td>
<td>1</td>
<td>0.700</td>
<td>0.871</td>
<td>1</td>
<td>0.833</td>
<td>1.100</td>
</tr>
<tr>
<td>Average for EPT : 20 cm</td>
<td>0.778</td>
<td>0.787</td>
<td>1</td>
<td>1</td>
<td>0.992</td>
<td>0.984</td>
<td>1</td>
<td>0.833</td>
<td>1.086</td>
</tr>
<tr>
<td>Average for EPT : 30 cm</td>
<td>0.783</td>
<td>0.900</td>
<td>1</td>
<td>1</td>
<td>0.883</td>
<td>0.945</td>
<td>1</td>
<td>0.875</td>
<td>1.040</td>
</tr>
</tbody>
</table>

Table B.4: Ratios of the dose delivery and image quality scores recorded when using the ‘Very Low’ and ‘Low’ acquisition mode protocols in the dose level 1 setting for all phantom and imaging configurations.
B.0.6 Comparison of the SHC and EHC high detail acquisition mode dose level 1 protocols: ‘Standard’ and ‘EHC-IQ+’

The ratio of ESD delivery and image quality scores recorded for the ‘Standard’ and ‘EHC-IQ+’ acquisition mode dose level 1 protocols have been presented tabularly in Table B.5. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was higher when using the ‘Standard’ protocol to image the 20 cm EPT configurations by an average of 22.3% and lower when using the ‘Standard’ protocol to image the 30 cm EPT configurations by an average of 12.4%.

- ICR scores were higher when using the ‘Standard’ protocol by an average of 19%.

- Spatial resolution scores were comparable between protocols.

- Total WTR scores comparable between protocols with exception to the 20 cm EPT, 12 cm OFS configuration in which the ‘Standard’ protocol acquired half as many high density WTR targets as the ‘EHC-IQ+’ protocol.

- Static MTR scores were comparable between protocols.

- Dynamic MTR scores were comparable between protocols.

- Image noise was lower when using the ‘Standard’ protocol by an average of 28.5% when imaging the 20 cm EPT configurations and lower by an average of 40.3% when imaging the 30 cm EPT configurations.
<table>
<thead>
<tr>
<th>Acquisition Mode 1 Standard : EHC-IQ+</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>EPT : 20 cm OFS : 12 cm</td>
<td>1.315</td>
<td>1.200</td>
<td>1</td>
<td>1</td>
<td>0.500</td>
<td>0.833</td>
<td>1</td>
<td>1</td>
<td>0.736</td>
</tr>
<tr>
<td>EPT : 20 cm OFS : 15 cm</td>
<td>1.130</td>
<td>1.133</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.694</td>
</tr>
<tr>
<td>EPT : 30 cm OFS : 12 cm</td>
<td>0.897</td>
<td>1.285</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.570</td>
</tr>
<tr>
<td>EPT : 30 cm OFS : 15 cm</td>
<td>0.855</td>
<td>1.142</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.624</td>
</tr>
<tr>
<td>Average</td>
<td>1.049</td>
<td>1.190</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.875</td>
<td>0.958</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>Maximum Value</td>
<td>1.315</td>
<td>1.285</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.736</td>
</tr>
<tr>
<td>Minimum Value</td>
<td>0.855</td>
<td>1.133</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.500</td>
<td>0.833</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>Average for OFS : 12 cm</td>
<td>1.223</td>
<td>1.167</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.750</td>
<td>0.917</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>Average for OFS : 15 cm</td>
<td>0.876</td>
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<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.597</td>
</tr>
<tr>
<td>Average for EPT : 20 cm</td>
<td>1.106</td>
<td>1.243</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.750</td>
<td>0.917</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>Average for EPT : 30 cm</td>
<td>0.993</td>
<td>1.138</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.659</td>
</tr>
</tbody>
</table>

Table B.5: Ratios of the dose delivery and image quality scores recorded when using the 'Standard' and 'EHC-IQ+' acquisition mode protocols in the dose level 1 setting for all phantom and imaging configurations.
B.0.7 Comparison of the SHC and EHC low dose rate acquisition mode dose level 1 protocols: ‘Low’ and ‘EHC-RDLS’

The ratio of ESD delivery and image quality scores recorded for the ‘Low’ and ‘EHC-RDLS’ acquisition mode dose level 1 protocols have been presented tabularly in Table B.6. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was higher when using the ‘Low’ protocol to image the 20 cm EPT configurations by an average of 41.8% and lower when using the ‘Low’ protocol to image the 30 cm EPT configurations by an average of 4.7%.

- ICR scores were higher when using the ‘Standard’ protocol by an average of 23.3% with exception to the 30 cm EPT, 15 cm OFS configuration in which the ‘Low’ protocol scored higher than the ‘EHC-RDLS’ protocol by 14.3%.

- Spatial resolution scores were comparable between protocols with exception to the 30 cm EPT, 15 cm OFS configuration in which the ‘Low’ protocol scored higher than the ‘EHC-RDLS’ protocol by 14.3%.
• Total WTR scores were comparable between protocols with exception to the 20 cm EPT, 12 cm OFS configuration in which the ‘Low’ protocol acquired twice as many high density WTR targets as the ‘EHC-RDLS’ protocol.

• Static MTR scores were comparable between protocols.

• Dynamic MTR scores were comparable between protocols with exception to the 30 cm EPT, 15 cm OFS configuration in which the ‘Low’ protocol scored lower than the ‘EHC-RDLS’ protocol by 33.3%.

• Image noise was lower when using the ‘Low’ protocol by an average of 40%.
## Table B.6:
Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘EHC-RDLS’ acquisition mode protocols in the dose level 1 setting for all phantom and imaging configurations.

<table>
<thead>
<tr>
<th>Acquisition Mode 1</th>
<th>Spatial Resolution</th>
<th>Image Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>Low: EHC-RDLS</td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>ESD</strong></td>
<td><strong>ICR</strong></td>
<td><strong>Static MTR</strong></td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td><strong>Dynamic MTR</strong></td>
<td><strong>Total MTR</strong></td>
</tr>
<tr>
<td><strong>Upper WTR</strong></td>
<td><strong>Lower WTR</strong></td>
<td><strong>WTR</strong></td>
</tr>
<tr>
<td><strong>EPT : 20 cm</strong></td>
<td>1.200</td>
<td>1.050</td>
</tr>
<tr>
<td><strong>OFS : 12 cm</strong></td>
<td>1</td>
<td>1.000</td>
</tr>
<tr>
<td><strong>EPT : 20 cm</strong></td>
<td>1.136</td>
<td>1.050</td>
</tr>
<tr>
<td><strong>OFS : 15 cm</strong></td>
<td>1</td>
<td>1.000</td>
</tr>
<tr>
<td><strong>EPT : 30 cm</strong></td>
<td>1.200</td>
<td>1.050</td>
</tr>
<tr>
<td><strong>OFS : 12 cm</strong></td>
<td>1</td>
<td>1.000</td>
</tr>
<tr>
<td><strong>EPT : 30 cm</strong></td>
<td>1.405</td>
<td>1.050</td>
</tr>
<tr>
<td><strong>OFS : 15 cm</strong></td>
<td>1</td>
<td>1.000</td>
</tr>
<tr>
<td><strong>Average</strong></td>
<td><strong>Maximum Value</strong></td>
<td></td>
</tr>
<tr>
<td><strong>Minimum Value</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Average for</strong></td>
<td><strong>WTR</strong></td>
<td><strong>WTR</strong></td>
</tr>
<tr>
<td><strong>OFS : 12 cm</strong></td>
<td>1</td>
<td>1.000</td>
</tr>
<tr>
<td><strong>OFS : 15 cm</strong></td>
<td>1</td>
<td>1.000</td>
</tr>
<tr>
<td><strong>Average for</strong></td>
<td><strong>WTR</strong></td>
<td><strong>WTR</strong></td>
</tr>
<tr>
<td><strong>OFS : 20 cm</strong></td>
<td>1</td>
<td>1.000</td>
</tr>
<tr>
<td><strong>OFS : 30 cm</strong></td>
<td>1</td>
<td>1.000</td>
</tr>
<tr>
<td><strong>Average for</strong></td>
<td><strong>WTR</strong></td>
<td><strong>WTR</strong></td>
</tr>
<tr>
<td><strong>OFS : 20 cm</strong></td>
<td>1</td>
<td>1.000</td>
</tr>
<tr>
<td><strong>OFS : 30 cm</strong></td>
<td>1</td>
<td>1.000</td>
</tr>
</tbody>
</table>

**Protocols**
- Protocol B: Performed Best
- Protocol A: Performed Worst

**Bold and Italicised Formatting**
- Most Improved EPT/FOV Configuration
- Protocols Comparable
B.0.8 Comparison of the post-upgrade acquisition mode dose level 2 protocols: ‘Standard’ and ‘CoroPlus’

The ratio of ESD delivery and image quality scores recorded for the ‘Standard’ and ‘CoroPlus’ acquisition mode dose level 2 protocols have been presented tabularly in Table B.7. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Standard’ protocol by an average of 31.0%.
- ICR scores were higher when using the ‘Standard’ protocol by an average of 27.7% with exception to the 30 cm EPT, 15 cm OFS configuration where both protocols scored comparably.
- Spatial resolution scores were lower when using the ‘Standard’ protocol by an average of 13.3%.
- Total WTR scores were comparable between protocols with exception to the 20 cm EPT, 12 cm OFS configuration in which the ‘Standard’ protocol acquired half as many high density WTR targets as the ‘CoroPlus’ protocol.
- Static MTR scores were lower when using the ‘Standard’ protocol by an average of 22.5% when imaging the 12 cm OFS configurations and comparable when imaging using the 15 cm OFS configuration.
- Dynamic MTR scores were comparable between protocols.
- Image noise was lower when using the ‘Standard’ protocol by an average of 35.8%.
<table>
<thead>
<tr>
<th>Acquisition Mode 2</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>Standard : CoroPlus</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 20 cm, OFS : 12 cm</td>
<td>0.590</td>
<td>1.130</td>
<td>0.890</td>
<td>1</td>
<td>2</td>
<td>1.200</td>
<td>0.800</td>
<td>1</td>
<td>0.840</td>
</tr>
<tr>
<td>EPT : 20 cm, OFS : 15 cm</td>
<td>0.637</td>
<td>1.130</td>
<td>0.790</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.650</td>
</tr>
<tr>
<td>EPT : 30 cm, OFS : 12 cm</td>
<td>0.675</td>
<td>1</td>
<td>0.880</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.750</td>
<td>1</td>
<td>0.580</td>
</tr>
<tr>
<td>EPT : 30 cm, OFS : 15 cm</td>
<td>0.863</td>
<td>1.570</td>
<td>0.910</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.500</td>
</tr>
<tr>
<td>Average Maximum Value</td>
<td>0.863</td>
<td>1.570</td>
<td>0.910</td>
<td>1</td>
<td>2</td>
<td>1.200</td>
<td>1</td>
<td>1</td>
<td>0.840</td>
</tr>
<tr>
<td>Average Minimum Value</td>
<td>0.590</td>
<td>1</td>
<td>0.790</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.750</td>
<td>1</td>
<td>0.500</td>
</tr>
<tr>
<td>Average for EPT : 20 cm, OFS : 12 cm</td>
<td>0.615</td>
<td>1.130</td>
<td>0.840</td>
<td>1</td>
<td>1.500</td>
<td>1.100</td>
<td>0.900</td>
<td>1</td>
<td>0.745</td>
</tr>
<tr>
<td>Average for EPT : 20 cm, OFS : 15 cm</td>
<td>0.765</td>
<td>1.285</td>
<td>0.895</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.875</td>
<td>1</td>
<td>0.540</td>
</tr>
<tr>
<td>Average for EPT : 30 cm, OFS : 12 cm</td>
<td>0.630</td>
<td>1.065</td>
<td>0.885</td>
<td>1</td>
<td>1.500</td>
<td>1.100</td>
<td>0.775</td>
<td>1</td>
<td>0.710</td>
</tr>
<tr>
<td>Average for EPT : 30 cm, OFS : 15 cm</td>
<td>0.750</td>
<td>1.350</td>
<td>0.850</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.575</td>
</tr>
</tbody>
</table>

**Table B.7:** Ratios of the dose delivery and image quality scores recorded when using the ‘Standard’ and ‘CoroPlus’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.
B.0.9 Comparison of the post-upgrade acquisition mode dose level 2 protocols: ‘Low’ and ‘RDLS’

The ratio of ESD delivery and image quality scores recorded for the ‘Low’ and ‘RDLS’ acquisition mode dose level 2 protocols have been presented tabularly in Table B.8. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was higher when using the ‘Low’ protocol by an average of 6.5% with exception to the 30 cm EPT, 12 cm OFS configuration where ESD delivery was lower when using the ‘Low’ protocol by 6.4%.

- ICR scores were higher when using the ‘Low’ protocol by an average of 18.3%.

- Spatial resolution scores were comparable between protocols.

- Total WTR scores were comparable between protocols.

- Static MTR scores were comparable between protocols.

- Dynamic MTR scores were comparable between protocols.

- Image noise was lower when using the ‘Low’ protocol by an average of 46% with the exception of the 20 cm EPT, 12 cm OFS configuration where image noise was higher by 149%.
Table B.8: Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘RDLS’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.
B.0.10 Comparison of the post-upgrade acquisition mode dose level 2 protocols: ‘Low’ and ‘Standard’

The ratio of ESD delivery and image quality scores recorded for the ‘Low’ and ‘Standard’ acquisition mode dose level 2 protocols have been presented tabularly in Table B.9. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Low’ protocol to image the 20 cm EPT configurations by an average of 36%, was lower when using the ‘Low’ protocol to image the 30 cm EPT, 12 cm OFS configuration by 3.4% and was higher when using the ‘Low’ protocol to image the 30 cm EPT, 15 cm OFS configuration by 6.1%.

- ICR scores were lower when using the ‘Low’ protocol by an average of 9.6%.

- Spatial resolution scores were comparable between protocols.

- Total WTR scores were comparable between protocols.

- Static MTR scores were comparable between protocols.

- Dynamic MTR scores were comparable between protocols.

- Image noise was lower when using the ‘Low’ protocol to image the 12 cm OFS configurations by an average of 2.7% and higher when using the ‘Low’ protocol to image the 15 cm OFS configurations by an average of 5%.
Table B.9: Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘Standard’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.
B.0.11 Comparison of the post-upgrade acquisition mode dose level 2 protocols: ‘Very Low’ and ‘Low’

The ratio of ESD delivery and image quality scores recorded for the ‘Very Low’ and ‘Low’ acquisition mode dose level 2 protocols have been presented tabularly in Table B.10. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Very Low’ protocol to image the 20 cm EPT configurations by an average of 31.9% and lower when using the ‘Very Low’ protocol to image the 30 cm EPT configuration by an average of 8.4%.

- ICR scores were comparable between protocols when imaging the 20 cm EPT, 12 cm OFS and 30 cm EPT, 15 cm OFS configurations, and lower when using the ‘Very Low’ protocol to image the 20 cm EPT, 15 cm OFS and 30 cm EPT, 12 cm OFS configurations by an average of 24%.

- Spatial resolution scores were comparable between protocols.
• Total WTR scores were higher when using the ‘Very Low’ protocol to image the 20 cm EPT configurations in which the ‘Very Low’ protocol acquired 17.5% more high density WTR targets than the ‘Low’ protocol and lower when using the ‘Very Low’ protocol to image the 30 cm EPT configurations in which the ‘Very Low’ protocol acquired 30% less high density WTR targets than the ‘Low’ protocol.

• Static MTR scores were comparable between protocols.

• Dynamic MTR scores were comparable between protocols with exception to the 30 cm EPT, 15 cm OFS configuration in which the ‘Very Low’ protocol scored lower than the ‘Low’ protocol by 33.3% .

• Image noise was higher when using the ‘Very Low’ protocol to image the 20 cm EPT, 12 cm OFS and 30 cm EPT, 15 cm OFS configurations, and lower when using the ‘Very Low’ protocol to image the 20 cm EPT, 15 cm OFS and 30 cm EPT, 12 cm OFS configurations by an average of 4.5% .
<table>
<thead>
<tr>
<th>Acquisition Mode 2</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
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<tr>
<td>Very Low : Low</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 20 cm</td>
<td>0.679</td>
<td>1</td>
<td>1</td>
<td>1</td>
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<td></td>
<td></td>
<td></td>
</tr>
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<td>1</td>
<td>0.930</td>
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<tr>
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<td></td>
<td></td>
<td></td>
<td></td>
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<tr>
<td>EPT : 30 cm</td>
<td>0.962</td>
<td>0.700</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.730</td>
<td>0.890</td>
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<td>OFS : 12 cm</td>
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<tr>
<td>EPT : 30 cm</td>
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<td>1</td>
<td>1</td>
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<td>0.670</td>
<td>0.860</td>
<td>1</td>
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<td></td>
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<td></td>
<td></td>
<td></td>
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</tr>
<tr>
<td>Average</td>
<td>0.798</td>
<td>0.880</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.938</td>
<td>0.965</td>
<td>1</td>
<td>0.918</td>
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<td>Maximum Value</td>
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<td>1</td>
<td>1.250</td>
<td>1.080</td>
<td>1</td>
<td>1.040</td>
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<td>1</td>
<td>1</td>
<td>1</td>
<td>0.670</td>
<td>0.860</td>
<td>1</td>
<td>0.670</td>
</tr>
<tr>
<td>Average for</td>
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<td>1</td>
<td>1</td>
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<td>1.055</td>
<td>1</td>
<td>0.985</td>
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<tr>
<td>Average for</td>
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<td>0.850</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.700</td>
<td>0.875</td>
<td>1</td>
<td>0.835</td>
</tr>
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<td>OFS : 15 cm</td>
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<td></td>
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<tr>
<td>Average for</td>
<td>0.820</td>
<td>0.850</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.990</td>
<td>0.985</td>
<td>1</td>
<td>0.835</td>
</tr>
<tr>
<td>EPT : 20 cm</td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
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<td>1</td>
<td>1</td>
<td>0.885</td>
<td>0.945</td>
<td>1</td>
<td>1.010</td>
</tr>
<tr>
<td>EPT : 30 cm</td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table B.10: Ratios of the dose delivery and image quality scores recorded when using the ‘Very Low’ and ‘Low’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.
B.0.12 Comparison of the SHC and EHC high detail acquisition mode dose level 2 protocols: ‘Standard’ and ‘EHC-IQ+’

The ratio of ESD delivery and image quality scores recorded for the ‘Standard’ and ‘EHC-IQ+’ acquisition mode dose level 2 protocols have been presented tabularly in Table B.11. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Standard’ protocol by an average of 16.8%.
- ICR scores were higher when using the ‘Standard’ protocol by an average of 33.3% with exception to the 30 cm EPT, 15 cm OFS configuration where ICR scores were comparable between protocols.
- Spatial resolution scores were comparable between protocols.
- Total WTR scores comparable between protocols.
- Static MTR scores were lower when using the ‘Standard’ protocol by an average of 23.3% with exception to the 20 cm EPT, 15 cm OFS configuration where static MTR scores were comparable between protocols.
- Dynamic MTR scores were comparable between protocols with exception to the 30 cm EPT, 12 cm OFS configuration where dynamic WTR scores were lower when using the ‘Standard’ protocol by an average of 25%.
- Image noise was lower when using the ‘Standard’ protocol by an average of 35.8%.
Table B.11: Ratios of the dose delivery and image quality scores recorded when using the 'Standard' and 'EHC-IQ+' acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.
B.0.13 Comparison of the SHC and EHC low dose rate acquisition mode dose level 2 protocols: ‘Low’ and ‘EHC-RDLS’

The ratio of ESD delivery and image quality scores recorded for the ‘Low’ and ‘EHC-RDLS’ acquisition mode dose level 2 protocols have been presented tabularly in Table B.12. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was higher when using the ‘Low’ protocol by an average of 37.3% with exception to the 30 cm EPT, 12 cm configuration where ESD delivery was lower when using the ‘Low’ protocol by 9%.

- ICR scores were higher when using the ‘Low’ protocol by an average of 26.3% with exception to the 30 cm EPT, 15 cm OFS configuration in which the ‘Low’ protocol scored lower than the ‘EHC-RDLS’ protocol by 18%.

- Spatial resolution scores were comparable between protocols with exception to the 20 cm EPT, 12 cm OFS configuration in which the ‘Low’ protocol scored higher than the ‘EHC-RDLS’ protocol by 14.3%.
• Total WTR scores were comparable between protocols with exception to the 20 cm EPT, 12 cm OFS configuration in which the ‘Low’ protocol acquired twice as many low density WTR targets as the ‘EHC-RDLS’ protocol.

• Static MTR scores were comparable between protocols with exception to the 30 cm EPT, 12 cm OFS configuration in which the ‘Low’ protocol scored lower than the ‘EHC-RDLS’ protocol by 25.0%.

• Dynamic MTR scores were comparable between protocols.

• Image noise was lower when using the ‘Low’ protocol by an average of 36.5%.
### Table B.12: Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘EHC-RDLS’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.

<table>
<thead>
<tr>
<th>Acquisition Mode 2 Low : EHC-RDLS</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>EPT : 20 cm OFS : 12 cm</td>
<td>1.270</td>
<td>1.130</td>
<td>1.140</td>
<td>1</td>
<td>2</td>
<td>1.200</td>
<td>1</td>
<td>1</td>
<td>0.650</td>
</tr>
<tr>
<td>EPT : 20 cm OFS : 15 cm</td>
<td>1.840</td>
<td>1.550</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.640</td>
</tr>
<tr>
<td>EPT : 30 cm OFS : 12 cm</td>
<td>0.910</td>
<td>1.110</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.750</td>
<td>1</td>
<td>0.590</td>
</tr>
<tr>
<td>EPT : 30 cm OFS : 15 cm</td>
<td>1.010</td>
<td>0.820</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.630</td>
</tr>
<tr>
<td>Average</td>
<td>1.258</td>
<td>1.153</td>
<td>1.035</td>
<td>1</td>
<td>1.250</td>
<td>1.050</td>
<td>0.938</td>
<td>1</td>
<td>0.628</td>
</tr>
<tr>
<td>Maximum Value</td>
<td>1.840</td>
<td>1.550</td>
<td>1.140</td>
<td>1</td>
<td>2</td>
<td>1.200</td>
<td>1</td>
<td>1</td>
<td>0.650</td>
</tr>
<tr>
<td>Minimum Value</td>
<td>0.910</td>
<td>1.110</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.750</td>
<td>1</td>
<td>0.590</td>
</tr>
<tr>
<td>Average for EPT : 20 cm</td>
<td>1.340</td>
<td>1.070</td>
<td>1</td>
<td>1</td>
<td>1.500</td>
<td>1.100</td>
<td>1</td>
<td>1</td>
<td>0.645</td>
</tr>
<tr>
<td>Average for EPT : 30 cm</td>
<td>1.090</td>
<td>1.070</td>
<td>1</td>
<td>1</td>
<td>1.500</td>
<td>1.100</td>
<td>0.875</td>
<td>1</td>
<td>0.620</td>
</tr>
<tr>
<td>Average for OFS : 12 cm</td>
<td>1.555</td>
<td>1.340</td>
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<td>1</td>
<td>1.500</td>
<td>1.100</td>
<td>1</td>
<td>1</td>
<td>0.650</td>
</tr>
<tr>
<td>Average for OFS : 15 cm</td>
<td>0.960</td>
<td>0.965</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.875</td>
<td>1</td>
<td>0.610</td>
</tr>
</tbody>
</table>

- Protocol B Performed Best
- Protocol B Performed Worst

**Bold and Italicised Formatting**

- Most Improved EPT/FOV Configuration
B.0.14 Comparison of the post-upgrade fluoroscopy mode dose level 1 protocols: ‘Standard’ and ‘CoroPlus’

The ratio of ESD delivery and image quality scores recorded for the ‘Standard’ and ‘CoroPlus’ fluoroscopy mode dose level 1 protocols have been presented tabularly in Table B.13. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Standard’ protocol by an average of 61.6%.

- ICR scores were lower when using the ‘Standard’ protocol by an average of 52.4% with exception to the 30 cm EPT, 15 cm OFS configuration where both protocols scored comparably.

- Spatial resolution was lower when using the ‘Standard’ protocol by an average of 17.4%.

- Total WTR scores were comparable between protocols with exception to the 30 cm EPT, 15 cm OFS configuration in which the ‘Standard’ protocol acquired 33.3% more high density WTR targets than the ‘CoroPlus’ protocol.
• Static MTR scores were comparable between protocols with exception to the 20 cm EPT, 15 cm OFS configuration in which the ‘Standard’ protocol scored lower than the ‘CoroPlus’ protocol by 25.0%.

• Dynamic MTR scores were comparable between protocols with exception to the 30 cm EPT, 12 cm OFS configuration in which the ‘Standard’ protocol scored higher than the ‘CoroPlus’ protocol by 100.0%.

• Image noise was higher when using the ‘Standard’ protocol by an average of 24.0% with exception to the 30 cm EPT, 12 cm OFS configuration where image noise was lower by 0.6%.
<table>
<thead>
<tr>
<th>Fluoroscopy Mode 1 Standard : CoroPlus</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>EPT : 20 cm OFS : 12 cm</td>
<td>0.399</td>
<td>0.500</td>
<td>0.806</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
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<tr>
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<td>0.502</td>
<td>0.429</td>
<td>0.710</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.750</td>
<td>1</td>
<td>1.191</td>
</tr>
<tr>
<td>EPT : 30 cm OFS : 12 cm</td>
<td>0.314</td>
<td>0.500</td>
<td>0.880</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>2</td>
<td>0.994</td>
</tr>
<tr>
<td>EPT : 30 cm OFS : 15 cm</td>
<td>0.319</td>
<td>0.909</td>
<td>1</td>
<td>1</td>
<td>1.333</td>
<td>1.143</td>
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<td>2</td>
<td>1.292</td>
</tr>
<tr>
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<td>1</td>
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<td>1.036</td>
<td>0.938</td>
<td>1.250</td>
<td>1.179</td>
</tr>
<tr>
<td>Maximum Value</td>
<td>0.502</td>
<td>1</td>
<td>0.909</td>
<td>1</td>
<td>1.333</td>
<td>1.143</td>
<td>1</td>
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<td>1.292</td>
</tr>
<tr>
<td>Minimum Value</td>
<td>0.314</td>
<td>0.429</td>
<td>0.710</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.750</td>
<td>1</td>
<td>0.994</td>
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<tr>
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<td>1</td>
<td>1.500</td>
<td>1.116</td>
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<td>1.072</td>
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</table>

Table B.13: Ratios of the dose delivery and image quality scores recorded when using the ‘Standard’ and ‘CoroPlus’ and fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.
B.0.15 Comparison of the post-upgrade fluoroscopy mode dose level 1 protocols: ‘Low’ and ‘RDLS’

The ratio of ESD delivery and image quality scores recorded for the ‘Low’ and ‘RDLS’ fluoroscopy mode dose level 1 protocols have been presented tabularly in Table B.14. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was higher when using the ‘Low’ protocol to image the 20 cm EPT configurations by an average of 249.4% and higher when using the ‘Low’ protocol to image the 30 cm EPT configurations by an average of 4.4%.

- ICR scores were lower when using the ‘Low’ protocol to image the 20 cm EPT configurations by an average of 26.8% and higher when using the ‘Low’ protocol to image the 30 cm EPT configurations by an average of 100.0%.

- Spatial resolution scores were lower when using the ‘Low’ protocol to image the 20 cm EPT configurations by an average of 20.4%, were higher when using the ‘Low’ protocol to image the 30 cm EPT, 12 cm OFS configuration by 11.1% and were comparable between protocols when imaging the 30 cm EPT, 15 cm OFS configuration.
- Total WTR scores were comparable between protocols.

- Static MTR scores were comparable between protocols with exception to the 30 cm EPT, 12 cm OFS configuration in which the ‘Low’ protocol scored higher than the ‘RDLS’ protocol by 100.0%.

- Dynamic MTR scores were higher when using the ‘Low’ protocol to image the 12 cm OFS configurations by an average of 75.0% and were comparable when imaging the 15 cm OFS configurations.

- Image noise was lower when using the ‘Low’ protocol to image the 12 cm OFS configurations by an average of 8.9% and higher when using the ‘Low’ protocol to image the 30 cm EPT configurations by an average of 25.4%.
<table>
<thead>
<tr>
<th>Fluoroscopy Mode 1 Low : RDLS</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>EPT : 20 cm</td>
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<td>0.937</td>
</tr>
<tr>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 20 cm</td>
<td>4.337</td>
<td>0.714</td>
<td>0.786</td>
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<td>1</td>
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<td>1</td>
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<td>1.274</td>
</tr>
<tr>
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<td></td>
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<td></td>
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<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 30 cm</td>
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<td>1</td>
<td>1</td>
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</tr>
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<td></td>
</tr>
<tr>
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<td>1</td>
<td>1</td>
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<td></td>
</tr>
<tr>
<td>Average</td>
<td>2.269</td>
<td>1.366</td>
<td>0.926</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.250</td>
<td>1.375</td>
<td>1.083</td>
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<td>4.337</td>
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<td>1.111</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>2</td>
<td>2</td>
<td>1.274</td>
</tr>
<tr>
<td>Minimum Value</td>
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<td>0.786</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.886</td>
</tr>
<tr>
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<td>3.494</td>
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<td>1.500</td>
<td>1.500</td>
<td>1.060</td>
</tr>
<tr>
<td>Average for EPT : 20 cm</td>
<td>1.859</td>
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<td>0.959</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.500</td>
<td>1.750</td>
<td>0.912</td>
</tr>
<tr>
<td>Average for EPT : 30 cm</td>
<td>2.679</td>
<td>1.357</td>
<td>0.893</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.254</td>
</tr>
</tbody>
</table>

Table B.14: Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘RDLS’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.
B.0.16 Comparison of the post-upgrade fluoroscopy mode dose level 1 protocols: ‘Low’ and ‘Standard’

The ratio of ESD delivery and image quality scores recorded for the ‘Low’ and ‘Standard’ fluoroscopy mode dose level 1 protocols have been presented tabularly in Table B.15. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was higher when using the ‘Low’ protocol by an average of 52.1%.

- ICR scores were comparable with exception to the 20 cm EPT, 15 cm OFS configuration where ICR scores were lower when using the ‘Low’ protocol by 16.7%.

- Spatial resolution scores were comparable with exception to the 30 cm EPT, 12 cm OFS configuration where spatial resolution was lower when using the ‘Low’ protocol by 9.1%.

- Total WTR scores were comparable between protocols with exception to the 30 cm EPT, 15 cm OFS configuration where the ‘Low’ protocol acquired 25.0% less high density WTR targets than the ‘Low’ protocol acquired.
• Static MTR scores were comparable between protocols when using the 20 cm EPT, 12 cm OFS configuration, were lower when using the ‘Low’ protocol to image the 15 cm OFS configurations by an average of 41.7% and were higher when using the ‘Low’ protocol to image the 30 cm EPT, 12 cm OFS configuration by 100.0%.

• Dynamic MTR scores were comparable between protocols with the exception of the 20 cm EPT, 15 cm OFS configuration where dynamic MTR scores were lower when using the ‘Low’ protocol by 33.3%.

• Image noise was lower when using the ‘Low’ protocol by an average of 16.4%.
<table>
<thead>
<tr>
<th>Fluoroscopy Mode 1</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>Low : Standard</td>
<td></td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 20 cm</td>
<td>1.422</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>0.719</td>
</tr>
<tr>
<td>OFS : 12 cm</td>
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<tr>
<td>EPT : 20 cm</td>
<td>1.654</td>
<td>0.833</td>
<td>0.833</td>
<td>0.833</td>
<td>0.833</td>
<td>0.833</td>
<td>0.667</td>
<td>0.667</td>
<td>0.948</td>
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<tr>
<td>OFS : 15 cm</td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 30 cm</td>
<td>1.517</td>
<td>1.000</td>
<td>0.909</td>
<td>0.909</td>
<td>0.909</td>
<td>0.909</td>
<td>1.000</td>
<td>1.000</td>
<td>0.895</td>
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<td>OFS : 12 cm</td>
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<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 30 cm</td>
<td>1.491</td>
<td>1.000</td>
<td>1.000</td>
<td>0.750</td>
<td>0.750</td>
<td>0.750</td>
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<td>0.500</td>
<td>0.783</td>
</tr>
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<td>OFS : 15 cm</td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Average</td>
<td>1.521</td>
<td>0.958</td>
<td>0.977</td>
<td>0.977</td>
<td>0.977</td>
<td>0.977</td>
<td>1.042</td>
<td>1.042</td>
<td>0.836</td>
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<tr>
<td>Maximum Value</td>
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<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>2.000</td>
<td>2.000</td>
<td>0.948</td>
</tr>
<tr>
<td>Minimum Value</td>
<td>1.422</td>
<td>0.833</td>
<td>0.909</td>
<td>0.750</td>
<td>0.750</td>
<td>0.750</td>
<td>0.500</td>
<td>0.500</td>
<td>0.719</td>
</tr>
<tr>
<td>Average for OFS : 12 cm</td>
<td>1.538</td>
<td>0.917</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>0.834</td>
<td>0.834</td>
<td>0.834</td>
</tr>
<tr>
<td>Average for OFS : 15 cm</td>
<td>1.504</td>
<td>1.000</td>
<td>0.955</td>
<td>0.875</td>
<td>0.875</td>
<td>0.875</td>
<td>1.250</td>
<td>1.250</td>
<td>0.839</td>
</tr>
<tr>
<td>Average for EPT : 20 cm</td>
<td>1.470</td>
<td>1.000</td>
<td>0.955</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.500</td>
<td>1.500</td>
<td>0.807</td>
</tr>
<tr>
<td>Average for EPT : 30 cm</td>
<td>1.573</td>
<td>0.917</td>
<td>1.000</td>
<td>0.875</td>
<td>0.875</td>
<td>0.875</td>
<td>0.584</td>
<td>0.584</td>
<td>0.866</td>
</tr>
</tbody>
</table>

Table B.15: Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘Standard’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.
B.0.17 Comparison of the post-upgrade fluoroscopy mode dose level 1 protocols: ‘Low (15 FPS)’ and ‘Low’

The ratio of ESD delivery and image quality scores recorded for the ‘Low (15 FPS)’ and ‘Low’ fluoroscopy mode dose level 1 protocols have been presented tabularly in Table B.16. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was higher when using the ‘Low’ protocol operated in the 15 FPS setting to image the 12 cm OFS configurations by an average of 112.1% and higher when using the ‘Low’ protocol operated in the 15 FPS setting to image the 15 cm OFS configurations by an average of 97.0%.

- ICR scores were higher when using the ‘Low’ protocol operated in the 15 FPS setting by an average of 82.2% with exception to the 20 cm EPT, 12 cm OFS configuration where both protocols scored comparably.

- Spatial resolution scores were comparable when imaging the 20 cm EPT configurations and higher when using the ‘Low’ protocol operated in the 15 FPS setting when imaging the 30 cm EPT configurations by an average of 10.0% .
- Total WTR scores were comparable between protocols with exception to the 20 cm EPT, 15 cm OFS configuration where the ‘Low’ protocol operated in the 15 FPS setting acquired 25.0% less low density WTR targets.

- Static MTR scores were comparable between protocols with exception to the 30 cm EPT, 15 cm OFS configuration in which the ‘Low’ protocol operated in the 15 FPS setting acquired twice as many MTR targets.

- Dynamic MTR scores were comparable between protocols when imaging the 12 cm OFS configurations and higher when using the ‘Low’ protocol operated in the 15 FPS setting by an average of 75.0%.

- Image noise was higher when using the ‘Low’ protocol operated in the 15 FPS setting by an average of 18.2%.
<table>
<thead>
<tr>
<th>Fluoroscopy Mode 1 Low (15FPS) : Low</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>EPT : 20 cm OFS : 12 cm</td>
<td>2.125</td>
<td>1.667</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.121</td>
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<tr>
<td>EPT : 20 cm OFS : 15 cm</td>
<td>1.979</td>
<td>1.800</td>
<td>1</td>
<td>0.750</td>
<td>1</td>
<td>0.833</td>
<td>1</td>
<td>1.500</td>
<td>1.091</td>
</tr>
<tr>
<td>EPT : 30 cm OFS : 12 cm</td>
<td>2.116</td>
<td>1</td>
<td>1.100</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.294</td>
</tr>
<tr>
<td>EPT : 30 cm OFS : 15 cm</td>
<td>1.961</td>
<td>2</td>
<td>1.100</td>
<td>1</td>
<td>1</td>
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<td>2</td>
<td>2</td>
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<td>2.045</td>
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<td>0.938</td>
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<td>0.958</td>
<td>1.250</td>
<td>1.375</td>
<td>1.182</td>
</tr>
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</tr>
<tr>
<td>Minimum Value</td>
<td>1.961</td>
<td>1</td>
<td>1</td>
<td>0.750</td>
<td>1</td>
<td>0.833</td>
<td>1</td>
<td>1</td>
<td>1.091</td>
</tr>
<tr>
<td>Average for OFS : 12 cm</td>
<td>2.052</td>
<td>1.734</td>
<td>1</td>
<td>0.875</td>
<td>1</td>
<td>0.917</td>
<td>1</td>
<td>1.250</td>
<td><strong>1.106</strong></td>
</tr>
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<td>1.500</td>
<td><strong>1.100</strong></td>
<td><strong>1</strong></td>
<td><strong>1</strong></td>
<td><strong>1</strong></td>
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<td><strong>1.500</strong></td>
<td>1.257</td>
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<td>Average for EPT : 20 cm</td>
<td>2.121</td>
<td>1.334</td>
<td>1.050</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
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<td>1.208</td>
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<td>Average for EPT : 30 cm</td>
<td><strong>1.970</strong></td>
<td><strong>1.900</strong></td>
<td>1.050</td>
<td>0.875</td>
<td>1</td>
<td>0.917</td>
<td><strong>1.500</strong></td>
<td><strong>1.750</strong></td>
<td><strong>1.156</strong></td>
</tr>
</tbody>
</table>

Table B.16: Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘Low (15 FPS)’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.
B.0.18 Comparison of the post-upgrade fluoroscopy mode dose level 1 protocols: ‘Very Low’ and ‘Low’

The ratio of ESD delivery and image quality scores recorded for the ‘Very Low’ and ‘Low’ fluoroscopy mode dose level 1 protocols have been presented tabularly in Table B.17. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Very Low’ protocol by an average of 53.0%.

- ICR was lower when using the ‘Very Low’ protocol to image the 20 cm EPT configurations by an average of 28.4% and lower when using the ‘Very Low’ protocol to image the 30 cm EPT configurations where no iodine dots were resolved in the acquired ‘Very Low’ protocol images.

- Spatial resolution scores were comparable between protocols when imaging the 12 cm OFS configurations and lower when using the ‘Very Low’ protocol to image the 15 cm OFS configurations by an average of 9.6%.

- Total WTR scores were comparable between protocols.

- Static MTR scores were comparable between protocols with exception to the 30 cm EPT, 12 cm OFS configuration where the ‘Very Low’ protocol acquired half as many motion targets as the ‘Low’ protocol.

- Dynamic MTR scores were lower when using the ‘Very Low’ protocol to image the 12 cm OFS configurations by an average of 58.4% and were comparable between protocols when imaging the 15 cm OFS configurations.

- Image noise was lower when using the ‘Very Low’ protocol by an average of 5.6%.
<table>
<thead>
<tr>
<th>Flurwoscopy Mode 1</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
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<td></td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 20 cm</td>
<td>0.462</td>
<td>0.833</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.667</td>
<td>0.984</td>
</tr>
<tr>
<td>OFS : 12 cm</td>
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</tr>
<tr>
<td>EPT : 20 cm</td>
<td>0.464</td>
<td>0.600</td>
<td>0.909</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>OFS : 15 cm</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 30 cm</td>
<td>0.477</td>
<td>0</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.500</td>
<td>0.500</td>
<td>0.917</td>
</tr>
<tr>
<td>OFS : 12 cm</td>
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<td></td>
<td></td>
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</tr>
<tr>
<td>EPT : 30 cm</td>
<td>0.477</td>
<td>0</td>
<td>0.900</td>
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<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
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<td>0.358</td>
<td>0.952</td>
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<td>1</td>
<td>1</td>
<td>0.875</td>
<td>0.792</td>
<td>0.945</td>
</tr>
<tr>
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<td>0.477</td>
<td>0.833</td>
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<td>0</td>
<td>0.900</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.500</td>
<td>0.500</td>
<td>0.906</td>
</tr>
<tr>
<td>Average for OFS : 12 cm</td>
<td>0.463</td>
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<td>0.955</td>
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<td>1</td>
<td>1</td>
<td>1</td>
<td>0.834</td>
<td>0.978</td>
</tr>
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<td>0.950</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.750</td>
<td>0.750</td>
<td>0.912</td>
</tr>
<tr>
<td>Average for EPT : 20 cm</td>
<td>0.470</td>
<td>0.417</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.750</td>
<td>0.584</td>
<td>0.951</td>
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<td>0.300</td>
<td>0.905</td>
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<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.939</td>
</tr>
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</table>

Table B.17: Ratios of the dose delivery and image quality scores recorded when using the ‘Very Low’ and ‘Low’ fluoroscopy mode protocols in the dose level 1 setting for all phantom and imaging configurations.
B.0.19 Comparison of the post-upgrade acquisition mode dose level 2 protocols: ‘Low’ and ‘Standard’

The ratio of ESD delivery and image quality scores recorded for the ‘Low’ and ‘Standard’ fluoroscopy mode dose level 2 protocols have been presented tabularly in Table B.18. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was lower when using the ‘Low’ protocol by an average of 3.8% with exception to the 30 cm EPT, 12 cm OFS configuration where ESD delivery was higher when using the ‘Low’ protocol by 0.7%.

- ICR scores were higher when using the ‘Low’ protocol by an average of 114.4% with exception to the 20 cm EPT, 15 cm OFS configuration where ICR scores were 40% lower when using the ‘Low’ protocol.

- Spatial resolution scores were comparable between protocols with exception to the 20 cm EPT, 12 cm OFS configuration where spatial resolution scores were 10.7% lower when using the ‘Low’ protocol.

- Total WTR scores were comparable between protocols with exception to the 30 cm EPT, 20 cm OFS configuration where 25.0% less high density WTR targets were acquired by the ‘Low’ protocol.

- Static MTR scores were comparable between protocols.

- Dynamic MTR scores were comparable between protocols with exception to the 20 cm EPT, 12 cm OFS configuration where the ‘Low’ protocol acquired 25% less motion targets than the ‘Low’ protocol.

- Image noise was lower when using the ‘Low’ protocol by an average of 13.9%.
<table>
<thead>
<tr>
<th>Fluoroscopy Mode 2</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>Low : Standard</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 20 cm</td>
<td>0.948</td>
<td>1.100</td>
<td>0.893</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.750</td>
<td>0.859</td>
</tr>
<tr>
<td>OFS : 12 cm</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 20 cm</td>
<td>0.969</td>
<td>0.600</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.915</td>
</tr>
<tr>
<td>OFS : 15 cm</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>EPT : 30 cm</td>
<td>1.007</td>
<td>1.330</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.778</td>
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<tr>
<td>OFS : 12 cm</td>
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<td></td>
<td></td>
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<tr>
<td>EPT : 30 cm</td>
<td>0.969</td>
<td>4</td>
<td>1</td>
<td>1</td>
<td>0.750</td>
<td>0.875</td>
<td>1</td>
<td>1</td>
<td>0.892</td>
</tr>
<tr>
<td>OFS : 15 cm</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
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<td></td>
<td></td>
</tr>
<tr>
<td>Average</td>
<td>0.973</td>
<td>1.758</td>
<td>0.973</td>
<td>1</td>
<td>0.938</td>
<td>0.969</td>
<td>1</td>
<td>0.938</td>
<td>0.861</td>
</tr>
<tr>
<td>Maximum Value</td>
<td>1.007</td>
<td>4</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.915</td>
</tr>
<tr>
<td>Minimum Value</td>
<td>0.948</td>
<td>0.600</td>
<td>0.893</td>
<td>1</td>
<td>0.750</td>
<td>0.875</td>
<td>1</td>
<td>0.750</td>
<td>0.778</td>
</tr>
<tr>
<td>Average for OFS : 12 cm</td>
<td>0.959</td>
<td>0.850</td>
<td>0.947</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.875</td>
<td>0.887</td>
</tr>
<tr>
<td>Average for OFS : 15 cm</td>
<td>0.988</td>
<td>2.667</td>
<td>1</td>
<td>1</td>
<td>0.875</td>
<td>0.938</td>
<td>1</td>
<td>1</td>
<td>0.835</td>
</tr>
<tr>
<td>Average for EPT : 20 cm</td>
<td>0.978</td>
<td>1.217</td>
<td>0.947</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>0.875</td>
<td>0.819</td>
</tr>
<tr>
<td>Average for EPT : 30 cm</td>
<td>0.969</td>
<td>2.300</td>
<td>1</td>
<td>1</td>
<td>0.875</td>
<td>0.938</td>
<td>1</td>
<td>1</td>
<td>0.904</td>
</tr>
</tbody>
</table>

Table B.18: Ratios of the dose delivery and image quality scores recorded when using the ‘Low’ and ‘Standard’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.
B.0.20 Comparison of the post-upgrade acquisition mode dose level 2 protocols: ‘Low (15 FPS)’ and ‘Low’

The ratio of ESD delivery and image quality scores recorded for the ‘Low (15 FPS)’ and ‘Low’ fluoroscopy mode dose level 2 protocols have been presented tabularly in Table B.19. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was higher when using the ‘Low’ protocol operated in the 15 FPS setting to image the 12 cm OFS configurations by an average of 109.9% and higher when using the ‘Low’ protocol operated in the 15 FPS setting to image the 15 cm OFS configurations by an average of 101.1%.

- ICR scores were higher when using the ‘Low’ protocol operated in the 15 FPS setting to image the 20 cm EPT, 12 cm OFS phantom configuration by 18.2%, were higher when using the ‘Low’ protocol operated in the 15 FPS setting to image the 15 cm OFS phantom configurations by an average of 58.4% and were lower when using the ‘Low’ protocol operated in the 15 FPS setting to image the 30 cm EPT, 12 cm OFS phantom configuration by 25%.

- Spatial resolution scores were higher when using the ‘Low’ protocol operated in the 15 FPS setting by an average of 11.9% with exception to the 20 cm EPT, 15 cm OFS configuration where spatial resolution was comparable between protocols.
- Total WTR scores were comparable between protocols.
- Static MTR scores were comparable between protocols.
- Dynamic MTR scores were comparable between protocols.
- Image noise was higher when using the ‘Low’ protocol operated in the 15 FPS setting to image the 20 cm EPT configurations by an average of 5.25% and was higher when using the ‘Low’ protocol operated in the 15 FPS setting to image the 30 cm EPT configurations by an average of 28.7%.
<table>
<thead>
<tr>
<th>Fluoroscopy Mode 2 Low (15FPS) : Low</th>
<th>ESD</th>
<th>ICR</th>
<th>Spatial Resolution</th>
<th>Upper WTR</th>
<th>Lower WTR</th>
<th>Total WTR</th>
<th>Static MTR</th>
<th>Dynamic MTR</th>
<th>Image Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>EPT : 20 cm OFS : 12 cm</td>
<td>2.099</td>
<td>1.182</td>
<td>1.120</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.097</td>
</tr>
<tr>
<td>EPT : 20 cm OFS : 15 cm</td>
<td>2.012</td>
<td>1.667</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.008</td>
</tr>
<tr>
<td>EPT : 30 cm OFS : 12 cm</td>
<td>2.099</td>
<td>0.750</td>
<td>1.136</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.311</td>
</tr>
<tr>
<td>EPT : 30 cm OFS : 15 cm</td>
<td>2.010</td>
<td>1.500</td>
<td>1.100</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.262</td>
</tr>
<tr>
<td>Average</td>
<td>2.055</td>
<td>1.275</td>
<td>1.089</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1.170</td>
</tr>
</tbody>
</table>

**Maximum Value**

| EPT : 20 cm OFS : 12 cm             | 2.099 | 1.182 | 1.120             | 1         | 1         | 1         | 1          | 1           | 1.311       |
| EPT : 20 cm OFS : 15 cm             | 2.010 | 1.667 | 1                 | 1         | 1         | 1         | 1          | 1           | 1.262       |
| Average                             | 2.055 | 1.275 | 1.089             | 1         | 1         | 1         | 1          | 1           | 1.170       |

**Minimum Value**

| EPT : 20 cm OFS : 12 cm             | 2.056 | 1.425 | 1.060             | 1         | 1         | 1         | 1          | 1           | 1.053       |
| EPT : 20 cm OFS : 15 cm             | 2.055 | 1.125 | 1.118             | 1         | 1         | 1         | 1          | 1           | 1.287       |
| Average                             | 2.099 | 0.966 | 1.128             | 1         | 1         | 1         | 1          | 1           | 1.204       |
| Average for EPT : 20 cm             | 2.011 | 1.584 | 1.050             | 1         | 1         | 1         | 1          | 1           | 1.135       |

**Protocol B Performed Best**

**Protocol B Performed Worst**

**Bold and Italicised Formatting**

**Table B.19**: Ratios of the dose delivery and image quality scores recorded when using the ‘Low (15 FPS)’ and ‘Low’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.
B.0.21 Comparison of the post-upgrade acquisition mode dose level 2 protocols: ‘Very Low’ and ‘Low’

The ratio of ESD delivery and image quality scores recorded for the ‘Very Low’ and ‘Low’ fluoroscopy mode dose level 2 protocols have been presented tabularly in Table B.20. The following comparative statements can be drawn regarding these two protocols:

- ESD delivery was higher when using the ‘Very Low’ protocol by an average of 9.7%.

- ICR scores were lower when using the ‘Very Low’ protocol to image the 12 cm OFS configurations by an average of 51.2%, comparable when imaging the 20 cm EPT, 15 cm OFS configuration and lower when using the ‘Very Low’ protocol to image the 30 cm EPT 15 cm OFS configuration where no iodine dots were resolved in the acquired ‘Very Low’ protocol images.

- Spatial resolution was comparable between protocols with the exception of the 30 cm EPT, 12 cm OFS configuration where spatial resolution was lower when using the ‘Very Low’ protocol by 9.1%.

- Total WTR was comparable between protocols.

- Static MTR was comparable between protocols with exception to the 20 cm EPT, 12 cm OFS configuration where the ‘Very Low’ protocol acquired 25% less motion targets than the ‘Low’ protocol.

- Dynamic MTR was comparable between protocols.

- Image noise was lower when using the ‘Very Low’ protocol by an average of 6.7% with exception to the 30 cm EPT, 12 cm OFS configuration where image noise was higher when using the ‘Very Low’ protocol by 9.6%.
### Table B.20

Ratios of the dose delivery and image quality scores recorded when using the ‘Very Low’ and ‘Low’ acquisition mode protocols in the dose level 2 setting for all phantom and imaging configurations.
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