Assessing the dosimetric and geometric accuracy of stereotactic radiosurgery

by

Alexis Dimitriadis

Submitted for the degree of
Doctor of Philosophy

Department of Physics
Faculty of Engineering and Physical Sciences
University of Surrey
Guildford, Surrey GU2 7XH, U.K.

December 2016

© Alexis Dimitriadis 2016
Dedication

Dedicated to my parents, my father Philippos and my mother Maro, who made it their life’s purpose to offer me everything. I hope to have inherited a mere fraction of their kindness and generosity, to be able to pay forward the vast amounts of love I received to my own children.

Αγιερωμένο στους γονείς μου, τον πατέρα μου Φίλιππο και την μητέρα μου Μάρω, που έκαναν σκοπό της ζωής τους να μου προσφέρουν τα πάντα. Εύχομαι να έχω κληρονομήσει έστω ένα κλάσμα της χαλάσματος και της γενναιότητάς τους, για να μπορέσω να κληροδοτήσω την απέραντη αγάπη που εισέπραξα στα δικά μου παιδιά.
Abstract

Stereotactic radiosurgery (SRS) is a non-invasive treatment predominantly used for the management of malignant and benign brain tumours. The treatment can be delivered by various platforms in a single fraction where a high dose of radiation is delivered to the target whilst the surrounding healthy tissue is spared. This requires a high degree of accuracy in terms of the dose level delivered but also in terms of geometric precision.

The purpose of this work was to identify the variations of SRS practice in the UK and develop a novel method compatible with all practices, capable of assessing the accuracy of delivery. The motivation behind this effort was to contribute to safety in SRS delivery, provide confidence through a quality assurance audit and form a basis to support standardisation in SRS.

A national survey was performed to investigate SRS practices in the UK and to help guide the methodology of this thesis. This resulted to the development of a method for an end-to-end audit of SRS. This was based on an anthropomorphic head phantom with a medium sized target located centrally in the brain, in close proximity to the brainstem. This realistic patient scenario was presented to all 26 radiosurgery centres in the UK who were asked to treat it with SRS. The dose delivered was assessed using two novel commercially available radiation detectors, a plastic scintillator and radiochromic film. These detectors were characterised for measuring the dose delivered in SRS. Another established dosimetry system, alanine, was also used alongside these detectors to assess the accuracy of each delivery.

The results allowed the assessment of SRS practices in the UK and the comparison of all centres that participated in the audit. The results were also used to evaluate the performance of the dosimeters used for the purposes of quality assurance measurements and audit.

Keywords: Medical physics, Radiotherapy, Radiosurgery, Stereotactic Radiosurgery, Dosimetry, Small fields, Plastic scintillation detector, Radiochromic film, Anthropomorphic phantom, End-to-end, Audit, Linear accelerator, Gamma Knife, CyberKnife.

email: a.dimitriadis@surrey.ac.uk

Statement of originality

This thesis and the work to which it refers are the results of my own efforts. Any ideas, data, images or text resulting from the work of others (whether published or unpublished) are fully identified as such within the work and attributed to their originator in the text, bibliography or in footnotes. This thesis has not been submitted in whole or in part for any other academic degree or professional qualification. I agree that the University has the right to submit my work to the plagiarism detection service TurnitinUK for originality checks. Whether or not drafts have been so-assessed, the University reserves the right to require an electronic version of the final document (as submitted) for assessment as above.

© Alexis Dimitriadis 2016
Acknowledgements

I would like to thank...

Professor Karen Kirkby who gave me the opportunity to embark on this journey and has done everything to make sure that I had all the resources needed to conduct my research. I will be forever thankful for your kindness and support.

Professor Andrew Nisbet who despite his busy schedule has always made time to meet me and discuss my concerns and questions to ensure my research was on track. I cannot thank you enough for your mentorship in my academic and professional progression.

Dr Catharine Clark who has been the champion of this endeavour. Thank you for your insightful scientific remarks that have illuminated my path throughout this voyage. I am eternally grateful to you for showing me tough love when necessary and being gentle when life was hard.

All staff at St. Luke’s Cancer Centre at the Royal Surrey County Hospital who supported my work by allowing access to the facilities and offering a helping hand whenever required. Special thanks to Shakardokht Jafari, Mohammad Hussein, Christopher Bunton, Matt Bolt, Clara Navarro, Chris South, James Earley, Leila Shelley, Mark Long, Matt Jones and Donna Rickard.

Everyone in the Radiation Dosimetry group at the National Physical Laboratory who together create the perfect environment for anyone working there to thrive. Their support has been instrumental in the completion of this work. Many thanks to Russell Thomas, Ileana Silvestre Patallo, Ilias Billas, Simon Duane, Hugo Bouchard, Michael Homer, Anna Subiel, Sebastian Galer, David Maughan, Sam Flynn, Matthew Cashmore, James Manning, Nigel Lee, Julia Snaith, Thorsten Sanders, Graham Bass and Rebecca Nutbrown.

Dr Tony Palmer who guided and supported my constant struggles with film dosimetry. Thank you for the immense help and also for permitting access to the facilities of Queen Alexandra Hospital in Portsmouth. Your contribution was an imperative part of this effort.

The NPL Chemical Dosimetry group whose hard work with alanine has contributed immensely to this work. Thank you to Clare Gouldstone, Gavin Fox, David Crossley and Peter Sharpe.

The Radiotherapy Trials and Quality Assurance team who have provided resources and support that facilitated the completion of this work. Thank you to David Eaton, Jonathan Lee, Rushil Patel, Rada Zotova and Elizabeth Miles.

The engineers of CIRS who customised parts of the phantom to my liking and provided a block of material for free. Thank you to Hunter Gall and Vlad Varchena.

The NPL Engineering Workshop who have been able to machine everything that was asked from them during the course of this project.
Acknowledgements

More to thank...

EPSRC that provided the funds which supported my PhD studies through EP/J500094 EPSRC Centre for Doctoral Training in Application of Next Generation Accelerators.

Also many thanks to RSCH and NPL that provided funds to cover equipment and audit costs and ESTRO for funding my placement to VUmc.

The UK’s SRS physics community who supported this effort from its beginning by finding time in their busy schedules to reply to the survey and offer access to their centres for conducting measurements. Special thanks to Kelvin Hiscoe, Martyn Gilmore, Phill Cooper, Yat Man Tsang and Ian Paddick.

The team of physicists, technicians and radiographers at VUmc, Amsterdam who hosted me at the early stages of my studies and provided me with an insightful understanding of linac-based radiosurgery. Thank you to Johan Cuijpers, Stan Heukelom, Leo Van Battum, Mustafa Zahir, Wenze Van Klink, Omar Bohoudi and Ingrid Kuijper.

The admin and secretarial staff at the University of Surrey that have been always willing to help. Special thanks to Francine Elson-Vining and Karen Arthur.

My bajanagh, fellow-PhD researcher and “partner-in-crime” Kamran Fathi, who has persevered alongside me and stumbled with me towards the end of the tunnel. Knowing that we were on the same boat made things easier. Thank you brother.

My good friends who showed interest in my work and encouraged me to complete it. Thank you being there along the way and offering a helping hand at difficult times: Luka Dimitrov, ChenChen Qiu, Maria Mellou, Pardis Emrouznejad, Andria and Paul Doolan, Rose and Rob Ryan, Emilia and Marcin Baran, Stefanie Kaisi and Nikolas Georgiou.

My beloved siblings, brother-in-law, sister-in-law, nephews and niece: Constandia, Marios, Rafaellos, Mireilina, Evelina, Demetris, Anna, Alexandros, Cassandra, Philippos and Stephanos, who have been a constant source of love and inspiration. Thank you for being there.

My beloved parents, Philippos and Maro, for their love and support throughout my life. Thank you for allowing me to realise my own potential. You are the architects of my success.

My handsome boys, Aris and Hector, who entered into my world halfway through the course of my studies, and changed my life forever. Thank you for the sleep deprivation, the distractions from my work, the vast amounts of love you show me every day and for making me a better man.

Finally, the love of my life, my wife Bruna. Thank you for your understanding, your patience, your support, your love and your extraordinary cooking during this difficult period. You were the force that kept me going at times when I was weak, despite being exhausted yourself. Thank you for pushing me to see this work to completion. I would have never been where I am without you. I love you forever...
Nomenclature

2D - Two-dimensional
3D - Three-dimensional
AAA - Analytical Anisotropic Algorithm
AAPM - The American Association of Physicists in Medicine
ABS - Acrylonitrile Butadiene Styrene
AVM - ArterioVenous Malformation
BED - Biologically Effective Dose
CBCT - Cone Beam Computed Tomography
CCA - Circular Cone Arcs
CCD - Charged-Coupled Device
Co60 - Cobalt-60 radioisotope
CLR - Cherenkov Light Ratio
CPE - Charged Particle Equilibrium
CT - Computed Tomography
CK - CyberKnife
DCA - Dynamic Conformal Arcs
DD - Dose Difference
DTA - Distance-To-Agreement
DMAX - Depth of MAXimum dose
EPR - Electron Paramagnetic Resonance
EPSRC - Engineering & Physical Sciences Research Council
ESTRO - European SocieTy for Radiotherapy and Oncology
FFF - Flattening Filter Free
FWHM - Full Width Half Maximum
GG - Global Gamma
GK - Gamma Knife
IAEA - International Atomic Energy Agency
ICRU - International Commission on Radiation Units and Measurements
ILGKS - International Leksell Gamma Knife Society
IPEM - Institute of Physics and Engineering in Medicine
IMRT - Intensity Modulated RadioTherapy
LB - Linear accelerator-Based
LG - Local Gamma
LINAC - LINear ACcelerator
LSRA - Lateral Scanner Response Artifact
MR - Magnetic Resonance (Imaging)
MC - Monte Carlo (simulations)
MU - Monitor Unit
MLC - Multi-Leaf Collimator
NC - Non-Coplanar
NHS - National Health Service
NPL - National Physical Laboratory
NVBB - Non-Voxel Broad Beam (algorithm)
OAR - Organ At Risk
OD - Optical Density
PDD - Percentage Depth Dose (a normalised depth dose curve)
PSD - Plastic Scintillation Dosimeter
PMMA - PolyMethyl MethAcrylate
PTV - Planning Target Volume
QA - Quality Assurance
RGB - Red Green Blue
RSCH - Royal Surrey County Hospital
SABR - Stereotactic Ablative Body Radiotherapy
SBRT - Stereotactic Body RadioTherapy
SCF - Static Conformal Fields
SDD - Source to Detector Distance
SNR - Signal-to-Noise Ratio
SRS - Stereotactic RadioSurgery
SRT - Stereotactic RadioTherapy
SSD - Source to Surface Distance
TCC - Triple-Channel Correction
TLD - ThermoLuminescence Dosimeter
TMR - Tissue Maximum Ratio
TPS - Treatment Planning System
**TT** - TomoTherapy
**UCLA** - University California Los Angeles
**UK** - United Kingdom
**USA** - United States of America
**VMAT** - Volumetric Modulated Arc Therapy
**WT1** - Water equivalent material
**Z_{eff}** - Effective atomic number
Contents

1 Background and Introduction
   1.1 Historic developments and background .................. 1
       1.1.1 The origin of stereotaxis ....................... 1
       1.1.2 Discovery of ionising radiation ................. 2
       1.1.3 Radiation dosimetry .......................... 2
       1.1.4 The origin of stereotactic radiosurgery ....... 3
       1.1.5 Linear accelerator radiosurgery ................ 4
   1.2 Contemporary stereotactic radiosurgery ................. 6
       1.2.1 Delivery platforms .......................... 6
       1.2.2 Immobilisation devices ....................... 11
       1.2.3 Prescription protocols ....................... 11
   1.3 Quality assurance in stereotactic radiosurgery ....... 14
       1.3.1 Local quality assurance ....................... 14
       1.3.2 External audit ................................ 16
   1.4 Scope of this work .................................. 19
   1.5 Summary of thesis .................................. 19
   1.6 Research Objectives .................................. 20
   1.7 List of publications and presentations arising from this work 21
       1.7.1 Journal Publications .......................... 21
       1.7.2 Published abstracts ........................... 22
       1.7.3 Conference posters/presentations and miscellaneous 22

2 Theory
   2.1 Small radiation fields ................................ 24
       2.1.1 Lack of charged particle equilibrium .......... 25
       2.1.2 Source occlusion & detector size ............... 26
       2.1.3 Considerations for accurate dosimetry in radiosurgery audit 27
   2.2 Small field detectors ................................ 29
   2.3 Alanine dosimetry .................................. 33
       2.3.1 Electron paramagnetic resonance ............... 34
   2.4 Radiochromic film dosimetry ......................... 36
       2.4.1 Triple-channel film dosimetry ................. 37
2.4.2 Scanner-related uncertainties ........................................... 37
2.5 Plastic scintillator dosimetry ........................................... 38
  2.5.1 The Cherenkov noise .................................................... 38
  2.5.2 The chromatic removal method ........................................ 40

3 Current status of stereotactic radiosurgery in the United Kingdom 42
  3.1 Rationale for conducting a survey ....................................... 42
  3.2 SRS Survey - Methods .................................................... 43
  3.3 SRS Survey - Results ..................................................... 44
    3.3.1 Generic Information, Equipment and Experience ............... 44
    3.3.2 Pathologies ........................................................... 47
    3.3.3 Treatment Planning Practices .................................... 49
    3.3.4 Quality assurance and verification ............................... 51
    3.3.5 Immobilisation and Imaging ...................................... 53
  3.4 SRS Survey - Discussion ................................................ 54
    3.4.1 Generic Information, Equipment and Experience ............... 54
    3.4.2 Pathologies ........................................................... 55
    3.4.3 Treatment Planning ................................................. 55
    3.4.4 Quality assurance and verification ............................... 57
    3.4.5 Immobilisation and imaging ...................................... 58
  3.5 SRS Survey - Conclusions .............................................. 59

4 Characterisation of a new commercial radiochromic film 60
  4.1 Gafchromic EBT-XD film ............................................... 60
  4.2 Initial characterisation of EBT-XD - Methods ....................... 62
    4.2.1 Film structure ....................................................... 62
    4.2.2 Calibration ........................................................... 62
    4.2.3 Lateral scanner response artifact ................................ 63
    4.2.4 Dose verification for SRS .......................................... 63
  4.3 Initial characterisation of EBT-XD - Results and Discussion ...... 64
    4.3.1 Film structure ....................................................... 64
    4.3.2 Calibration ........................................................... 65
    4.3.3 Lateral scanner response artifact ................................ 66
    4.3.4 Dose verification for SRS .......................................... 67
  4.4 Initial characterisation of EBT-XD film - Conclusions .......... 71
  4.5 Development of a film dosimetry protocol .......................... 72
    4.5.1 Scanner response ..................................................... 72
    4.5.2 Positioning jig ....................................................... 74
    4.5.3 Scanner resolution ................................................... 74
    4.5.4 Scanner stability .................................................... 75
    4.5.5 Film Energy dependence ............................................. 76
    4.5.6 Post-irradiation darkening ....................................... 76
    4.5.7 Film orientation ..................................................... 77
    4.5.9 Uncertainty budget .................................................. 77
  4.6 Development of a film dosimetry protocol - Conclusions .......... 78
5 Characterisation of a new commercial plastic scintillator 79
  5.1 The Exradin W1 plastic scintillation detector 79
  5.2 Characterisation of the Exradin W1 - Methods 82
    5.2.1 Dose response, collection mode and short-term repeatability 83
    5.2.2 Dose rate and dose-per-pulse 83
    5.2.3 Angular dependence 84
    5.2.4 Temperature dependence 85
    5.2.5 Energy dependence 85
    5.2.6 Long-term stability 86
    5.2.7 Evaluation of detector response in small fields 87
    5.2.8 SuperMAX electrometer characterisation 87
  5.3 Characterisation of the Exradin W1 - Results 88
    5.3.1 Dose response, collection mode and short-term repeatability 88
    5.3.2 Dose rate and dose-per-pulse 88
    5.3.3 Angular response 88
    5.3.4 Temperature dependence 90
    5.3.5 Energy dependence 90
    5.3.6 Long-term stability 92
    5.3.7 Evaluation of detector response in small fields 92
    5.3.8 SuperMAX electrometer characterisation 93
  5.4 Characterisation of the Exradin W1 - Discussion 93
    5.4.1 Uncertainty budget 96
  5.5 Characterisation of the Exradin W1 - Conclusions 97

6 Adaptation and validation of a phantom for radiosurgery audit 98
  6.1 Phantom adaptation - Introduction 98
  6.2 Phantom adaptation - Methods 99
    6.2.1 Positioning accuracy of inserts 103
    6.2.2 Basic plan dose verification with ionisation chamber 104
    6.2.3 Radiochromic film measurements 104
    6.2.4 End-to-end validation test 105
  6.3 Phantom adaptation - Results 106
    6.3.1 Positioning accuracy of inserts 106
    6.3.2 Basic plan dose verification with ionisation chamber 106
    6.3.3 Radiochromic film measurements 107
    6.3.4 End-to-end validation test 108
  6.4 Phantom adaptation - Discussion 110
  6.5 Phantom adaptation - Conclusions 112

7 Methodology of the national audit for stereotactic radiosurgery 113
  7.1 National SRS Audit - Introduction 113
    7.1.1 Initial investigations of a pilot audit study 114
  7.2 National SRS Audit - Methods 115
    7.2.1 Schedule of audits 115
    7.2.2 Audit image set 116
    7.2.3 Immobilisation and CT scan 117
    7.2.4 Image fusion and contouring 118
<table>
<thead>
<tr>
<th>Section</th>
<th>Title</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>7.2.5</td>
<td>Treatment planning</td>
<td>121</td>
</tr>
<tr>
<td>7.2.6</td>
<td>Treatment delivery</td>
<td>122</td>
</tr>
<tr>
<td>7.2.7</td>
<td>Output measurement</td>
<td>122</td>
</tr>
<tr>
<td>7.2.8</td>
<td>EBT-XD film measurements</td>
<td>122</td>
</tr>
<tr>
<td>7.2.9</td>
<td>Exradin W1 measurements</td>
<td>124</td>
</tr>
<tr>
<td>7.2.10</td>
<td>Alanine measurements</td>
<td>125</td>
</tr>
<tr>
<td>8</td>
<td>Results of the national audit for stereotactic radiosurgery</td>
<td>126</td>
</tr>
<tr>
<td>8.1</td>
<td>National SRS Audit - Results</td>
<td>126</td>
</tr>
<tr>
<td>8.1.1</td>
<td>Output measurements</td>
<td>129</td>
</tr>
<tr>
<td>8.1.2</td>
<td>EBT-XD film measurements</td>
<td>129</td>
</tr>
<tr>
<td>8.1.3</td>
<td>Exradin W1 measurements</td>
<td>132</td>
</tr>
<tr>
<td>8.1.4</td>
<td>Alanine measurements</td>
<td>134</td>
</tr>
<tr>
<td>8.2</td>
<td>National SRS Audit - Discussion</td>
<td>137</td>
</tr>
<tr>
<td>8.2.1</td>
<td>Output measurements</td>
<td>137</td>
</tr>
<tr>
<td>8.2.2</td>
<td>EBT-XD film measurements</td>
<td>138</td>
</tr>
<tr>
<td>8.2.3</td>
<td>Exradin W1 measurements</td>
<td>140</td>
</tr>
<tr>
<td>8.2.4</td>
<td>Alanine measurements</td>
<td>141</td>
</tr>
<tr>
<td>8.3</td>
<td>National SRS Audit - Conclusions</td>
<td>143</td>
</tr>
<tr>
<td>9</td>
<td>Conclusions and Further Work</td>
<td>144</td>
</tr>
<tr>
<td>9.1</td>
<td>Conclusions</td>
<td>144</td>
</tr>
<tr>
<td>9.1.1</td>
<td>Dosimetric accuracy</td>
<td>144</td>
</tr>
<tr>
<td>9.1.2</td>
<td>Geometric accuracy</td>
<td>145</td>
</tr>
<tr>
<td>9.1.3</td>
<td>Treatment Plan Quality</td>
<td>145</td>
</tr>
<tr>
<td>9.1.4</td>
<td>Biologically Effective Dose</td>
<td>146</td>
</tr>
<tr>
<td>9.2</td>
<td>Further work</td>
<td>147</td>
</tr>
<tr>
<td>9.2.1</td>
<td>Dosimetric accuracy</td>
<td>147</td>
</tr>
<tr>
<td>9.2.2</td>
<td>Geometric accuracy</td>
<td>147</td>
</tr>
<tr>
<td>9.2.3</td>
<td>Dosimetry systems</td>
<td>147</td>
</tr>
<tr>
<td>9.2.4</td>
<td>Standardisation in SRS</td>
<td>149</td>
</tr>
<tr>
<td>9.3</td>
<td>Closing remarks</td>
<td>151</td>
</tr>
</tbody>
</table>

**Bibliography**

**A** Questions included in the survey questionnaire (Section 3.2) 178

**B** Protocol followed for film handling and scanning 180

**C** Contouring for SRS audit 182

**D** Film analysis results for three centres 183

**E** Example of an Audit Report 184

**F** Published abstracts 192

F.1 Journal publications 192

F.2 Conference presentations 196
List of Figures

1.1 Illustration of a Gamma Knife unit ........................................ 7
1.2 Illustration of a CyberKnife unit .......................................... 8
1.3 Illustration of a Novalis unit ............................................... 9

2.1 Illustration of the lack of charged particle equilibrium in small photon fields ................................................................. 25
2.2 Illustration of source occlusion effect and the impact of detector size .................................................................................. 27
2.3 The light signal from a plastic scintillation detector .......................................................... 39

3.1 The number of SRS centres in the UK ........................................ 45
3.2 The average number of patients treated per month with SRS in the UK .......................................................... 46
3.3 The number of SRS centres treating the indicated pathologies and the expected growth ........................................ 47
3.4 Imaging modalities used in SRS for target definition .................. 49
3.5 Treatment planning systems and algorithms used for SRS planning ....................................................... 50
3.6 The most common prescription isodoses used in each centre ........ 51
3.7 Phantoms and detectors used for Quality Assurance measurements ................................................................................ 52
3.8 The setup accuracies considered acceptable by SRS centres .... 53

4.1 Structural comparison between EBT3 and EBT-XD .................... 61
4.2 Calibration curves for EBT3 and EBT-XD .................................. 65
4.3 Lateral scanner response artifact for EBT3 and EBT-XD ................. 67
4.4 Repeated measurements with EBT-XD film in SRS-type dose distributions ....................................................... 68
4.5 The lateral and longitudinal scanner response artifact of the Epson Expression 10000XL ......................................................... 73
4.6 Photograph of the scanner setup used for film calibration .......... 75
4.7 Epson 10000XL scanner response over 30 minutes of activity .... 76

5.1 Photographs of the components of the Exradin W1 plastic scintillation detector, SuperMAX electrometer and calibration slab ...................... 80
5.2 Angular dependence of the Exradin W1 plastic scintillation detector ........................................................... 89
5.3 Temperature dependence of the Exradin W1 plastic scintillation detector ....................................................... 90
5.4 Experimentally measured and simulated energy dependence of the Cherenkov light ratio ...................................................... 91
5.5 Long term stability of the Exradin W1 plastic scintillation detector .......................................................... 92
5.6 Small field output factors measured for a 10 MV FFF beam using various detectors ........................................................... 93
6.1 The STE2EV anthropomorphic phantom, bespoke inserts and detector sleeves developed ........................................ 99
6.2 CT scan sections of the bespoke inserts developed for STE2EV phantom 100
6.3 Pictures of the all parts of the dosimetry insert ................................. 101
6.4 Schematic representation of the sagittal plane through the middle of the STE2EV phantom showing all detector positions ...................... 102
6.5 Dose distribution comparisons between film and TPS for axial and the sagittal planes .................................................. 109

7.1 Map of the UK showing the geographical locations and equipment groups of the centres visited ..................................... 116
7.2 Immobilisation systems used by different participating centres in the audit 117
7.3 Example of a primary data set CT scan acquired during an audit visit .. 118
7.4 Sagittal section through the middle of the phantom showing copied and repositioned contours ............................... 119
7.5 Sagittal sections of the phantom showing fusion of the primary and secondary CT data sets (a & b) and contours for all structures (c) .... 120
7.6 Example of a audit treatment plan ............................................. 121
7.7 Setup used for scanning audit test film ...................................... 123

8.1 Output measurements in local reference conditions for the 28 platforms that participated in the audit ................................. 129
8.2 Axial and Sagittal film passing rates for 3% - 2 mm Local Gamma ................................. 130
8.3 Axial and Sagittal film passing rates for 5% - 1 mm Global Gamma ................................. 131
8.4 Film-TPS dose difference maps for the axial and sagittal films from Centre 1 .......................................... 132
8.5 Plastic scintillator measurements in the target compared to alanine pellet measurements in the same position within the phantom ................................. 133
8.6 Plastic scintillator measurements in the target compared to the Treatment Planning System predicted dose ................................. 134
8.7 Alanine pellet measurements performed in the target during the audit ................................. 135
8.8 Alanine pellet measurements performed in the organ at risk during the audit .......................................... 136

C.1 Diagram with schematic representation of contour structures ................................. 182

D.1 Gamma passing rates recorded for three centres ..................................... 183

E.1 Example of an Audit report - Page 1 of 8 ...................................... 184
E.2 Example of an Audit report - Page 2 of 8 ...................................... 185
E.3 Example of an Audit report - Page 3 of 8 ...................................... 186
E.4 Example of an Audit report - Page 4 of 8 ...................................... 187
E.5 Example of an Audit report - Page 5 of 8 ...................................... 188
E.6 Example of an Audit report - Page 6 of 8 ...................................... 189
E.7 Example of an Audit report - Page 7 of 8 ...................................... 190
E.8 Example of an Audit report - Page 8 of 8 ...................................... 191
# List of Tables

1.1 Comparison of radiosurgery systems ........................................ 10

2.1 Uncertainty budget calculated for the purposes of this work by the alanine dosimetry service operated at NPL ................................. 34

3.1 The percentage of UK centres treating each pathology with SRS .... 48

4.1 Gamma analysis passing rates for EBT3 and EBT-XD in an SRS dose verification test ............................................................. 69

4.2 Uncertainty budget for Gafchromic EBT-XD film ........................ 78

5.1 Impact of energy-specific correction factors used with the Exradin W1 plastic scintillation detector ............................................ 91

5.2 Uncertainty budget for the Exradin W1 plastic scintillation detector .. 96

6.1 Measurements of a basic plan dose verification using the STE\textsubscript{2}EV phantom and a Semiflex ionisation chamber ......................... 106

6.2 Gamma passing rates for EBT-XD axial film measurements with and without the PSD/alanine abutting the film plane ......................... 107

6.3 Dose measurements in the target and OAR compared to TPS predicted doses ................................................................. 108

8.1 Equipment, techniques and prescription practices of centres that participated in the audit ......................................................... 128
Chapter 1

Background and Introduction

1.1 Historic developments and background

1.1.1 The origin of stereotaxis

The words stereotactic and stereotaxic are both comprised of the Greek words “stereo” which means “solid” or “three-dimensional”, and “taxis” which means “order” or the “orderly arrangement of something”. Therefore, these compound words have been created to elicit the meaning of the orderly arrangement of a three-dimensional solid object, within a three-dimensional space.

The word “stereotaxic” appeared probably for the first time in the writings of Horsley and Clarke in 1906 [1]. It was used to describe the method which they adopted in investigating the cerebellum of a primate. This method employed the use of a metal frame attached on the skull of a primate, which provided external coordinates for precise intracranial navigation. They studied the functions of various anatomical sections of the brain by destroying them. This method formed the basis for many techniques used in research and medicine including computer-aided navigation and surgery, robotic surgery and radiosurgery. We find medical applications in humans of the stereotactic frame technique as early as 1947, with Spiegel et al establishing the less invasive nature of this technique compared to previously used techniques for operations on the human brain [2]. A series of other stereotactic devices were developed in the following years including the device of Lars Leksell, a Swedish neurosurgeon, who published the design
of his own stereotaxic apparatus for intracerebral surgery in 1950 [3]. Leksell’s design fashioned an “arc-quadrant” and an electrode that was inserted in the patient’s brain in order to perform neurosurgical procedures. This design was a major improvement to previous devices that used the Cartesian axis system (x, y, z), as this allowed movement along a polar angle of choice, depth into tissue and anterior-posterior location.

1.1.2 Discovery of ionising radiation

Just over a decade before the invention of the stereotactic frame by Horsley and Clark, Wilhem Röntgen was the first to identify electromagnetic radiation in the range that is today known as X-rays [4]. Within days of Röntgen’s discovery, the biological effects of X-rays were noticed and a few weeks later X-rays were being used for medical treatments in many parts of the world. In the following 50 years many attempts for producing medical radiation delivery machines, with a variety of engineering approaches, were published [5–7]. Around the same time that X-rays were used therapeutically for the first time [8], Henri Becquerel detected the existence of naturally occurring radioactive materials [9]. Likewise, Becquerel’s discovery raised a lot of interest and radioactive sources were just as popular for therapy as X-rays were.

1.1.3 Radiation dosimetry

Radiation dosimetry is the science investigating the measurement of absorbed dose in medical and industrial applications. For the purposes of this thesis, absorbed dose refers to absorbed dose to water, which is used as a standard in medical applications, as the human body is mostly comprised of water and its average density is $1 \, g/cm^3$. The unit of absorbed dose, the Gray (Gy), is fundamentally defined as the absorption of energy per unit mass where $1 \, \text{Gy} = 1 \, \text{Joule/kg}$. National standards laboratories, define dose in this quantity using calorimeters, often referred to as primary standards. Calorimetry quantifies the increase in temperature over absorbed dose. Secondary standard dosimeters, usually ionisation chambers, are calibrated against calorimeters and with the application of suitable factors the recorded quantity can be converted to absolute dose in Gy. Other detectors can be calibrated against secondary standards and used as tertiary standards for absolute dosimetry.
1.1.4 The origin of stereotactic radiosurgery

*Stereotactic radiosurgery* is a non-surgical, non-invasive treatment that focuses ionising radiation to an affected area within the brain, whilst sparing healthy tissue.

The idea of cross-firing small radiation beams to elicit a desired biological effect within the human cranium was proposed in the late 1940’s [10]. Lars Leksell was the first to propose a method that could accurately guide these beams. By putting together the developing fields of stereotaxis and radiotherapy he proposed a new method and coined the term “Stereotactic RadioSurgery” (SRS) to elicit a procedure with geometrical precision equivalent to surgery [11]. These developments took place in 1951 and were based upon the principles of his stereotactic apparatus [3]. He replaced the electrode of his frame with a 200 kVp orthovoltage X-ray unit whilst stating that a beam of higher energy would have been preferable if available. He defined it as: “The administration, through the intact skull, of a single high dose of radiation, stereotactically directed to an intracranial region of interest; May be from X-rays, gamma rays, protons or heavy particle” [11]. As reported by Leksell, the first time this method was used medically, was for the treatment of trigeminal neuralgia in 1953 [12]. Since the conception of this idea, more efforts for delivering radiosurgery were undertaken adopting different ionising radiation sources. The use of heavy particles and protons was growing in popularity as the Bragg-Peak energy deposition of these particles in human tissue was preferred. This was originally suggested in 1946 [13] and it was also then considered ideal for radiosurgery [14, 15]. Leksell also pursued particle SRS by partnering with physicists Borje Larsson and Kurt Liden who contributed immensely [16]. Despite the progress achieved and the clinical success of particle radiosurgery, it appeared that the synchrotron accelerator that produced the beams was unreliable and impractical for clinical use. Not long after that realisation, Larsson’s and Liden’s attempts for radiosurgery were redirected towards a more practical approach which was the use of Cobalt-60 ($^{60}$Co) gamma rays. These efforts resulted in the first Gamma Knife (GK) prototype built for clinical research in 1967. This was a machine with a helmet appliance that contained multiple $^{60}$Co sources. It was dedicated for delivering high doses of radiation into small intracranial volumes and it was predominantly used for
treating functional, behavioural and mobility disorders; all of these are pathologies that were previously treated with conventional neurosurgery. It was also tested for the treatment of malignancies and arteriovenous malformations (AVMs) showing promising results [12,17]. “Gamma Knife I” was installed at Hospital Sofiahemmet in Stockholm, Sweden and produced very successful outcomes in its first few years of operation [16]. It was then gifted to University California Los Angeles (UCLA) in the United States of America (USA) to be used for clinical research for most of the 80’s. A refined prototype was installed at the Karolinska Hospital in 1975, this time containing 179 $^{60}$Co sources whose beams were collimated into ellipsoids instead of squares. This second unit was again used for treating the traditional neurosurgical targets but was also used to investigate further the treatment of cranial tumours and AVMs [18]. Leksell reported that the number of patients treated with SRS by 1983 was approximately 700 [16]. It should be noted that this number does not account for all the patients who received particle radiosurgery at Berkeley and Harvard under the care of physicians Lawrence and Kjellberg. Also, the newly emerging field of linear accelerator (linac) radiosurgery was not accounted for either. After further refinement, a third unit with 201 $^{60}$Co sources was then commercialised and installed in Buenos Aires in the early 80’s. Two identical units, the first radiosurgery machine in the United Kingdom (UK) was commissioned at Weston Park Hospital in Sheffield during the summer of 1985 [19] and the first in the US was installed at the Presbyterian University Hospital of Pittsburgh in 1987 [20].

1.1.5 Linear accelerator radiosurgery

The use of linacs for therapeutic purposes was proposed in the late 1940’s [21], predating Leksell’s proposal for X-ray radiosurgery. Despite that, the progress was slower and the first treatment using a linac was performed in 1953 at Hammersmith Hospital in London using an 8 MeV beam, followed by a 6 MV linac used in Stanford in 1956 [22]. Both of these prototype systems had a fixed beam but with the advent of 360° rotational gantries by Varian in 1960, linacs became very popular radiotherapy tools [23]. They had many advantages over $^{60}$Co radiotherapy machines therefore more centres were replacing their equipment. During this period, SRS was also gaining popularity but
linac systems were unable to provide the geometric and dosimetric accuracy required to perform such procedures. In 1974, Larsson et al stated that should accelerators improve and reach the desired level of accuracy they “would seem a most attractive alternative” to $^{60}$Co systems for radiosurgery [24]. Hence, the 1960’s and the 1970’s was a period where SRS was predominantly used for neurosurgical procedures but it was also starting to invade into the field of radiation oncology. On the other side, linac radiotherapy was recognised as a valuable tool in the fractionated treatment of malignant disease, with a potential future role to play in radiosurgery that had not yet been pursued. Leksell noticed this trend in 1983 by saying that: “…the use of narrow beams of ionising radiation has little to do with radiotherapy in its conventional meaning, but the communication lines between the territories must remain open” [16].

It was not until 1982 that the adaptation of linacs to perform radiosurgery was initiated by Betti and Derechinsky in Argentina using a Varian Clinac [25]. In the same period, collaborations of neurosurgeons and physicists were happening in many countries of the world in order to achieve similar results. The list of pioneers includes Winston and Lutz in Pittsburgh [26], Friedman and Bova in Florida [27], Colombo in Vicenza [28] and Hartmann in Heidelberg [29]. The aforementioned and others proved the feasibility of linac radiosurgery and gave the green light for expanding the world of radiosurgery beyond the GK community. In spite of the evidence, linac-SRS was still considered unsafe by some attributed to the fact that linacs had more moving parts than GK units and therefore more sources of uncertainty [30]. Partly due to this reason, in the early 1990’s efforts were focused on developing dedicated SRS linacs with the first one made by Phillips, the “SRS200” and the second by Varian, the “600SR”. Linac SRS was becoming more popular, however there were still improvements to be made. It was a fact that all linac radiosurgery systems used circular collimators that were not ideal as the target volume was rarely spherical. Leavitt et al identified this issue and suggested a different collimation system [31]. These suggestions were later realised and developed into a commercial micro-Multi Leaf Collimator (MLC) system by the German Cancer Research Centre that significantly improved beam shaping [32]. In order to keep up with these developments Varian Medical Systems and BrainLab partnered to develop a dynamic micro-MLC addition to the Varian dedicated SRS linac.
1.2. Contemporary stereotactic radiosurgery

This collimation system was able to change the shape of the beam every 10° of gantry rotation, a system later named “Novalis” [33]. This partnership was also imperative in giving rise to Intensity Modulated RadioTherapy (IMRT). Neurosurgeon Mark Carol identified the possibility of focusing the beam not just in shaping the beam to match the tumour shape but also in better conforming dose within the tumour itself [34]. Physicist Tim Solberg was convinced by the idea and persuaded BrainLab to include IMRT in their Treatment Planning System (TPS) [33]. The fourth dedicated SRS linac to hit the market followed a new approach to treatment delivery. This system, envisioned by Stanford neurosurgeon John Adler, adopted a compact 6 MV linac attached to a robotic arm and was originally named “Neurotron 1000” to be later renamed “CyberKnife” (CK) [35]. The robotic arm was able to move around the patient to irradiate the target from multiple positions in manner similar to the GK. Moreover, the advent of a robotic arm linac allowed for the first time to take radiosurgical practices outside the cranium [36], something that was later achieved by gantry linacs as well [30].

1.2 Contemporary stereotactic radiosurgery

1.2.1 Delivery platforms

There are currently a few manufacturers that produce delivery platforms marketed specifically for SRS. There are also some commercial systems that are capable of performing SRS but are not necessarily designed or marketed for this purpose. Elekta (Stockholm, Sweden) and Varian (Palo Alto, CA, USA) produce linacs for routine radiotherapy use, which if equipped with additional SRS specific software and hardware, they are capable of delivering SRS. Also, Accuray (Sunnyvale, CA, USA) has a rotational radiotherapy delivery system called “Tomotherapy”, which although it was not marketed as a radiosurgery machine, has been used to deliver SRS.

The most prominent SRS systems in the market (at least in the UK) are the GK (Elekta AB, Stockholm, Sweden), the CK (Accuray, Sunnyvale, California, USA) and the Novalis (Brainlab, Feldkirchen, Germany). In order to highlight the differences in SRS delivery methods a brief description is provided for each machine and a comparison table at the end of the section (Table 1.1). While the images below might not
1.2. Contemporary stereotactic radiosurgery

represent the latest or high-end systems available from their manufacturers, they serve the purpose of illustrating the major components and the architecture of each unit. This section focuses on the most common approaches to radiosurgery practices within the UK.

GK units (Figure 1.1), are equipped with multiple $^{60}$Co sources (the “Perfexion” model has 192) that are positioned in a hemispherical array. Tungsten collimators of different sizes can be positioned between the sources and the patient to shape each “shot” delivered. The patient is fitted and imaged, usually in a Magnetic Resonance Imaging (MRI) scanner, with a stereotactic frame that defines the stereotactic space within which the treatment target is located. For the treatment delivery, the patient is localised by attaching the fitted stereotactic frame onto the GK robotic couch. The couch then moves the patient with respect to the beam isocentre and multiple shots can be used to achieve conformal dose coverage to irregular shaped targets. Previous models did not have any on-board imaging. However, the latest model (GK Icon) has a kV X-ray gantry with Cone-Beam CT (CBCT) capabilities.

Figure 1.1: Illustration of a Gamma knife unit [37].
1.2. Contemporary stereotactic radiosurgery

CKs (Figure 1.2) are comprised of a compact 6MV linac attached to a robotic arm that allows movement in six degrees of freedom. The beam is shaped using circular collimators, but an MLC attachment is also available. The patient is typically scanned and imaged in a thermoplastic mask. Two X-ray imaging sources on the ceiling and two detectors on the floor provide stereoscopic imaging before and throughout a treatment. These images are compared to the reference Computed Tomography (CT) scan and if a deviation is detected the robotic couch moves and repositions the patient.

The Novalis system by Brainlab (Figure 1.3), is the most prominent linac-based SRS system. It is equipped with a micro-MLC that can also have circular collimators attached to it. It comes with the ExacTrac positioning system which employs the principles of stereo-photogrammetry to monitor reflective markers on the patient’s skin using cameras. It also has the capability of performing stereoscopic imaging, before or after

Figure 1.2: Illustration of a CyberKnife unit [38].
each treatment field, with two X-ray sources in the floor and two imaging detectors on
the ceiling, opposite to the CK. It is also equipped with a robotic couch that can move
in six degrees of freedom to correct any deviations from the reference CT scan.

Figure 1.3: Illustration of a Novalis unit [39].

There are other linac platforms, such as the Varian TrueBeam linac, that can be fitted
with the ExacTrac positioning system and therefore utilise favourable features available
of other linac platforms, such as CBCT or different beam energies. To be inclusive of
all the possible variations of equipment in linac-based SRS, Table 1.1 includes a column
for linac-based systems in general.
# Table 1.1: Comparison of the radiosurgery systems.

<table>
<thead>
<tr>
<th>Features</th>
<th>Gamma Knife Perfexion</th>
<th>CyberKnife v.10.5</th>
<th>Linac-based systems</th>
</tr>
</thead>
<tbody>
<tr>
<td>Source</td>
<td>192 $^{60}$Co sources - Gamma rays ($\approx$1.25MeV)</td>
<td>Xrays (6 MV)</td>
<td>X-rays (6 MV/10 MV)</td>
</tr>
<tr>
<td>Flattening filter</td>
<td>N/A</td>
<td>No</td>
<td>Yes/No</td>
</tr>
<tr>
<td>Image guidance</td>
<td>Yes (latest version) - CBCT</td>
<td>Yes - Planar X-rays</td>
<td>Yes - Planar X-rays &amp; CBCT</td>
</tr>
<tr>
<td>Output</td>
<td>$\approx$3.5 Gy/min at installation</td>
<td>up to 1000 MU/min</td>
<td>Up to 2400 MU/min</td>
</tr>
<tr>
<td>Treatment time</td>
<td>Longest</td>
<td>Longer</td>
<td>Shortest</td>
</tr>
<tr>
<td>Daily capacity</td>
<td>Low</td>
<td>Medium</td>
<td>High</td>
</tr>
<tr>
<td>Collimation</td>
<td>Circular</td>
<td>Circular, IRIS &amp; MLC</td>
<td>Circular &amp; MLC</td>
</tr>
<tr>
<td>Delivery Technique</td>
<td>“Spherical” shots</td>
<td>Static fields</td>
<td>Dynamic &amp; Modulated fields</td>
</tr>
<tr>
<td>Target dose homogeneity</td>
<td>Lowest</td>
<td>Higher</td>
<td>Highest</td>
</tr>
<tr>
<td>Dose fall off</td>
<td>Highest</td>
<td>High</td>
<td>Lowest</td>
</tr>
<tr>
<td>Extracranial</td>
<td>No</td>
<td>Yes</td>
<td>Yes</td>
</tr>
<tr>
<td>Immobilisation</td>
<td>Fixed-frame or frameless (latest model)</td>
<td>Frameless</td>
<td>Fixed-frame or frameless</td>
</tr>
<tr>
<td>Fractionation</td>
<td>Yes (on the latest model)</td>
<td>Yes</td>
<td>Yes</td>
</tr>
<tr>
<td>Functional SRS</td>
<td>Ideal</td>
<td>Not ideal</td>
<td>Not ideal</td>
</tr>
<tr>
<td>Versatility</td>
<td>Least</td>
<td>More</td>
<td>Most</td>
</tr>
<tr>
<td>Quality Assurance</td>
<td>Simple</td>
<td>Complex</td>
<td>Complex</td>
</tr>
<tr>
<td>Maintenance</td>
<td>Simple</td>
<td>Rigorous</td>
<td>Rigorous</td>
</tr>
<tr>
<td>Reliability</td>
<td>Good</td>
<td>Variable</td>
<td>Variable</td>
</tr>
<tr>
<td>Source reload</td>
<td>Every 5 years</td>
<td>N/A</td>
<td>N/A</td>
</tr>
</tbody>
</table>
1.2.2 Immobilisation devices

Alongside the evolution of delivery units discussed in sections 1.1.4 and 1.1.5, patient immobilisation systems also evolved. Initially, fixed stereotactic head frames were adapted to become more robust, allow access from more treatment angles but also to improve patient comfort. Moreover, they became compatible with CT and MR imaging in order to allow more precise delineation of the targeted tissue [40]. Nowadays, all major manufacturers provide frameless solutions but some still provide fixed-frame solutions as they may be preferred over frameless system for certain indications. Frameless systems are most commonly employing thermoplastic masks with or without the use of a mouth-bite immobilisation device. Up until recently, GK treatments were only performed with a fixed-frame until Elekta released the “Icon” model in 2015 and thermoplastic masks also became available [41]. The CK robotic radiosurgery system was developed with frameless SRS as its main purpose and although originally a stereotactic frame was used [30], it is now a truly frameless radiosurgery technique. The Novalis system is also compatible with both frameless and a fixed-frame solutions.

1.2.3 Prescription protocols

In recent years, it was observed that fractionating a radiosurgical procedure might be beneficial for improving radiation induced side effects [42]. Typically, some centres prefer to fractionate when the target is in close proximity to an organ at risk or when the target volume is larger than usual. These practices expanded the field of SRS, from only single fraction deliveries, to include treatments that are delivered within two to five fractions. This is sometimes referred to as Stereotactic Radiotherapy (SRT) and when it is delivered extracranially it is called Stereotactic Body Radiotherapy (SBRT) or Stereotactic Ablative Body Radiotherapy (SABR). Although these treatments are similar and they even produce similar results for some pathologies [43], the theory that underpins them is different. On one hand, SRS is used to ablate the target and obliterate it whilst sparing nearby healthy structures. On the other hand, SRT is based on the principles of radiobiology and utilises the differences in radiosensitivity of the target and healthy tissue [44] whilst allowing for repair, redistribution, repopulation and reoxygenation [45]. The two regimes complement each other as certain indications
produced good outcomes with SRS but not SRT and vice versa [43]. Ongoing trials will be able to provide more evidence on the merits and drawbacks of each modality for specific patient groups.

The pathologies treated with the stereotactic techniques in oncology nowadays include acoustic neuromas, arteriovenous malformations in the brain, primary brain tumours and brain metastases, lung cancer, spinal metastases, metastatic liver disease, prostate cancer and others [46–54]. Moreover, apart from malignant and benign lesions, a number of functional disorders may also be treated to produce a desired radiobiological response. These include trigeminal neuralgia, glossopharyngeal neuralgia, epilepsy, essential tremor and Parkinson’s tremor or rigidity and others [46–49,55–57]. The clinical decision making process before prescribing SRS takes into account a number of factors. The lesion has to be of a compatible size, location and type but also the patients’ general health and age are taken into account. Very often it takes months for the effect of treatment to be evident due to the fact that the tumour is not removed but biologically deactivated. SRS is used worldwide to treat large numbers of patients with the aforementioned pathologies each year.

It can be argued that neurosurgeons have been the most influential group of professionals in the development and evolution of radiosurgery. From the moment of the conception of the idea until at least three decades after that, GK SRS was a tool exclusively available to neurosurgeons as it was a treatment option available for most indications previously treated with neurosurgery. The prescription regime followed during this time was focused on ablation of the target, with good target coverage and steep dose fall off, with no interest regarding the homogeneity of dose inside the target. This practice is still existent in present times predominantly in GK centres. Due to the fact that the desired outcome of the treatment is biological deactivation of the target and local control, a steep dose gradient is preferred and dose inhomogeneity is encouraged. Despite this standardised approach and universal approach within the GK community, there are examples of other aspects of prescription practices where consensus has not been achieved. Two of the most influential institutions for GK radiosurgery follow different planning approaches. At the Karolinska Institute, target volumes are outlined and the prescribed dose is delivered to 95% of the target volume. The remaining 5%
of the volume will receive a slightly lower dose than the target. This ideology supports the theory that it is not necessary to treat 100% of the target to the full prescription dose to achieve local control. A different approach is followed at Pittsburgh where target volumes are not outlined at all. The lesion to be treated is covered completely by the prescription isodose and target coverage is inspected visually. Reported results of retrospective outlining of the target volume showed that in most cases 100% of the target was covered [58].

During the 1980’s, when linac-SRS was proving feasible, linacs were ubiquitous in radiotherapy centres. This is also the case today and it is therefore not surprising that linac-based SRS programs are predominantly run by oncologists. In effect, this group of professionals has traditionally followed alternative methods in the use of radiation, which in turn has influenced the way linac-SRS is delivered today. The target volume is always outlined and it is then enlarged by a margin to account for various uncertainties in the delivery. The prescribed isodose typically covers 100% of the enlarged target volume as opposed to GK practice where a lower percentage of the target volume is covered (95%) which has not been enlarged by a margin. Moreover, the prescription isodose chosen is much higher than the prescription isodoses used in GK practices. Whilst GK clinicians prescribe to the 40-60% isodose, linac clinicians prescribe to the 80-100% isodose. Another source of variation is that prescribed isodose is normalised to the maximum dose in GK practices, whereas in linac-SRS it can be a percentage of the dose to the isocentre, which is rarely the maximum dose. Moreover, dose homogeneity within the target volume is always sought after in oncology but as mentioned previously it considered irrelevant in neurological radiosurgery. CK practices tend to be the middle ground between GK and linac-based radiosurgery practices.

In summary, the differences in SRS practices described above show variations in dose prescription, target coverage and homogeneity of dose. This is mainly equipment and clinician dependent, although there are examples of different practices followed in different countries of the world. The practices outlined in the reports of International Commission of Radiation Units and Measurements (ICRU) are hardly followed, especially when it comes to reporting dose to a delivery reference point [59]. This can cause confusion and misunderstanding in communication between SRS practitioners. More-
over, when prescription doses are included in publications they do not convey the detail of the treatments delivered unless additional information is included. It is important to establish ground rules for this communication that will facilitate better understanding and direct comparisons of outcomes. This can in turn open new avenues for research and clinical trials. A strong mechanism in facilitating this process within the UK SRS community could be the initiation of a national, inter-departmental and cross-platform comparison of radiosurgery services.

1.3 Quality assurance in stereotactic radiosurgery

1.3.1 Local quality assurance

SRS demands extraordinary attention to Quality Assurance (QA) issues. This is related to the high geometric and dosimetric accuracy needed to perform a successful procedure, the accuracy demanded by the proximity of the target lesion to organs at risk and the high dose delivered in one or a few sessions. A geographical miss in SRS is not only sparing the problematic target volume but also ablating healthy tissue with a lethal dose of radiation.

The first steps for producing guidelines for radiosurgery were taken in the early 1990’s when many of the pioneers of radiosurgery from both the neurosurgical and oncology communities in the US published a consensus statement manuscript [60]. The major purpose of this paper was to emphasise the necessity for quality improvement, which was deemed achievable by enhancing QA procedures. During the same period, more documents with guidelines were published in an attempt to standardise the quality of SRS worldwide [61, 62]. These efforts mainly focused on the details of how commissioning of a new SRS units should be undertaken giving advice for both linac-based systems and the GK. The reports also highlighted the multi-disciplinary team component that should be apparent in a radiosurgical setting and the need for double-checking all parameters before delivery. There was wide agreement in the fact that apart from the initial testing of the unit, routine testing procedures need to be performed daily, monthly and annually [55]. Apart from these guideline documents, there were a number of studies published outlining QA procedures performed for both GK [63, 64] and
1.3. Quality assurance in stereotactic radiosurgery

linac-SRS [65,66]. The recommended accuracy for the beam delivery system alone was recently reported to be below 1mm [55]. As technology developed and computer capabilities increased, QA procedures necessarily became more rigorous. In particular, the emergence of high-resolution CT and MR has enabled targeting lesions with an overall accuracy of 2.5 - 2.7 mm [67]. QA efforts are now capable of allowing for such accuracy to be assessed and attained routinely in practice. According to the Radiological Physics Centre, “the clinically applicable 95% confidence level of positional uncertainty is 1.8 mm” and 1.6% for the dose delivered. The uncertainty in the dose prescription line is around 1mm [68]. In comparison to a non-stereotactic radiation treatment of a brain lesion, one study found this uncertainty to be up to 7 mm [69]. These figures are indicative of the positional certainty required for SRS delivery. For a radiosurgery program to run confidently it is essential to have good knowledge of the characteristics of all the systems in place. The overall positional accuracy relies on the following [68]:

a) The rigidity of the stereotactic frame system or frameless localisation equipment
b) The pixel and slice separation dimensions of the pre-treatment images
c) The spatial accuracy of the treatment unit; this may be subdivided to the mechanical and geometric accuracy of the unit, the dose delivered to the tumour and surrounding healthy tissue but also the treatment planning system (TPS) that is used
d) The motion of internal anatomy between the imaging and treatment steps.

Patient immobilisation is sometimes dependent on the indication treated. As explained in section 1.2.2, there are a few devices available for stereotactic delivery including removable frames and frameless approaches to enhance fractionated treatments. Whatever the system used, it is recommended that it thoroughly assessed in order to quantify the uncertainty it may contribute to the delivery of SRS [67].

It is important to highlight point b) because tolerances applicable in radiosurgical procedures are such that the slice thickness and resolution chosen for pre-treatment images, if inadequate, can have a significant impact on treatment accuracy. A study showed that target localisation is highly dependent on imaging resolution [40]. In this study a scan of 3 mm slice separation typically contributed to 4.5 mm of target position uncertainty, irrespective of other issues. The overall uncertainty of linac-based
procedures using images of 1 mm slice separation was estimated to be 2.4 mm [68]. Therefore, high resolution pre-treatment imaging is imperative for SRS.

The spatial accuracy of all SRS treatment units should be assessed routinely using established methods [26,70] to maintain the high degree of accuracy required. Moreover, all radiosurgery delivery systems have to be routinely assessed for their radiation output and where possible these measurements should test the ability of the planning system to predict them. If this is not the case, additional tests for the planning system should be undertaken.

Finally, to account for internal motion, approaches vary in a number of ways. Some consider negligible internal motion and proceed to treatment; some utilise imaging technologies only once before the treatment delivery whilst others acquire a number of images throughout the treatment or use tracking systems to continuously adjust treatment delivery. Regardless of the practices used, a thorough understanding of all the factors that can contribute to inaccuracies is of utmost importance. QA checks may then be tailored accordingly and performed routinely to address the consistency of the procedure in order to identify any problems that may contribute negatively.

1.3.2 External audit

Even with rigorous local QA practices, undetected errors may still occur. External audit is a useful mechanism to reveal such errors. Apart from the fact that the word “audit” connotes a performance test that is done by an independent mechanism, the differences between QA and audit are minor. Stamatis defines audit as: “a planned and documented activity to determine by investigation, examination, or evaluation of objective evidence, the adequacy and compliance with established procedures and the effectiveness of an implementation” [71]. In healthcare, the audit process has been associated with procedures that aim to assess patient care against the quality expected. Clinical audit procedures are therefore mechanisms used for reviewing the performance of clinical services. In the instances where the performance status is not satisfactory, the identification of errors acts as constructive feedback to the service providers, followed-up by improvements to the service in order to avoid errors in the future. These procedures were officially introduced in the National Health Service (NHS) in 1993 as a means of
improving the standard of patient care [72].

The field of radiotherapy in the UK was one of the early adopters of independent clinical audit with multi-departmental dosimetric comparisons initiated in 1987 [73], six years before the formalisation of audit by the NHS. As radiotherapy was developing and becoming more advanced, dosimetric audit in the UK was also evolving. Firstly taking on electron radiotherapy [74], then a series of more complex audits performed as part of clinical trials for lung [75], prostate [76] and breast cancer [77], and recently assessing the services offered with the advents of IMRT [78] and rotational radiotherapy [79]. Despite the presence of SRS in the UK since 1985, radiosurgery services have not yet been subjected to an independent, cross-platform dosimetry audit at a national level.

Dosimetry audits in radiotherapy can be categorised into three main areas: basic dosimetry audits, “plan and treat” audits and end-to-end audits. In basic dosimetry audits the auditor performs a series of measurements to compare their results with the local centre. The parameters compared are usually basic beam characteristics such as the output in reference conditions, Percentage Depth-Dose (PDD) curves, field sizes, the quality index of a beam (defined as the Tissue Phantom Ratio (TPR) of measurements at different depths, i.e: TPR_{20cm}^{10cm}, etc. In “plan and treat” audits, the centre prepares a treatment plan, typically calculated on a phantom, and then the auditor performs measurements and compares the measured doses with TPS-predicted doses. The findings give an indication of the overall accuracy of the treatment planning and delivery chain. This type of audit can be combined with some basic dosimetry measurements to perform a more comprehensive test. End-to-end audits are designed to test every step that can contribute to uncertainties in the treatment delivery. This includes patient immobilisation devices, pre-treatment imaging, contouring, image fusion etc. In comparison to other tests, it may be more difficult to evaluate the level of contribution to errors of each step tested, but they are the only types of audit that provide an assessment of the overall accuracy of the pathway assessed. With the addition of more measurements during an end-to-end audit, some sources of uncertainty can be accounted for independently. For example, a basic measurement can evaluate the differences in output at the time of the measurement from the reference conditions which are defined on the TPS for dose calculations. The ratio of the measured over expected
1.3. Quality assurance in stereotactic radiosurgery

Output can then be used as a correction factor. As radiotherapy practices are becoming more complex and the sources contributing to errors are increasing, end-to-end audits are becoming more valuable as they are inclusive of all steps in the patient pathway. Therefore, this type of audit should be the preferred method for a national SRS audit.

With regards to the specific aspects of treatment planning as part of an audit, there are two common approaches. The most common, which is typically used in clinical trials, involves a set of dose constraints fixed by the auditor to be delivered by the service provider to a target volume that contains one or more radiation detectors. This approach was followed for the UK rotational radiotherapy audit as it also served to credential the centres for clinical trials [79]. The other approach does not impose dose constraints and the service provider is asked to deliver the treatment of interest in the same way that it is routinely performed in their centre. This approach was used for the IMRT audit as it was a postal audit designed to give a snapshot of the practice of IMRT in the UK at the time [78]. For both approaches, the target volume tends to be a pre-defined area within a phantom device or a delineated volume in an anonymised patient scan. This exercise is adapted to represent a clinical treatment procedure as much as possible and the audited centre is assessed upon the quality of the treatment delivered. Comparative assessments between all the service providers audited may contribute valuable information as to how each centre compares to the rest of the population.

It would be inappropriate to set up an audit with dose constraints, where the goal is to assess the local practice with different modalities used for SRS, due to the differences in treatment protocols used within the SRS community. As explained in section 1.2.3, there is a multitude of variables in the field of SRS and it would not be representative of clinical practice if centres were asked conform to a prescribed treatment delivery. Considering these discrepancies, the second and more flexible approach is more appropriate; where centres are asked to deliver a treatment following to the local protocol for a known indication. The methodology chosen for an SRS audit has to take into account the idiosyncrasies of the service. It is therefore crucial to understand the nature and complexity of small radiation fields before choosing appropriate methods for measuring them.
1.4 Scope of this work

SRS is field of medicine without standardised practice as over the years it has been subject to influences from neurosurgery and radiation oncology. It is essential to understand the variations existing in SRS practices in order to facilitate a level of standardisation. This will produce a cohesive national approach which would encourage scientific communication and development of the field, facilitate research studies and enable clinical trials to take place. Audit is a strong intervention that supports new clinical implementations whilst identifying errors and providing constructive feedback for eliminating them. The development of a national SRS audit, inclusive of GK, CK and linac-based SRS services is a novel approach that can be the driving force behind this effort. Within the framework of an audit, a number of important research questions can also be tackled. The dosimetric and geometric accuracy of each modality used in SRS can uncover important findings and areas of potential improvement. The same applies to treatment plan quality and TPS calculation algorithm accuracies. Finally, a national audit presents a unique opportunity to access a large number of facilities and test novel instrumentation for the purposes of radiation dosimetry.

1.5 Summary of thesis

- Chapter 2: This chapter briefly describes the properties of the small radiation fields used in radiosurgery and the implications apparent in small-field dosimetry. This is followed by a review of existing detection technology to identify good candidates for SRS dosimetry. The theory of three detection systems is then briefly summarised.

- Chapter 3: This chapter presents the results of a national survey conducted to investigate the current status and practices of radiosurgery in the UK.

- Chapter 4: This chapter presents the characterisation of a new radiochromic film and the development of a protocol suitable for audit purposes.

- Chapter 5: This chapter presents the characterisation of a new plastic scintillation detector and the development of a methodology suitable for audit purposes.
Chapter 6: In this chapter the detectors characterised are utilised in an adapted phantom to develop a procedure for conducting an end-to-end test for stereotactic radiosurgery. The system is validated for this purpose.

Chapter 7: This chapter describes all steps of the methodology developed for performing an end-to-end assessment of all SRS systems active in the UK.

Chapter 8: This chapter presents and discusses the results of the national UK audit for stereotactic radiosurgery.

Chapter 9: The final chapter summarises the findings of this thesis and proposes future directions for dosimetry, audit and standardisation in SRS.

1.6 Research Objectives

a) To conduct a survey in order to investigate current SRS practices in the UK

b) To understand the complexities of performing accurate dosimetry in clinical SRS deliveries and characterise suitable detectors for SRS audit

c) To test novel detector systems for suitability in dosimetry audits

d) To develop and test an end-to-end phantom methodology for SRS audit, compatible with all SRS platforms in the UK

e) To conduct a national UK SRS audit in order to assess the dosimetric and geometric accuracies of SRS practices

f) To identify areas of SRS practices that may benefit from standardisation
1.7 List of publications and presentations arising from this work

1.7.1 Journal Publications


1.7.2 Published abstracts


1.7.3 Conference posters/presentations and miscellaneous

- 1. A. Dimitriadis (2014) “Intracranial stereotactic radiosurgery with the Novalis Tx system”. Report on the TTG placement to attend a 2 weeks at the VUmc, Amsterdam, Netherlands published in the ESTRO newsletter.


Chapter 2

Theory

2.1 Small radiation fields

Radiosurgery targets are typically small in volume. In order to tightly conform a high dose around these targets whilst sparing surrounding healthy tissue, small (smaller than 3 x 3 cm) photon fields are used in all mainstream SRS practises and delivery platforms. GKs and CKs use small static fields, whereas linacs may use static or composite (modulated/dynamic) fields. In all cases, the characteristics of these fields are far removed from reference measurement conditions and are considered non-standard due to the difficulty in measuring or calculating them accurately [80]. For the accurate calculation of dose distributions in a patient, TPSs require a number of accurate measurements in the small fields that will be used clinically. These include output factors, dose profile measurements and percentage depth dose plots. Hence, special considerations are required in these measurements in order produce an accurate beam model on the TPS that can predict resulting dose distributions within the patient with high accuracy.

There are many factors contributing to inaccuracies in small photon field measurements, which have been previously described in great detail [81–83]. For the purposes of this chapter, a brief review of the major contributing factors is included, leading to their relevance and impact in a dosimetry audit for SRS. Finally, a review of current detector technologies has been conducted to identify the best candidates.
2.1.1 Lack of charged particle equilibrium

Charged particle equilibrium (CPE) can be defined in simple terms as the state where the energy carried by the charged particles entering into a region of interest is equal to the energy carried by the particles exiting from that region. For instance, when charged particles that are leaving the volume of interest are replaced by particles entering of the same mean energy, a state of CPE is assumed. When particles leaving the volume are not replaced by particles carrying the same mean energy, charged particle equilibrium cannot be assumed. In photon beams, charged particles refers predominantly to scattered electrons. The lack of lateral electronic equilibrium, which is apparent in small photon fields, is demonstrated in Figure 2.1.

![Figure 2.1: Illustration of the lack of charged particle equilibrium in small photon fields.](image)

As illustrated in Figure 2.1, the forward range of electrons does not have an impact on CPE within the volume of interest, as exiting electrons are replaced by entering electrons moving in the direction of the photon beam. Where the lateral range of the electrons is concerned, if the range exceeds the boundaries of the field edge (from the point of measurement), the field is considered as small. Das et al demonstrate that one of the problems caused by the lack of CPE in small fields is the overestimation of
field sizes [81]. This is also dependent on source occlusion which is discussed further in Section 2.1.2. In large photon beams, where lateral CPE is present, cavity theory assumes that when a detector is inserted into these fields, which have a uniform flux of particles, the measurements performed are accurate as any perturbations caused by the detector are negligible [84]. On the contrary, in small photon fields where there is no lateral CPE, cavity theory is no longer applicable. Moreover, the lateral range of electrons varies with the density of the medium traversed and therefore prolonged electron tracks are produced in areas of low-density, whereas the opposite effect occurs in areas of high-density [83]. Therefore, unless the detector has the same density as the surrounding medium, it will contribute further to the level of disequilibrium. This is an important consideration as detectors with near water-equivalent densities are ideal for SRS dose verification.

On a similar note, density inhomogeneities are often met in SRS where scattered electrons are crossing through dense skull bone and air-filled sinuses in the cranium. As density effects are not always taken into account by TPS algorithms, the use of an anthropomorphic phantom with realistic densities is re-emphasised and the question as to how accurate such algorithms are is raised.

2.1.2 Source occlusion & detector size

With modern radiosurgery collimation systems, occlusion of the direct photon radiation source as seen from the point of measurement is common practice. This results in errors when calculating the field size as the full width half maximum (FWHM) is overestimating the field size. The same problem arises when the detector size, relative to the field size, is bigger than desired causing partial view of the field. Figure 2.2 adapted from IPEM report 103 [82] demonstrates these conditions. These two conditions contribute to what is known as “volume averaging effects” of the detector and lead to underestimation of the dose within the field and overestimation of the penumbra. Consequently, inaccuracies in small field dosimetry can lead to both geometric and dosimetric uncertainties in SRS delivery if inappropriate methodologies are utilised in creating the beam model of the TPS. Therefore, another important consideration for accurate dosimetry in SRS is the size of the detector, or the resolution of the detector
2.1. Small radiation fields

if a 2D dosimetry technique is employed.

Figure 2.2: Illustration of source occlusion effect and the impact of detector size.

2.1.3 Considerations for accurate dosimetry in radiosurgery audit

The effects discussed in Sections 2.1.1 and 2.1.2 are the defining factors of small field dosimetry and big contributors to uncertainty in the characterisation of small fields. Despite these, one can still produce accurate beam models on a TPS if suitable steps are taken to account for these effects. However, there are more considerations to be taken when selecting detectors for dose verification in SRS treatments.

SRS plans are usually delivered in a non-coplanar fashion to achieve high levels of conformity around the target and steep dose fall off from diseased tissue to healthy tissue [85,86]. Therefore a detector whose response is dependent on the angle of irradiation will yield significant errors in the measurement of an SRS plan, which may mask mistakes
2.1. Small radiation fields

in the beam model or the delivery.

SRS plans, particularly those delivered on linacs, utilise variable dose rates to achieve better dose distributions. Therefore, even detectors with dose rate dependence may generate erroneous measurements. Furthermore, the detectors utilised in a multi-platform audit should be compatible with all delivery methods to enable direct comparisons. Inherently, across the population of SRS centres in the UK dose rates will vary due to differences in source activity (GK) or dose-per-pulse (linac & CK).

In recent years, it was noticed that the flattening filter present in linacs, used to flatten the central section of the beam, did not have a significant contribution in small fields and stereotactic applications as the central region of the field is still relatively flat even in the absence of a flattening filter [87]. However, the removal of the flattening filter results in substantially higher dose rates. Flattening Filter Free (FFF) beams were therefore adopted in clinical practise to achieve faster treatments. This requires extra attention in dosimetry as FFF beams have different characteristics to flattened beams.

The use of FFF beams by some SRS centres in the UK re-emphasises the need for a detector that is dose rate independent, due to the higher dose rates. The removal of the flattening filter also causes a spectral change in the beam as low energy contributions are not filtered causing the beam to have a lower average energy and be less penetrating.

The performance of the detectors chosen should therefore be evaluated in FFF beams. The energy dependence of the detector must also be considered due to the large energy range across the population, as shown previously in Table 1.1 (1.25MeV - 10MV). Detectors that are highly energy sensitive over this range of beam energies should be avoided.
2.2 Small field detectors

Section 1.3 highlighted the need for accurate dosimetry in SRS and section 2.1 outlined how inappropriate dosimeters may yield erroneous measurements and result to patient overdose or underdose. Hence, the ideal dosimeter for an SRS audit should have the following characteristics:

a) High spatial resolution (especially lateral) - small size
b) Soft tissue (or water) equivalence
c) Low field size dependence
d) Low photon beam energy dependence
e) Low angle of incidence dependence
f) Low dose rate dependence
g) Linear response in the dose range of interest
h) Perform stable and reproducible measurements

The detectors for which there is documented evidence of their use in SRS include: ionisation chambers, Thermo-Luminescent Dosimeters (TLDs), micro-TLDs, metal oxide semiconductor field-effect transistor (MOSFET) detectors, diamond detectors, silicon diodes, alanine pellets, radiochromic film, polymer gels and plastic scintillators [59, 88–97]. Most of these detectors, approach the ideal only to an extent and only over a limited range of conditions. Many of these dosimeters are dependent on photon energy and beam angle, and the spatial resolution varies. Multi-dimensional high spatial resolution detectors are appealing for SRS measurements as they provide a better assessment of the treatment accuracy and they should be preferred when they are able to achieve low uncertainties. Unfortunately, some dosimeters which may give high spatial resolution (eg: detector arrays) are hard or impossible to use for measurements in anthropomorphic phantoms due to difficulties in accurate positioning and handling. Other detectors with high spatial resolution (eg: TLDs, gels) have significant energy dependence and may require strenuous preparation or calibration [59,98].

The most reliable and well-documented detector in dosimetry is the ion chamber. There are many studies that have assessed the efficacy of small ion chambers in small fields [99–101]. Many inaccuracies were found attributed to volume averaging effects and
their non-tissue equivalence. A comparative study measuring various small field characteristics with radiochromic film, a TLD and a gel dosimeter demonstrated that the three dosimeters were in good agreement with each other and that an ionisation chamber underestimated the absorbed dose [102]. It is widely recognised that ion-chambers, regardless of how small they are or whether they are gas or liquid filled, are not ideal for SRS dosimetry.

Semiconductor diode detectors may have a role to play in SRS dosimetry. Their small size, real-time measurements and superiority to ion chambers are key advantages. On the other hand, their high Z value, temperature-dependence, dose rate dependence and in some cases their finite lifetime are considered major flaws that make them unsuitable for an SRS audit [103].

There are numerous publications investigating the performance of polymer gel dosimetry in stereotactic applications [88, 93, 99, 104–106]. The attention that this field is receiving is indicative of the promising nature of gel dosimetry and its acceptance from the research community as an efficient 3D dosimeter. Unfortunately, there are some serious drawbacks and barriers that are difficult to overcome and it is therefore unlikely that this technique will become mainstream in the near future. The limitations are mostly related with the difficulties in producing gel (consistently), handling and scanning the material for readout after radiation exposure. Considering these factors, it is apparent that despite the many benefits exhibited by gel dosimeters, they will be highly impractical for dosimetry audits.

Diamond detectors are appealing for small field dosimetry due to their near tissue equivalence and small size. It was shown that natural diamonds can be dose rate dependent, but if this dependence is corrected, they produce more accurate measurements than film and diodes [90]. Synthetic diamonds showed very good performance in small field measurements, IMRT and VMAT plans [52, 96, 107, 108]. There are some disadvantages such as the need for pre-irradiation to maintain stability and a degree of angular dependence along the polar axis of the detector [52]. Diamond detectors may have a role to play in SRS dosimetry but high dose rates, treatment angles and the need for pre-irradiation need to be taken into consideration. Bearing in mind the variations in dose rates and angles of irradiation in SRS delivery systems, it may not be
possible to directly compare measurements performed with a diamond detector within the framework of an audit.

Electron paramagnetic resonance (EPR) dosimeters, such as alanine and lithium formate, are promising methods in SRS dosimetry. Lithium formate custom shaped dosimeters were used in an anthropomorphic head phantom and proved the feasibility of this technique for SRS dose verification [109]. As far as alanine is concerned, there is a growing body of evidence for its use in small fields and in dosimetry audit [79, 95, 107, 110–112]. The drawbacks of EPR dosimetry is mainly the non-automatic and expensive readout process, and the compromise between detector size and sensitivity. On the other hand, it is well-established, allows for small size detectors to be manufactured and it is a near water-equivalent detection system with little dependence to any other factors [113]. Another key benefit of alanine is that it can be independent of the ionisation chamber traceability chain.

Some common and practical methods for SRS dosimetry utilise TLDs and radiochromic films to acquire 2D measurements [114], with film being more popular [97, 115, 116] and also used in stacks for acquiring 3D dose distribution measurements [117]. In the USA, there are at least three dosimetry services that offer mailed dosimetry for SRS by evaluating the absorbed dose and/or dose distribution using radiochromic film and TLDs [55]. However, there are limitations with these techniques mostly due to the calibration and post-exposure processing [59]. On the other hand, there are some very attractive advantages to be gained if a good system for handling and processing radiochromic film is established. Film has a high resolution, very little dependence to other factors and has the ability to measure complex geometries, which are attributes that are close to ideal for SRS. Moreover, it is a practical audit dosimeter which is easy to cut and position in phantoms [97, 110, 114, 118, 119], although it is essential that this is done accurately.

Another appealing approach adopts plastic scintillators attached to fibre-optic cables [120–122]. There are a number of advantages in the use of plastic scintillation dosimeters (PSDs), for example the fact that corrections for pressure and humidity are not necessary [123]. They can be very small, they are tissue equivalent and capable of taking real-time measurements. There is currently only one commercially available PSD
2.2. Small field detectors

(Exradin W1 by Standard Imaging, Middleton, WI, USA), which could present a very good option to be used in a dosimetry audit. Initial characterisations of the detector showed mainly a good behaviour [124, 125] but raised some concerns about the long term stability of the detector, the stem effect and its energy dependence [126, 127]. Nevertheless, any dependence shown appears to be correctable, making the detector able to produce low uncertainty measurements. The Exradin W1 has also been tested for the measurement of small field output factors producing good agreement with other suitable small field detectors and Monte Carlo calculated output factors [128–130]. The detector was also used in an Italian multi-centre study for the measurement of output factors, which confirmed its suitability for small field measurements [131]. There is little evidence of PSDs being used for dose verification purposes [132, 133] but the evidence available suggests that the Exradin W1 is a good candidate for an SRS dosimetry audit.

To summarise, there is currently no ideal detector for SRS plan dose verification. Despite the fact that ionisation chambers are the gold standard in radiation detection, it has been shown that they are inadequate for measurements in small fields. There is promise shown by diamond detectors but their dose rate and angular dependence are not ideal for an SRS audit. Likewise, gel dosimeters have distinct advantages but the labour-intensive task of handling and scanning is impractical for an audit. A detector with a good combination of characteristics for an audit is the Exradin W1 PSD. However, as it is not a well-established product yet, if used in an audit its performance should be assessed against a well-established, well-characterised detector. Radiochromic film and alanine require significant attention to the methodologies used and the analysis can be labour intensive. However, there is a large body of evidence which shows that they are suitable detectors for SRS and for dosimetry audits.
2.3 Alanine dosimetry

Alanine is one of the most abundant amino acids in the human body used in the synthesis of proteins. It can also be chemically synthesised and used in radiation dosimetry as it has favourable properties. When alanine is irradiated, it goes through a deamination process and forms stable free radicals as by-products [134]. This effect occurs in both solid and aqueous solutions and has been used for radiotherapy dosimetry for at least two decades [135]. The NPL alanine service, produces pellets made of 90% L-α-alanine and 10% paraffin wax. They are manufactured in a cylindrical shape with 5 mm diameter and 2.5 mm height. They have a density of 1.2 g/cm³ and weight approximately 55 mg [113]. They can be used in stacks and placed in a number of holders for irradiation in plastic or water phantoms. In order to achieve reasonable signal-to-noise ratios (SNR) the service operates a therapy level threshold of doses above 10 Gy only. This poses an implication for routine radiotherapy doses that are normally 2 Gy per fraction. In order to overcome this issue, centres are commonly asked to deliver multiple fractions in order to scale to higher doses and achieve a high SNR with lower uncertainty in the measurement. This does not pose an issue for SRS doses in the target, as prescriptions are usually above 15 Gy, which makes alanine ideal for this application. However, measurements performed outside the target may still suffer from low SNRs. For the purposes of this work, the NPL alanine dosimetry service produced the uncertainty budget shown in Table 2.1, which also accounts for doses in the region of 5 Gy.
<table>
<thead>
<tr>
<th>Source</th>
<th>Standard uncertainty (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dose delivery during calibration in $^{60}$Co</td>
<td>±0.65</td>
</tr>
<tr>
<td>Energy dependence for MV range</td>
<td>±0.35</td>
</tr>
<tr>
<td>Correction for dose received during transport</td>
<td>±0.4 at 5 Gy, ±0.2 at 10 Gy</td>
</tr>
<tr>
<td>Correction for fading</td>
<td>±0.2</td>
</tr>
<tr>
<td>Temperature dependence</td>
<td>±0.3</td>
</tr>
<tr>
<td>Statistical uncertainty in calibration line</td>
<td>±0.4 at 5 Gy, ±0.2 at 10 Gy</td>
</tr>
<tr>
<td>Pellet-to-pellet reproducibility</td>
<td>±1.0 at 5 Gy, ±0.5 at 10 Gy</td>
</tr>
<tr>
<td>Combined</td>
<td>For $^{60}$Co: ±1.9 at 5 Gy, ±1.6 at 10 Gy</td>
</tr>
<tr>
<td></td>
<td>For MV X-rays: ±2.0 at 5 Gy, ±1.7 at 10 Gy</td>
</tr>
</tbody>
</table>

Table 2.1: Uncertainty budget calculated for the purposes of this work by the alanine dosimetry service operated at NPL.

### 2.3.1 Electron paramagnetic resonance

A signal from irradiated alanine pellets can be acquired using Electron Paramagnetic Resonance (EPR) spectroscopy. This method uses a magnetic field to detect free electrons within a sample by exciting their electronic spin. The signal from post-irradiation free radicals in the alanine sample is detected by EPR spectroscopy. The most abundant free radical species produced in alanine after irradiation is CH$_3$-C*H-COOH, which upon scanning with an EPR spectrometer produces characteristic peaks. The method
used at NPL for the analysis of the signal is based on a peak-to-peak evaluation method, which has been proven to be superior to more sophisticated analysis methods [136]. A pellet-specific calibration function can be developed from pellet measurements in a $^{60}$Co irradiator. This function is then used for converting the signal to dose-to-water. The dose response is linear for low doses and saturates above approximately 100 kGy where free radicals start to recombine and the pellets also start to suffer radiation damage. Pellets are typically scanned 48 hours post-irradiation to reduce changes in the EPR spectra to negligible levels. EPR spectroscopy is a non-destructive technique and theoretically pellets could be rescanned if necessary.
2.4 Radiochromic film dosimetry

The term “radiochromic” is reserved for materials that exhibit a change in colour after exposure to radiation. These materials are passive dosimeters and require a readout process. Despite that, the effect of energy absorption from radiation is immediate and therefore additional development processes are not required. Various radiochromic materials have been investigated over the last three decades for their application in medical dosimetry [137]. These efforts led to the development of a number of commercial radiochromic films that were successfully implemented for QA purposes [138]. Their advantages include water-equivalent density, energy independence over the clinical range of MV beams and high-resolution two-dimensional (2D) spatial information [139]. The most widely used radiochromic films for radiotherapy dosimetry are commercially available under the name Gafchromic (Ashland ISP, NJ, USA) and the last two generations were EBT3 and EBT-XD. These films are different to their predecessor EBT2 in that they are symmetric when viewed in cross section, which eliminates the need for a specific orientation in use. Moreover, whilst EBT2 was susceptible to scanning artifacts (Newton’s rings) these have been eliminated with EBT3 and EBT-XD due to their matte finish [140]. The latest films are comprised of an active layer of lithium-10,12-pentacosdiynoate (LiPCDA) micro-crystalline monomers laminated between flexible, clear polyester sheets. When irradiated, these crystal monomers undergo topochemical photo-polymerisation and form rod-shaped micro-polymers that are visible and stable on the polyester base [141]. The level of polymerisation increases with absorbed dose, which also increases colouration of the film. This detection method enables the measurement of complex and inhomogeneous dose distributions with very high spatial resolution. Due to many potential sources of uncertainty in film dosimetry, many studies using radiochromic film only report relative dosimetry measurements [142–145]. This approach assumes that systematic errors occur throughout the film and the measured dose planes are used qualitatively. In this approach, another dosimeter is used to verify the dosimetric accuracy of the delivery. However, many studies have demonstrated that by using advanced practices, absolute film dosimetry is possible [118, 146, 147].
2.4. Radiochromic film dosimetry

2.4.1 Triple-channel film dosimetry

It is now common practice to scan films using conventional (high-end) flat-bed document scanners to acquire digitised images for calibration and analysis. Various mathematical models have been developed in attempts to better explain the relationship between film colouration and dose [139, 143, 148–152].

Most well-established methods involve scanning films in Red-Green-Blue (RGB) format and acquiring 48bit images. Each pixel of the image contains three values, one for each colour channel. These values are then used to calibrate film response to dose. Many studies have demonstrated that multi-channel methods are superior to single-channel methods as they minimise many sources of uncertainty [139, 144, 149–152]. Therefore, the use of multi-channel methods is recommended and should be utilised where possible.

2.4.2 Scanner-related uncertainties

In order to perform accurate film dosimetry, it is essential to appreciate that the film only comprises half of the dosimetry system. Many errors can occur if a poor scanning protocol is adopted. The post-exposure time allowed before scanning films should be considered carefully. Due to post-irradiation darkening of the film, it is recommended to wait for up to 48 hours after exposure before scanning it [140, 148]. However, some methods have been proposed to correct for this effect and allow scanning sooner [153].

A well-known problem with flat-bed scanners is the reduction of light output laterally to the central axis of the scanner [154]. Two recent studies identified the cause of this effect and described additional effects that can contribute to errors [155, 156]. Careful considerations of the scanning resolution, film size (or region of interest), orientation and position on the scanner can diminish these effects to acceptable levels. The effect of film curvature on the scanner bed has been shown to contribute to errors of up to 4% in the dose measured therefore the use of a glass compression plate to keep films flat is recommended [157]. Other sources of error include scanner warm-up times and dirty scanner surfaces [148]. Therefore in order to perform accurate film dosimetry special attention is required for appropriate scanning practices.
2.5 Plastic scintillator dosimetry

The use of PSDs for radiation dosimetry applications has been investigated for over two decades [158]. Their characteristics are very appealing for small field dosimetry due to their near water equivalence (negligible perturbation effects) and high spatial resolution. Early recommendations for SRS dosimetry recognised the ability of these detectors in small fields even at a time when the technology was still developing [32].

The basis of their operation lies in the sensitive volume, which is typically an organic material, doped with scintillating elements. Upon irradiation by a photon beam, scattered electrons interact with the sensitive volume of the detector, depositing energy which excites the electrons in the scintillator material. These excited orbital electrons are then returning to ground state emitting a fast component of light (fluorescence) in the visible spectrum, within a few nanoseconds. There is a slow component of light emitted, phosphorescence, a process that takes place tenths of milliseconds after irradiation [121]. A number of approaches have been used in the detection of light from PSDs. A common approach couples optical fibres to the sensitive volume, which act as the light guide from the scintillator to a photo-detector. The photo-detector could be a photomultiplier tube [159] a photodiode [160] or a Charged-Coupled Device (CCD) camera [161].

2.5.1 The Cherenkov noise

The instrumentation used for light detection in PSDs suffers from Cherenkov radiation emissions “noise” produced in the optic fibre. The light emitted from the scintillator travels through the coupled optic fibre significantly slower than light in vacuum ($c$). The speed of light in the fibre ($v$) is governed by the refractive index ($n$) of the material as $v = c/n$. As the refractive index of the fibre material $n \approx 1.5$, $v \approx 2/3$ $c$. Electrons travelling through the fibre with an energy above $\approx 150$ keV, are travelling faster than $v$. When these conditions are met in a dielectric medium, Cherenkov radiation is released by interaction between the charged particle and the medium. Cherenkov light peaks at a wavelength ($\lambda$) near the violet region of the visible spectrum and its intensity in other regions of the spectrum varies as $\lambda^{-3}$. This light is also transmitted through the
optical fibre and overlaps with scintillation. The optical fibre also produces delayed fluorescence (phosphorescence) like the scintillator. The amount of Cherenkov in the wavelength region of interest is usually relatively low, but can be up to 20% of the signal if a large amount of fibre is irradiated. Therefore, steps need to be taken to remove the Cherenkov contributions from the signal in order to perform accurate dosimetry. Various approaches have been taken in decomposing the signal: the use of another fibre to measure the Cherenkov noise only [158], the optical filtering method to remove the majority of Cherenkov from the signal [162,163], a time-resolved approach to discern between scintillation and Cherenkov by their time-stamp [164] and the “chromatic removal” method [165–168]. The fast nanosecond components of the detected signal are graphically represented in Figure 2.3.

Figure 2.3: The light signal from a plastic scintillation detector.
2.5.2 The chromatic removal method

The chromatic removal method is the most widely used method for calibrating PSDs. This is largely attributed to the fact that currently there is only one commercial system available (Exradin W1, Standard Imaging, Middleton, WI, USA) and the manufacturer recommends this method for its calibration [169]. The calibration method was formulated and validated by Guillot et al for a PSD system that was very similar to the Exradin W1 [168]. The basis of this method is that the amount of Cherenkov produced is proportional to the amount of fibre irradiated. By using dichroic colour filters to split the spectrum into two wavelength bands, it is then possible to perform simultaneous measurements in both bands. These two regions of the spectrum (Figure 2.3) are:

1) a region dominated by light from the scintillator
2) a region dominated by Cherenkov and fluorescence produced from the fibre

The detector is then irradiated in two setup conditions where the only parameter changed is the amount of fibre irradiated. In one setup condition a small amount of fibre is irradiated (min) and in the other a large amount (max). The two simultaneous measurements, in regions 1 and 2, taken in the two setup conditions, max and min, enable the determination of a correction factor that accounts for the removal of the Cherenkov component from the total signal. This correction factor is annotated as the Cherenkov Light Ratio (CLR) and is acquired using equation 2.1.

\[
CLR = \frac{(R_{1\text{max}} - R_{1\text{min}})}{(R_{2\text{max}} - R_{2\text{min}})}
\]  

(2.1)

Where \( R_1 \) and \( R_2 \) correspond to readings in regions 1 and 2, \( R_{1\text{max}} \) and \( R_{2\text{max}} \) are measured simultaneously in the max fibre setup condition, \( R_{1\text{min}} \) and \( R_{2\text{min}} \) are measured simultaneously in the min fibre setup condition.

A cross-calibration factor is also required to convert the signal measured to dose to water. This requires a measurement in reference conditions, with an ionisation chamber traceable to a primary standard. A measurement with the PSD is then taken in the same conditions and the “Gain” factor is then defined using Equation 2.2:

\[
Gain = \frac{Dose_{\text{ref}}}{(R_1 - R_2 \times CLR)}
\]  

(2.2)
Where $Dose_{ref}$ is the dose measured by the ionisation chamber in reference conditions.

Following the determination of these two factors, any subsequent measurements can be used to derive dose using Equation 2.3:

$$Dose = Gain \times (R1 - R2 \times CLR) \quad (2.3)$$

Note that in Equation 2.2 and 2.3 the product of $CLR$ and $R2$ is the Cherenkov over-response. This is then subtracted from the total signal measured in $R1$, and the result is the signal from the scintillator only. In this work, the above method was used for measuring dose with the Exradin W1 PSD.
Chapter 3

Current status of stereotactic radiosurgery in the United Kingdom

3.1 Rationale for conducting a survey

The rapid evolution of SRS has been described in Section 1.1. It is a treatment undergoing constant development that is now established as the preferred route for a large number of indications [170–173]. The UK was one of the early adopters of SRS, as the first GK unit to be installed in the UK (Weston Park Hospital, Sheffield) in 1985 was only the second commercial unit in the world [19]. The first Linac-Based (LB) service in the UK was initiated in 1989 by St. Bartholomew’s Hospital in London [57] and in 2009 CK radiosurgery was introduced for the first time in the UK by the Harley Street clinic [174].

As outlined in Section 1.2, there is a variety of SRS practices which makes it difficult to perform comparisons for assessing the efficacy of this technique, but also causes confusion in science communications. Recently, there have been efforts put into improving standardisation in SRS. The ICRU has proposed a new report on “Prescribing, Recording, and Reporting Stereotactic Treatments with Small Photon Beams” that is in preparation [175]. Also, the International Leksell Gamma Knife Society (ILGKS)
has recently published a report that attempts to standardise the terminology used in radiosurgery. This report deals with the variations in nomenclature and aims to standardise them, not only for GK users but across the field, to facilitate collaborations between radiosurgical technologies [176]. Alongside these international initiatives it is an appropriate time to facilitate collaborations and communication nationally. Moreover, NHS England recently published a consultation for SRS and stated that due to the inability to determine whether one type of SRS machine produces better outcomes there is a need for more robust reviews and technology appraisals on SRS in the UK [177]. The development of a national audit inclusive of all SRS platforms can contribute significantly in facilitating collaborations and communication nationally. Furthermore, it will constitute a robust appraisal to the technologies available for SRS.

The aim of this chapter was to investigate the current status of clinical and dosimetric practice for SRS. It is planned to employ the results to help develop the methodology for an SRS audit, an approach which was previously successfully implemented in a national audit setting [178]. The findings will also provide a useful reference as a basis for benchmarking and future comparisons, but also in assisting new centres to launch their SRS programs. Moreover, a better understanding of the current practices would facilitate better communication in the UK community and support the standardisation of practices between users of different equipment. The results may also provide relevant information for protocol design in clinical trials. The findings of the survey presented below were published in the British Journal of Radiology [179].

### 3.2 SRS Survey - Methods

An online questionnaire (Appendix A) was sent to the Heads of Radiotherapy Physics at 70 UK radiotherapy centres (63 NHS and 7 private) in June 2014. The survey defined “cranial radiosurgery” as “a single high dose of photon radiotherapy in a small volume within the cranium” and requested that participants only submit replies for intracranial radiosurgery. For respondents who did not have plans to implement SRS in the near future only a few questions were required to be answered.

The aim was to identify the centres with active SRS programs and those working to-
wards implementation, in order to account for the logistical implications of undertaking a national audit. The survey also aimed to obtain details of the current issues and variations in clinical practices to ensure compatibility of the methods chosen for the audit with all current practices. The questions were divided into five sections:

1. Generic information and experience
2. Pathologies treated
3. Treatment planning practices
4. QA and verification
5. Immobilisation and imaging

The results reported are presented as fractions/percentages of the centres that responded to each question due to the fact that some partial replies were submitted.

3.3 SRS Survey - Results

3.3.1 Generic Information, Equipment and Experience

68/70 centres responded by December 2014, six months after the launch of the survey. 21/68 centres were performing SRS clinically, 5/68 were in the process of implementing and planning to be clinical within one year. 7/68 were planning to implement within two years and the remaining 35/68 centres did not have active programs nor did they plan to have one in the next two years. Centres were also asked for how long they have been delivering SRS: 13/21 indicated that they have been clinical for more than 5 years, 4/21 for 3-5 years, 3/21 for 1-3 years and 1/21 began treating in the last year. Figure 3.1 shows the diversity and numbers of vendors used in the 21 centres that were treating with SRS in the UK at the end of the survey.

According to the responses, there are 31 radiotherapy treatment machines in total used clinically for SRS in the UK: 16/31 are linacs, 6/31 are CKs, 7/31 are GKs and 2/31 are TomoTherapy (TT).

2/21 centres (both GK) indicated that they have treated over 1000 cases, 5/21 (2 GK, 2 LB and 1 CK) had treated 500-1000 cases, 10/21 (2 GK, 5 LB and 3 CK) had treated 100-500 with the remaining 4/21 (2 CK, 1 LB and 1 TT) having treated less than 100
Figure 3.1: The number of SRS centres in the UK using equipment from each manufacturer indicated.

cases. GK centres have the highest patient throughput per month followed by LB and CK centres as shown on Figure 3.2.

The centres were asked if they wished to expand their current SRS programs; this expansion was differentiated between increasing the indications which they already treat (60% of centres said “Yes”) and the numbers of cases per week (75% of centres said “Yes”). 40% of centres wanted to expand on both areas and 5% had no expansion plans. 11/21 centres did not limit the number of cases treated with the remaining 10 limiting them. The reasons indicated as the limiting factors for limiting/not expanding SRS programs were: resources for delivery (5/10), planning resources (4/10), contouring resources (3/10) and NHS funding (2/10).

All GK centres (6/21) use the GK Perfexion beam array as a collimation system and 4/6 CK centres use circular collimators whereas the remaining 2/6 CK use both circular collimators and the CK IRIS system. 6 LB and 1 TT centre use only MLCs (7/21), 1 LB centre uses only circular collimators (1/21) and 1 LB centre uses both (1/21). The majority of the centres that use MLCs adopt micro-MLCs (2.5 mm) although there are two centres that use wider MLCs (5 mm and 6.25 mm). The CK and LB centres that use more than one collimation systems stated that the collimation system of choice is
Figure 3.2: The average number of patients treated with SRS per month in the UK grouped under three frequencies. The equipment used in each group is also indicated on the chart.

dependent on the pathology, its size, its location and its proximity to organs at risk (OARs).

The nominal photon energies used for SRS delivery are $^{60}$Co (1.25 MeV) used by the 6 GK centres, 6 MV used by 14/21 centres and 10 MV used by 1 centre. Of the 8 LB centres, 1/8 indicated that they use FFF mode (10MV FFF). 3/8 said that they do not use FFF but they plan to within two years whilst 4/8 stated that they have no plans to use FFF within the next two years. GK, CK and TT do not use flattening filters by default. The most common delivery technique is the use of non-coplanar static fields (89% of respondents use it), however modulated fields or arcs with MLCs or cones are also being used, but less often (5-21% of respondents).
3.3.2 Pathologies

The majority of centres are treating solitary and multiple brain metastases but many other sites are also being treated. The centres were asked to state the number of patients treated for each indication per month. They were also asked about their expansion plans in terms of numbers of patients treated and indications treated. It is anticipated that there will be an increase for both as shown in Figure 3.3 and Table 3.1.

Figure 3.3: The number of centres treating the indicated pathologies and expected increase by the end of 2016.

There are differences in the indications treated between equipment groups. GK centres treat the largest variety of clinical indications (mean: 9.2, range: 6-13), followed by CK (mean: 6.5, range: 2-10) and LB (mean: 4.9, range: 1-10). LB is mostly focused on brain metastatic diseases, acoustic neuromas and arteriovenous malformations. The number of indications treated was also found to increase with the experience of the centre, as more experienced centres treat more indications than less experienced centres.
3.3. SRS Survey - Results

<table>
<thead>
<tr>
<th>Indications</th>
<th>Not within 2 years (%)</th>
<th>Start within 2 years (%)</th>
<th>Currently ≤4/month (%)</th>
<th>Currently 5-8/month (%)</th>
<th>Currently ≥9/month (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Solitary brain metastasis</td>
<td>4.8</td>
<td>-</td>
<td>57.1</td>
<td>33.3</td>
<td>4.8</td>
</tr>
<tr>
<td>Multiple brain metastases</td>
<td>4.8</td>
<td>4.8</td>
<td>61.9</td>
<td>28.6</td>
<td>-</td>
</tr>
<tr>
<td>Acoustic neuromas</td>
<td>23.8</td>
<td>4.8</td>
<td>52.4</td>
<td>9.5</td>
<td>9.5</td>
</tr>
<tr>
<td>Meningiomas</td>
<td>28.6</td>
<td>-</td>
<td>61.9</td>
<td>4.8</td>
<td>4.8</td>
</tr>
<tr>
<td>Pituitary adenomas</td>
<td>28.6</td>
<td>4.8</td>
<td>66.7</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Arteriovenous malformations</td>
<td>33.3</td>
<td>14.3</td>
<td>47.6</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Trigeminal Neuralgia</td>
<td>57.1</td>
<td>-</td>
<td>38.1</td>
<td>4.8</td>
<td>-</td>
</tr>
<tr>
<td>Primary CNS tumours</td>
<td>52.4</td>
<td>4.8</td>
<td>42.9</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Glomus jugulare tumours</td>
<td>52.4</td>
<td>9.5</td>
<td>42.9</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Cranio-pharyngiomas</td>
<td>52.4</td>
<td>9.5</td>
<td>38.1</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

Table 3.1: The percentage of centres that treat the indicated anatomical sites. The results are categorised in three frequency groups. The percentage of centres that are and are not planning to to treat within the next 2 years are also presented. The indications have been sorted with the most common indications at the top and least common indications at the top and least common indications at the bottom.
3.3.3 Treatment Planning Practices

Participants were asked to state the imaging modalities they use for SRS target and OAR visualisation and outlining. Multiple answers were allowed as centres often decide to use a different modality depending on the availability of modalities, the equipment used and the pathology to be treated. The results show that there is a wide range of imaging-modalities used with fused CT-MR being the most common (Figure 3.4).

![Imaging modalities used for Outlining](image)

Figure 3.4: The percentage of centres using the indicated imaging modalities for SRS target and OAR definition (multiple answers were allowed)

The structures outlined are dependent on the location of the target volume as only proximal OARs are usually delineated. The target volume is delineated by all centres at all times. The respondents indicated the following structures as those that are most often delineated: optic chiasm (90%), optic nerves (90%), brainstem (86%), eyes (76%), lenses (71%), cochlea (24%), trigeminal nerve (19%), whole brain (14%), hippocampus (10%), lacrimal gland (5%), pituitary (5%), scalp (5%) and temporal lobe (5%). One centre reported that OAR contouring is only performed retrospectively to the plan and is dependent on the clinician’s judgement which OARs are to be contoured after reviewing the dose distribution. Figure 3.5 shows the range of TPSs used. 6/21 respondents reported that they may not use their TPS for delineation and use different software instead.
3.3. SRS Survey - Results

Figure 3.5: The percentage of centres using each Treatment Planning System and Algorithms (multiple answers were allowed).

All GK centres use an MRI data set in the TPS for defining the target and OARs as well as the stereotactic space and were using a tissue maximum ratio (TMR) algorithm, which assumes that everything inside the patient has water density, in order to calculate dose distributions. 2 GK centres indicated that they may use CT for certain pathologies or when the MR distortion is significant. The remaining centres use a CT data set for dose calculation.

GK and CK centres only use non-coplanar static fields and the TT centre only uses coplanar IMRT. There is a variety of techniques used in the 8 linac centres but 7 centres employ non-coplanar techniques only and one centre may also use coplanar techniques. The most common techniques used by linac centres are non-coplanar static fields (5/8) and non-coplanar dynamic conformal arcs (4/8). 2 centres use non-coplanar circular collimator arcs and only one centre uses VMAT.

The centres were asked to indicate the most commonly used prescription isodose (Fig-
3.3. SRS Survey - Results

The reported values range from 45-50% to 95-100% indicating that there are different prescription practices employed by each equipment group. GKS prescribe in the range of 45-55%, CKs within 55-80% and LB with TT within 80-100%.

![Most common prescription isodose used](image)

Figure 3.6: The most common prescription isodoses used in each centre.

### 3.3.4 Quality assurance and verification

Participants were asked to state the quantity of patient-specific QA (plan dose verification) performed. 6/21 reported that they do not perform such measurements routinely (5 GK, 1 CK) and 7/21 perform measurements on every plan (3 LB, 3 CK and 1 TT). 4/21 perform QA measurements for new techniques/sites only (2 LB and 2 CK) and 3/21 perform it as part of a regular QA program (2 LB and 1 GK). 1 LB centre performs QA for less than 10% of all plans. For the centres that had reduced the amount of patient-specific QA measurements made, 87.5% stated that they reduced it after 10-25 plans. 70% these of centres reduced the amount of patient-specific QA following an experience-based decision and the remaining stated that main reason for reducing them was insufficient time (15%) and the lack of a suitable phantom/detector (15%). There is a large range of phantoms and detectors used for QA measurements. These are illustrated in Figure 3.7.
The majority of respondents (13/18) reported that they measure both point doses and dose distributions (4 CK, 5 LB, 3 GK and 1 TT). 4/18 stated that they only measure point doses (2 LB and 2 CK) and one centre only dose distributions (1 LB).
3.3.5 Immobilisation and Imaging

The majority of centres (12/21) use thermoplastic masks for the immobilisation of patients (6 LB and 6 CK). All GKS always use invasive fixed frames (Leksell) to immobilise the patient and 2 LB may use masks or frames depending on the indication treated (one centre stated that frames are always used for AVMs).

Due to the GK Perfexion design, the centres that have this system do not perform any imaging guidance for localisation before or during treatment. The 6 CK centres perform orthogonal kV X-rays before and throughout the treatment. The TT centre performs a pre-treatment MV CT scan and no further imaging during treatment. Between LB practitioners, 3/8 centres acquire a CBCT scan before treatment and do not perform any other imaging during treatment. 4/8 centres, all of which use the Novalis ExacTrac system, start by taking a set of orthogonal kV X-rays to ensure precise patient positioning and repeat these images after each couch movement and before each beam delivery. Only one LB centre does not perform any image guidance before or during treatment but it was reported that they are intending to use CBCT in the near future. Figure 3.8 shows the action level below which setup accuracy is considered acceptable by each centre.

![Action level below which setup accuracy is considered acceptable]

Figure 3.8: The reported values of setup accuracy below which treatment is considered acceptable. The different equipment groups are indicated.
3.4 SRS Survey - Discussion

3.4.1 Generic Information, Equipment and Experience

The use of SRS has been steadily increasing since its introduction in the UK in 1985. By 2009, at least 13 centres were active and in the last five years the number has increased to 21. The results also suggest that this growth will continue as almost half the UK’s centres stated that they were planning to offer cranial radiosurgery services by the end of 2016. The rapid increase over the last few years is mostly attributed to CK and LB radiosurgery, as well as the growing interest in hypofractionated treatment schemes. The survey responses suggest that this expansion will continue to occur both for the pathologies treated and the number of cases per month. In order to support this development and maintain high quality in clinical delivery, a national audit for SRS services is essential.

Since GK units are dedicated to cranial radiosurgery it is not surprising that the results showed them having higher patient throughput. The disadvantage in having a versatile SRS unit that can be used for other techniques is that it is unable to match a dedicated unit in the number of cases treated per month. This is also reflected in the responses from the 5 centres who limit the numbers of patients they treat due to resources for SRS delivery who are all LB.

The only LB centre that currently uses an FFF beam is also the only centre that uses 10 MV. This is perhaps less striking when considering the fact that an FFF beam is softer (has a lower average energy) than a filtered beam of the same nominal energy. As more LB centres switch to FFF beams (4 centres are intending to do so), in order to benefit from faster delivery times, it will be interesting to see if 10 MV beams are adopted by more of these centres. The results indicate that the advent of FFF beam delivery is slowly becoming prominent in the clinic. It is therefore essential that rigorous QA monitors and tests the systems in place. A national radiosurgery audit can contribute significantly in ensuring that FFF dosimetry is accurate, at least in the centres that use such beams for SRS.
3.4.2 Pathologies

The results suggest that the most commonly treated pathologies are brain metastases and acoustic neuromas, with a single brain metastasis being the most common. The treatment of these pathologies is likely to increase, as new centres launching their programs in the near future will probably take on these indications first. A realistic and patient-like audit scenario that allows for all centres to participate should therefore be based on single brain metastasis. This way, even centres that have not yet started performing SRS clinically can participate in the audit to practise on a plausible clinical case and have their SRS pathway assessed in an end-to-end test. Simple geometric phantoms are considered inadequate for performing end-to-end audits as they are not representative of patient-like conditions [97]. Considering the above, an anthropomorphic head phantom should be adopted for the purposes of an audit, with a solitary brain metastasis as a target.

As shown in Section 3.3.2, there is a general intention by centres to increase the number of pathologies treated and the number of patients treated for each pathology. This highlights further the importance of an SRS audit, as this is additional evidence of the growth of SRS and the potential for detriment if rigorous QA is not undertaken.

3.4.3 Treatment Planning

The majority of centres use fused CT and MRI for the purposes of contouring and planning, except GK centres that tend to use MRI only. It would be beneficial to incorporate both CT and MR in an end-to-end audit as any errors that may occur in image fusion can be factored in to the final results. However, the limited visibility of phantoms on MR-scans and the restricted availability of MR-scanning time in most NHS hospitals may prove to be impractical for an end-to-end audit, in which case a CT only methodology may need to be adopted even for GK centres.

A number of studies have looked at the inter-observer variations in target contouring on patient cases and reported large disagreements in the contoured volumes [180,181]. It is widely accepted that delineation may be the weakest link in radiotherapy and radiosurgery practices due to these large variations. The survey results indicate that
there is a multitude of contouring practices currently in place for SRS. This may cause some smaller but significant variations in volume size that are inherent to the scan slice thickness, voxel size and delineation software used. The collection of audit plans will allow for these variations to be quantified.

As shown in Figure 3.5 there are at least 9 different combinations of planning systems and calculation algorithms used for SRS. Some of these perform heterogeneity corrections whereas others do not, and some perform convolution, pencil beam or Monte Carlo (MC) type calculations to predict dose distributions. A dosimetric comparison between measured doses and expected doses can provide valuable information as to which algorithms perform better in SRS. The use of a head phantom with tissue-like electron densities and effective atomic numbers is crucial in revealing any differences between calculation algorithms that may be clinically significant.

The survey responses show a large number of techniques used for SRS with non-coplanar static fields and non-coplanar arcs (including dynamic conformal arcs, VMAT and circular collimator arcs) to be the most common. A few centres reported the use of coplanar VMAT and IMRT. It would be of interest and clinical benefit to establish whether certain techniques produce superior results in a dosimetry comparison. The survey findings could be indicative that some techniques, such as coplanar IMRT or coplanar VMAT, are less common because they produce inferior plans. A retrospective analysis of the plans produced during the audit may provide valuable evidence for recommendations to the use of only certain techniques for SRS. This will create a more cohesive approach to SRS but will also prevent upcoming centres from undertaking timely investigations in assessing the suitability of a range of techniques.

Figure 3.6 illustrates that the most commonly used prescription isodoses differ between equipment groups. The survey replies have not been reported with a percentage of target coverage. Also, the centres were not asked to indicate whether any target margins are being used in these prescriptions. If target coverage and margins are taken into account this distribution may change as these parameters influence the prescription isodose. The variation presented is a demonstration of the differences in prescription practices. In addition to this, it should be mentioned that the level of dose homogeneity in the target volume is also variable. Typically, LB radiosurgery adheres to
ICRU practices and aims for more homogeneous dose distributions within the target volume. On the other hand, some dose inhomogeneity in the target volume is acceptable for CK users and it is actively sought in GK radiosurgery. The large variation in prescription practices requires discussion and perhaps indicates that it will benefit from regulation/standardisation. The proposed end-to-end audit will present the same patient-like scenario to all centres, which will retrospectively invite the opportunity for the variations in prescription practices to be studied. The survey findings are supportive of the argument hypothesised in Section 1.3.2 that performing an audit with dose constrains would be inappropriate and a more flexible approach should be preferred.

### 3.4.4 Quality assurance and verification

The number of patient-specific QA measurements performed by each centre suggests that some centres have more confidence in their method of treatment delivery than others. The predominant reason for reducing the amount of QA was an experience based decision and the reduction was mostly introduced after 10-25 treatment plans. Interestingly, some experienced centres continue to perform QA on every plan whereas some less experienced centres have reduced their QA. Also, the majority of GK centres do not perform measurements routinely, which indicates higher confidence, possibly due to the simpler design of the system with fewer moving parts and the well-known activity related output of the sources.

The detectors and phantoms in use are diverse with the exception of GK users who follow similar practices. Solid water blocks are used by the majority of centres and ionisation chambers are the most commonly used detectors despite their limitations in small field dosimetry capabilities discussed previously (Section 2.2). Some evidence of recognition that these limitations exist is seen as some centres are using other detectors such as Gafchromic film, diodes, diamond and thermoluminescence detectors in their practices. Additional comments provided with the survey submission as well as personal communications with SRS practitioners, stressed the need for the verification of multi-dimensional dose distributions as opposed to point doses. Modern TPSs allow for heavily modulated and highly conformal plans to be simulated on the computer screen at the expense of time and resources. Even though plans of high complexity are often
verified during local QA measurements, there is still lack of confidence in the results as 2D or 3D dosimeters are not always available in the clinic. It is therefore suggested that the audit methodology introduces a target of irregular shape in close proximity to OARs in an attempt to produce a challenging but realistic case. This, in combination with suitable 2D/3D dosimeters can provide valuable findings for SRS centres.

### 3.4.5 Immobilisation and imaging

The results illustrate the wide range of immobilisation devices and image guidance protocols used by the SRS community. It is essential that the audit methodology is compatible with all immobilisation devices used. It should also allow validation of the accuracy of all image guidance systems in place. This re-enforces the significance of an anthropomorphic phantom with realistic tissue densities as it will provide the closest representation possible to patient-like conditions.

As a result of the different immobilisation and imaging systems in place, a range of acceptable set-up accuracy levels are in use, as shown in Figure 3.8. LB and TT centres assess acceptable positional accuracies via the integrated imaging systems but this is not possible for GK centres due to the lack of an imaging system. The answers reported by GK centres on their acceptable positional accuracy are reflections of the accuracy believed to be achievable by their units based on their QA measurements and empirical knowledge. This raises concerns regarding the accuracy of delivery due to the lack of on-board imaging for positional verification and the reduced levels of routine QA discussed in Section 3.4.4. Arguably, the use of a fixed frame in GK SRS eliminates the need for imaging verification, but it is essential to test this in an independent audit. CK centres are also able to assess the setup accuracy via the integrated imaging system; however, there was no consensus on the reported setup accuracy levels. It should be noted that 4 out of 6 CK centres submitted their replies to this question with additional commentary that allows some insights into the reasons behind the large spread of the levels reported. These 4 submissions commented on the nature of the CK system that automatically repositions the patient after each pair of kV images is acquired. The machine therefore “corrects” misalignments above an action level set by the user. However, there are still some inherent positioning errors that need to be taken into account which can be
3.5 SRS Survey - Conclusions

assessed in the audit.

LB centres generally agree and report a level of 1 mm with the exception of one centre, reporting a level of 2 mm. Responses from GK users range between 0.2 – 1 mm. The largest range is seen in CK responses which vary from 0.1 mm to 1.5 mm. The spread of the reported acceptable setup accuracies also indicates benefit from standardisation. With only two centres reporting a setup accuracy above 1 mm, this suggests that a reasonable national level for SRS setup accuracy would be less than 1 mm. Quantitative evidence to support this proposal for standardisation can be acquired in the national SRS audit.

3.5 SRS Survey - Conclusions

SRS in the UK has undergone a rapid increase since its introduction in 1985. This is particularly the case with LB radiosurgery which is now overtaking GK and CK in terms of number of units and patients treated. This increasing trend will probably continue in the coming years, not only for the number of centres delivering SRS but also the number of patients treated and the range of pathologies treated. There is variation in the practices followed between different centres for most aspects of radiosurgery. It is proposed that a national end-to-end audit, compatible with all practices, is conducted to evaluate the dosimetric and geometric accuracy of SRS in the UK. Such an audit should adopt an anthropomorphic phantom with realistic tissue densities and capabilities for multi-dimensional dose verification. The target should be modelled on a single brain metastasis of irregular shape which is in close proximity to an OAR.
Chapter 4

Characterisation of a new commercial radiochromic film

4.1 Gafchromic EBT-XD film

EBT-XD became commercially available to the UK in the Spring of 2015. This film has a slightly different structure to its well-established predecessor EBT3. Both films are composed of two identical polyester surfaces that are 125 $\mu$m thick, with a density of 1.35 g/cm$^3$. In their sensitive layers, between the two polyester sheets, they have different atomic compositions with the new film featuring the addition of 4 new elements and higher density. Figure 4.1 illustrates a schematic representation of the two films showing their atomic compositions and densities. EBT-XD (XD stands for eXtended Dose) was marketed as the film of choice for high-dose radiotherapy applications, such as SRS and SABR. Whereas EBT3’s optimum dose range is 0.1 Gy - 10 Gy, according to the manufacturer EBT-XD can measure doses up to 100 Gy. The size of LiPCDA crystals in the active layer has also been changed. These crystals are needle-shaped particles, which in EBT3 measure 15 $\mu m$ - 20 $\mu m$ in length and 1 $\mu m$ - 2 $\mu m$ in diameter. In EBT-XD they have the same diameter but are a lot shorter in length, measuring 2 $\mu m$ - 4 $\mu m$ [182]. This is a substantial change in aspect ratio, going from approximately 10:1 to 2:1. The re-sized crystals are expected to improve the homogeneity of the active layer distribution, as preferential alignment will be decreased and the particles will be more prone to random Brownian movement before lamination.
4.1. Gafchromic EBT-XD film

As a result, film thickness is expected to show improved uniformity and therefore improved response to dose. Moreover, scanning orientation effects are expected to be minimised. It is also speculated that the lateral scanner response artifact will be less pronounced on EBT-XD due to reduced polarisation. This film therefore is a very appealing dosimeter for use in high dose applications, like SRS. The higher dose range, near water-equivalent structure and ability to perform high-resolution measurements, with potentially improved response to EBT3 are favourable characteristics.

The purpose of this chapter was to evaluate EBT-XD by comparison to EBT3 and assess its performance in SRS dose verification. Another objective was the development of a suitable film dosimetry protocol for the purposes of an SRS audit. The work in this chapter was partly published in the Journal of Physics and Medicine in Biology [183].

---

1 Initial investigations described in Section 4.2 were carried out in collaboration with Antony Palmer at Queen Alexandra Hospital, Portsmouth. Subsequent investigations (Section 4.5) were performed at NPL.
4.2 Initial characterisation of EBT-XD - Methods

4.2.1 Film structure

The information on the atomic composition of the films shown in Figure 4.1 was collected via communication with the manufacturer. Effective atomic numbers \((Z_{eff})\) for the two films were calculated using a power-law method which uses the atomic number of each element in the compound and the proportion of electrons associated with it [184]. Although, it is acknowledged that this method is simplistic, it was considered suitable for comparison of the two dosimeters and for comparing them to water.

4.2.2 Calibration

All measurements were performed with EBT-XD batch no.0108501 and EBT3 batch no.12171303. Ten pieces of film from each batch were used for calibration purposes. Each film was placed in a WT1 phantom at 95 cm SSD, 5 cm deep and irradiated in a 10 cm\(^2\) field with a 6 MV linac beam. The range of doses used for irradiation were from 0 Gy - 40 Gy and the absolute dose was measured using an ionisation chamber with a traceable calibration to NPL. The recommendations of AAPM TG-55 for film handling were followed [138]: latex gloves were worn, films were always kept at a clean and controlled with minimum exposure to ambient light. The scanning methods adopted, were previously evaluated and implemented for brachytherapy dosimetry audit [152]. Films were scanned in consistent orientation, 48 hours after exposure on an Epson Expression 11000XL flat-bed scanner. The scanning resolution used was 72 dots-per-inch (dpi) and a 3 mm thick piece of glass was used to press film flat on the scanner. The scanner was switched on for several minutes and a number of warm-up scans were performed before any film was scanned. The digitised film images were acquired in RGB 48bit TIFF format (16bit per colour channel) using transmission mode without colour or sharpness corrections. The software package FilmQAPro version 5 (Ashland ISP Advanced Materials, NJ, USA) was used for generating calibrations curves. This was done for both films using a conventional single (red) channel method and also using the red channel with triple-channel correction (TCC) [149].
4.2.3 Lateral scanner response artifact

The Lateral Scanner Response Artifact (LSRA) is attributed to a number of effects, but the dominating source is the scanner light polarisation as it passes through the film [156]. This results to a non-uniform lateral (perpendicular) response to the direction of the scan. In order to avoid or minimise errors caused by this effect, films must be placed on the central axis of the scanner. This effect is also known to vary with dose and to be different for each colour channel. Depending on the methods used, large errors could be caused even for small pieces of films that do not extend very far from the central axis of the scanner [185]. To evaluate the LSRA, four strips of film were exposed to doses of 0 Gy, 13 Gy, 20 Gy and 40 Gy. Each strip was then cut into three pieces to create identical samples. One piece was placed on the central axis of the scanner while the other two identical pieces were placed 3 cm laterally to the central piece. Eight scans were acquired with the off-axis films moved successively further away from the central axis until they reached the boundaries of the scanner at 15 cm off-axis. The artifact is quantified in the ratio of the pixel values between the off-axis and on-axis films. This was calculated for all colour channels.

4.2.4 Dose verification for SRS

In order to compare the two types of film in SRS-type measurements, with minimum influence from other factors, a simple 30 cm cubical WT1 phantom was used for this test. 6 cm x 9 cm pieces from each film (test films) were placed in the frontal plane of the phantom at 5 cm deep. The phantom was CT-scanned using 1 mm slice thickness and the images were imported in the Eclipse TPS version 13 (Varian, Palo Alto, CA, USA). A single-arc VMAT plan for a 2.5 cm diameter spherical target was re-calculated on the phantom using the anisotropic analytical algorithm (AAA). The plan was positioned so that the test film bisected the sphere of the high-dose region. Output measurements were performed before and after the film measurements to account for any variations in the delivery. The plan was delivered three times for each film and a 1 mm resolution coronal plane was exported from the TPS for comparison. Film scanning and analysis were performed following the methods described in Section 4.2.2. The film dose-linear scaling method [153] was used to recalibrate the film. This method requires two pieces of
film, one un-irradiated and one irradiated at a known dose, to be scanned together with
the test film. The two reference films act as an additional calibration and are used to re-
scale the test film response map to mitigate any variations caused by post-irradiation
darkening and/or scanner light-source variations. For the purposes of this test, the
reference films used had doses of 0 Gy and 20 Gy (prescription isodose). Test-films
were auto-aligned with the TPS dose map in FilmQAPro to eliminate any positional
errors that may have occurred. Gamma analysis \cite{186} was used to evaluate the level
of agreement between the two dose distributions. This method is used for comparing
two dose distributions and quantifying the percentage of pixels in agreement within
predefined criteria for distance-to-agreement (DTA) and dose difference (DD). Such
tests can be performed to compare pixels to the local value (local gamma) or to a
normalisation value (global gamma), which is typically the maximum dose.

4.3 Initial characterisation of EBT-XD - Results and Discussion

4.3.1 Film structure

In examining the structure of the films, it was noted that the smaller EBT-XD sam-
pies had a substantially larger natural curl than EBT3 samples. This emphasised the
importance of using a glass plate during scanning. The effective atomic number of the
sensitive layers of the two films are 7.26 for EBT3 and 7.37 for EBT-XD. These are
both comparable to the effective atomic number of water. The two films are almost
identical in structure as they both measure approximately 275 \( \mu m \), and the bulk of
them is composed of the same polyester sheets. Small differences are seen in the ef-
fective atomic number of their sensitive layers, and therefore both films are considered
water-equivalent the MV energy range clinically used for SRS. These findings are sup-
ported by experimental evidence from Grams et al who showed that measurements at
the same dose levels with 6 MV and 18 MV gave the same values of OD \cite{185}. 
4.3.2 Calibration

As expected, EBT3 was darker than EBT-XD at all dose levels. This is translated to lower pixel values on all colour channels of the calibration curve. Figure 4.2 shows the calibration curves for the three colour channels of each film over the dose range of 0 Gy - 40 Gy. The curves have similar appearances but they are offset in the y-axis (pixel value). EBT3 has a steeper gradient on all three colour channel curves for doses approximately up to 10 Gy. Above 10 Gy, EBT-XD curves have steeper gradients whereas EBT3 curves become almost flat as the film response comes close to saturation. EBT-XD colour channels appear to have not reached plateaux regions at 40 Gy, the end of the dose range investigated. SRS prescription isodoses tend to be above 15 Gy. Hence, for the doses of interest in SRS dosimetry EBT-XD is expected to be superior.

Figure 4.2: Calibration curves for EBT3 and EBT-XD.
The model used in FilmQAPro for fitting calibration curves is based on the rational function shown in Equation 4.1 [149].

\[ R(D) = \frac{a + bD}{c + D} \] (4.1)

Where \( R(D) \) is the scanner response, in the range of 0 to 1, at dose \( D \), and \( a \), \( b \) and \( c \) are the constant equation parameters that are fitted.

Rational functions, like the one in Equation 4.1, are considered to be better representations of film-dose response. This is due to Optical Density (OD) not showing a linear relationship with dose and therefore showing no benefit in converting transmittance to OD [187]. On the other hand, a corrected transmittance value, using a rational function, shows a linear relationship with dose [188]. A uniformity correction is also used in FilmQAPro based on the paper by Micke et al [149]. This method is capable of separating dose-dependent and dose-independent parts of the signal using the values acquired from the three colour channels. This way any perturbations, such as film thickness variations or noise, can be reduced. The method is more effective when the slopes of each colour channel are sufficiently different to each other. The steeper gradients seen by EBT-XD in this high dose range, are therefore essential in better measurement accuracy. It also benefits from larger differentials between colour channel values to determine the correction. The triple-channel uniformity correction can be applied on any colour channel but the red channel is most commonly used. Since EBT-XD colour channels show bigger separations, it is expected that the uncertainty of the measurement will be lower than that of EBT3.

### 4.3.3 Lateral scanner response artifact

The LSRA of an Epson Expression 11000XL scanner was evaluated for both films, in all three colour channels and at different dose levels. The relative change in pixel value for different off-axis displacements is shown in Figure 4.3. An additional dose level of 13 Gy was chosen for the evaluation of EBT3, as it had the same optical density with EBT-XD at the 40 Gy level.

As shown in Figure 4.3, the artifact is more pronounced at higher dose levels. It is expressed as a decrease in the pixel value and therefore an increase in the reported
dose. At the extreme case where a 40 Gy film was placed 15 cm off axis, the pixel value was decreased by 35% for EBT3 and 16% for EBT-XD. The results highlight the importance of film position on the scanner bed and are indicative of the error contributed depending on the off-axis extend of the film used. If for example, a 6 cm wide film is irradiated at 40 Gy, the LSRA would be up to 0.5% for EBT-XD. In comparison, this is four times larger for EBT3 reaching up to 2%. The LSRA is generally smaller for EBT-XD due to lower optical densities in comparison to EBT3. However, the effect is still smaller for the 40 Gy EBT-XD film when compared to the 13 Gy EBT3 that has the same optical density. This must be related to the re-sized crystals in the sensitive layer of EBT-XD [155]. The findings of this section are in agreement with other published studies [182,185,189].

![Figure 4.3: Lateral scanner response artifact for EBT3 and EBT-XD.](image)

### 4.3.4 Dose verification for SRS

The average output of the linac, evaluated by two sets of ionisation chamber measurements in reference conditions, before and after film measurements, was 0.997 cGy/MU.
This suggested a deviation of only 0.3% from TPS reference conditions and therefore no corrections were applied.

The film sizes used were small enough to assume negligible errors from the LSRA considering the findings of Section 4.3.3. The maximum off-axis displacement of the 6 cm wide film was 3 cm, where the edges of the film were exposed to dose levels well below 20 Gy. Therefore, the measurement uncertainty attributed to the LSRA was estimated to be up to 0.2%.

Three independent measurements were performed with each type of film in an SRS dose distribution. Each set of three films was assessed to evaluate the repeatability of the measurement. Both types of film showed a high degree of repeatability. All isodose levels between 2 Gy and 24 Gy were found to be repeatable within 1 mm distance. This value is indicative of the reproducibility of film measurements performed in this work, as they were carried out under controlled conditions with little impact from other sources of uncertainty. These sub-millimetre differences seen were therefore accounted in the uncertainty budget. Figure 4.4 shows the three repeated measurements performed with EBT-XD film, overlayed with the TPS-calculated isodose map.

![Figure 4.4: Repeated measurements with EBT-XD film in SRS-type dose distributions. (Thin lines are film-measured isodoses, thick lines are TPS-calculated isodoses).](image)

When triple-channel corrected EBT-XD films were compared to the TPS-calculated isodoses, they agreed within 2 mm in the range of 5 Gy - 15 Gy and 1.5 mm in the range between 15 Gy - 22 Gy. In comparison, triple-channel corrected EBT3 was inferior as agreements were found to be within 2 mm and 6 mm respectively for the
4.3. Initial characterisation of EBT-XD - Results and Discussion

The agreement between films and TPS was also evaluated using gamma analysis for a number of criteria. Both local gamma (normalised to the local pixel dose) and global gamma (normalised to the film pixel with the maximum dose) criteria were tested. For local gamma (LG) criteria, a 20% threshold was applied. The difference in passing rates was also investigated between single (red channel) and red channel with TCC. The results from this analysis are shown in Table 4.1. For all criteria investigated, EBT-XD using triple-channel correction had the highest gamma passing rates. EBT-XD film at 3% DD and 1.5 mm DTA global gamma (GG) had an average passing rate of 99.8%, and for the same criterion but LG the average passing rate was 97.0%. Single-channel corrected EBT-XD film also showed high passing rates but these were, on average, 5.3% lower than triple-channel passing rates. Higher percentage passing rate losses, when switching from triple to single correction, were seen for the strictest criteria used.

EBT3 unexpectedly showed higher passing rates with single rather than with TCC. The lower passing rates seen with TCC may be caused by the lack of separation in calibration curves at high doses, increasing the uncertainty of the corrections applied. Single-channel EBT3 improved passing rates by an average of 37.3%. Despite this improvement, when comparing the highest EBT3 (single) passing rates to the highest EBT-XD (triple) passing rates, EBT-XD rates were (consistently) higher by an average of 6.7%. At the most challenging criteria, significantly better agreement was seen with EBT-XD reaching up to 18.1% passing rate increase. The improved agreement with the

<table>
<thead>
<tr>
<th>Film type and method</th>
<th>Local Gamma (%)</th>
<th>Global Gamma (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>3%</td>
<td>3%</td>
</tr>
<tr>
<td></td>
<td>2 mm</td>
<td>1.5 mm</td>
</tr>
<tr>
<td>EBT-XD, triple-channel</td>
<td>98.5</td>
<td>97.0</td>
</tr>
<tr>
<td>EBT-XD, single channel</td>
<td>94.9</td>
<td>91.3</td>
</tr>
<tr>
<td>EBT3, triple channel</td>
<td>55.6</td>
<td>45.0</td>
</tr>
<tr>
<td>EBT3, single channel</td>
<td>94.2</td>
<td>90.5</td>
</tr>
</tbody>
</table>

Table 4.1: Gamma analysis passing rates for EBT3 and EBT-XD in an SRS dose verification test.
TPS shown by EBT-XD and the high degree of repeatability shown are complimentary to the characteristics of this dosimeter and supportive of its use for SRS audit.
4.4 Initial characterisation of EBT-XD film - Conclusions

Initial characterisations demonstrated the differences between the two films in terms of their structure, water equivalence and response to dose. Both films are considered to be water-equivalent and structurally very similar. The extended dose range and improved response shown by EBT-XD at high doses are valuable features for SRS dosimetry, where the maximum doses measured can be up to 40 Gy.

The susceptibility of both films to the lateral artifact of a Epson Expression 11000XL scanner has been quantified. This was assessed for all colour channels and different dose levels. Based on these findings, a suitable scanning protocol can be adopted to minimise any perturbations to the measurement caused by the scanner.

The findings demonstrated the feasibility of achieving good agreement between film dosimeters and TPS, for a complex SRS dose distribution in a simple geometric phantom. High levels of repeatability were seen for both films. EBT-XD showed better agreement with the TPS, whilst allowing for TCC to be applied, whereas TCC was not effective for EBT3, but single-channel EBT3 measurements yielded high passing rates. There are however many advantages in applying TCC, mainly in compensating for film thickness variations, scanner noise and inter-scanning variations and resilience to contamination. Considering these, EBT-XD with TCC should be the preferred method.

On a final note, the levels of agreement shown are also attributed to good film handling practices such as using a glass compression plate to compensate for film curvature, wearing latex gloves, thoroughly cleaning the scanner and film, minimising exposure to ambient light and maintaining film in controlled conditions without exposure to extreme temperature, pressure or moisture.
4.5 Development of a film dosimetry protocol

Following on from the initial film characterisation, a film protocol was developed for the purposes of radiosurgery audit and using the advocated film type, EBT-XD. It was decided that for practical reasons this would be based at NPL. The facilities at NPL would allow for controlled film handling, irradiations when they were needed and scanning on an Epson Expression 10000XL scanner. Due to the fact that a different scanner would be used, a new calibration curve was generated. Seven pieces of EBT-XD film were irradiated in the dose range of 0 Gy - 40 Gy and scanned 72 hours after irradiation to generate new scanner-specific calibration curves. The response observed was identical to that shown previously in Figure 4.2 with a small expected shift on the y-axis, due to differences in pixel values obtained from two different scanners.

The subsections that follow outline some additional tests that were carried out to quantify the uncertainties associated with the film scanning process. The energy dependence was measured experimentally and additional considerations for accurate film dosimetry are described.

4.5.1 Scanner response

The Epson 10000XL and 11000XL are successive scanner models with the same geometry. They have some differences in their instrumentation, with the newer scanner allowing for higher scanning resolutions, but it is anticipated that these differences are irrelevant for the studied application. In order to assess the LSRA of the NPL scanner, a similar test to that described in Section 4.2.3 was performed. Figure 4.5a shows a comparison of the off axis response between the two scanners. As the response was symmetric on the central axis, only one lateral displacement is shown. The differences observed are negligible and confirm the reliability of the scanner with regards to the LSRA. As with the previous scanner assessed, if the films used are up to 6 cm wide (maximum off-axis distance of 3 cm), the uncertainty contributed in within 0.2%.

The response of the scanner was also tested along the scanning axis to investigate its variations. This was also performed using a method similar to the previous test, by placing three films irradiated with known dose along the scanning axis and recording
Figure 4.5: a) The lateral scanner response artifact of an Epson Expression 10000XL compared to the 11000XL model. b) The on-axis response of the Epson Expression 10000XL scanner.
4.5. Development of a film dosimetry protocol

the differences seen with respect to the central point of the scanner. The results shown in Figure 4.5b, demonstrate a more uniform response along the axis in comparison with off-axis measurements. The edges of the scanner show a tendency towards higher pixel values in comparison to the centre. A possible explanation for this is that ambient light entering at the edges of the scanner is responsible for this effect. This is consistent with the findings and explanation of another study [154]. In the central region of the scanner the effect is within 0.5%.

4.5.2 Positioning jig

In order to minimise the impact of the lateral and longitudinal effects, a rectangular frame was constructed to exclude regions at the edges of the scanner where the effects shown are larger. This is also used as a positioning jig allowing repeatable placement of films on the scanner bed, in a region where the scanner response is as uniform as possible. Figure 4.6 shows a photograph of the positioning jig (made of black non-reflective paper) securing the calibration films in a central position on the scanner, pressed against the scanner-bed with the use of a glass plate. The longitudinal and lateral scanner directions are indicated.

Two scans were performed to assess any perturbations caused by the presence of the positioning jig. Pixel values with and without the jig were compared for all colour channels throughout the calibration dose range (0 Gy - 40 Gy). The differences seen were within 0.5%, equal to the level of inter-scanner variations and therefore not indicative of any additional disturbances caused by the presence of the jig.

4.5.3 Scanner resolution

A scanning resolution of 96 dpi was considered reasonable for the purposes of the audit as it corresponds to 3.8 pixels per mm, an adequate resolution for comparison to a typical high-resolution 1 mm TPS dose grid. Moreover, it allows for quick scanning times and manageable image sizes. The noise contributed at this scanning resolution was assessed by calculating the standard deviation from the mean pixel value for consistent 4 cm² central regions of interest in the homogeneously irradiated calibration
4.5. Development of a film dosimetry protocol

Figure 4.6: Photograph of the scanner setup used for film calibration: Calibration films, glass compression plate and positioning jig are shown on the scanner bed.

films. This assessment was performed for all three colour channels and the level of noise was found to increase with dose, reaching a maximum of 0.8% at 40 Gy. It should be noted that these levels of noise were seen on the raw images using on a open-access image processing software (ImageJ), without any corrections applied. Non-uniformity corrections applied in triple-channel dosimetry have been demonstrated to lower this noise significantly [149].

4.5.4 Scanner stability

Following recommendations and suggestions from other film studies [138,152,154,190], the scanner was switched on, allowed several minutes to stabilise and 5 full length scans were performed to warm-up the scanner. The response of the scanner to three pieces of films irradiated at different dose levels was recorded in 11 scans acquired at 3 minute intervals over a total period of 30 minutes. The results for films of 0 Gy, 20 Gy and 40 Gy are shown in Figure 4.7. The scanner response at all dose levels tested did not exceed differences above 0.9%, after appropriate steps were taken to warm-up the scanner. As with the previous test, raw images were analysed to assess
the scanner response and no corrections were applied. Film-dose linear scaling has been demonstrated to decrease inter-scanning variations [153].

![Post-warmup scanner response graph](image)

Figure 4.7: Consistency of the Espon 10000XL scanner response over 30 minutes of activity.

### 4.5.5 Film Energy dependence

The energy dependence of the film was measured experimentally in a method similar to another study but for a different energy range to reflect the clinical energy range used for SRS [185]. This was done by comparing the mean pixel values in central 4 cm² regions of three film samples irradiated with a dose of 18 Gy: one in a ⁶⁰Co beam and the other two in nominal 6 MV and 10 MV linac beams. The three films were scanned together in the central region of the scanner, and their pixel value differences for all three colour channels were within 0.5%. This is comparable to the findings of Grams et al who showed differences within 0.8%.

### 4.5.6 Post-irradiation darkening

Previous studies have shown or suggested that post-irradiation darkening of radiochromic films occurs mostly in the first 48 hours after irradiation [140, 148, 152]. Due to this effect, it is recommended that films are scanned after this initial period of rapid changes.
in optical density. Regardless of this, for the purposes of a multi-centre audit, scanning films before 48 hours would be impractical. It was therefore decided to allow 72 hours post-irradiation for scanning test films, consistent with the practice followed for film calibration.

4.5.7 Film orientation

According to the manufacturer the re-sized crystals in EBT-XD should minimise any scanning orientation effects. For the practical reasons and for the purposes of this work, a consistent film orientation was used for all films. Orientation effects were therefore not evaluated as they did not constitute a source of uncertainty in the method presented.

4.5.8 Protocol

Taking into account the findings of the initial film characterisation and the subsequent tests performed, a step-by-step protocol was developed for the purposes of handling and scanning film for radiosurgery audit. This protocol outlines the steps and precautions to be undertaken when preparing, irradiating and scanning film samples (Appendix B).

4.5.9 Uncertainty budget

The findings presented in this chapter allow for the calculation of an uncertainty budget for dose measurements performed with EBT-XD film. The analysis of uncertainty follows the Joint Commitee for Guides in Metrology (JCGM) Guide to the Expression of Uncertainty in Measurement [191]. Uncertainties evaluated by statistical analysis are grouped as type A and the rest are grouped as type B. These are added in quadrature to give a combined standard uncertainty with coverage factor $k = 1$. Table 4.2 shows the calculated film uncertainty.
### Table 4.2: Uncertainty budget for Gafchromic EBT-XD film.

<table>
<thead>
<tr>
<th>Source</th>
<th>Standard uncertainty (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Film position (±0.5mm)</td>
<td>±2*</td>
</tr>
<tr>
<td>Inter-scanning variations</td>
<td>±0.9</td>
</tr>
<tr>
<td>Ion chamber measurement uncertainty</td>
<td>±0.8</td>
</tr>
<tr>
<td>Calibration fit</td>
<td>±0.5</td>
</tr>
<tr>
<td>Film repeatability</td>
<td>±0.5</td>
</tr>
<tr>
<td>Energy dependence</td>
<td>±0.5</td>
</tr>
<tr>
<td>Intra-scanning variations</td>
<td>±0.5</td>
</tr>
<tr>
<td>Lateral scanner effect</td>
<td>±0.2</td>
</tr>
<tr>
<td>Combined</td>
<td>±2.5</td>
</tr>
</tbody>
</table>

* Estimated film positional uncertainty in phantom of 0.5 mm - Dose value of 4% estimated for a typical maximum dose gradient of 8% per mm for k = 2

### 4.6 Development of a film dosimetry protocol - Conclusions

Scanner related uncertainties and the energy dependence of the film were quantified. The findings of this section, along with considerations from recommended good film practices were taken into account for the development of an appropriate protocol that aims to minimise uncertainties in film dosimetry. This protocol is suitable for the purposes of radiosurgery dosimetry audit.
Chapter 5

Characterisation of a new commercial plastic scintillator

5.1 The Exradin W1 plastic scintillation detector

The Exradin W1 became commercially available in the UK in the summer of 2014. The sensitive volume of the detector is a 3 mm (length) by 1 mm (diameter) polystyrene cylinder doped with scintillating agents. It is encased within an opaque enclosure made of epoxy and acrylonitrile butadiene styrene (ABS). The scintillator is coupled to an optical fibre with a 1 mm diameter polymethyl methacrylate (PMMA) core and a 2.2 mm diameter polyethylene jacket. The scintillator-fibre coupling is externally protected with a polyimide sheath. The fibre is 3 metres long and is attached to a metal photodiode box. Although the manufacturer does not provide a detailed account of the contents of the photodiode box, it is likely to include instrumentation for the chromatic separation of light (dichroic filters and photodiodes). The other end of the photodiode box has two connectors for transmitting the electrical charges collected by a dual channel electrometer. Two individual electrometers could also be used, as long as they are able to detect the small charges produced by the PSD which are in the pico-Coulomb range. The company also provides a 30 x 30 cm polystyrene calibration slab that allows placement of the detector in the min and max fibre orientations. The instrumentation described above is shown in Figure 5.1.
5.1. The Exradin W1 plastic scintillation detector

Figure 5.1: Photographs of the components of the Exradin W1 plastic scintillation detector, SuperMAX electrometer and calibration slab.
A study that performed MC simulations of different detectors in small fields reported that the Exradin W1 showed the best behaviour and its response was accurate within 1% [192]. Similar findings were experimentally validated by another study in the measurement of small field output factors [130]. The PSD was subsequently used to determine correction factors for other small field detectors. The available evidence supports the use of this detector in small fields. However, all characteristics of the detector need to be investigated to assess its suitability for SRS dose verification.

The Exradin W1 has been recently characterised by two independent and almost simultaneously published studies [124,125]. Some aspects were investigated by both studies and some were only presented in one of the two. The findings show good characteristics and minimal dependence to most factors evaluated. Slightly different findings were seen for the energy dependence and long term stability of the detector which were highlighted and discussed in a letter to the editor [126]. A response to the letter commented on the possibility that such differences may be observed between different detectors and highlighted the need for more studies investigating the Exradin W1 [127]. There are also some crucial investigations that will determine whether the detector is suitable for SRS dose verification which have not been within the scope of these two studies. The angular dependence of the detector along its polar axis and the manual collection mode for high dose measurements need to be investigated.

The purpose of this chapter was to conduct a characterisation of the Exradin W1, to verify published findings and investigate further, with the aim of adopting it in the methodology of a national SRS audit. The contents of this chapter were partly presented as an oral presentation at the 2nd International Conference in Dosimetry Applications in Guildford, UK and submitted to the Journal of radiation physics and chemistry for publication in a special edition volume of the conference proceedings.
5.2 Characterisation of the Exradin W1 - Methods

The detector was connected to a SuperMAX dual channel electrometer that is commercially available from the same manufacturer. The readings from both channels were acquired in the low range (pC), using both triggered and manual collection modes. All factors and dose measurements were calculated manually using the methods and equations shown in Section 2.5.2. Channel 1 of the electrometer collected the signal produced mainly from the scintillator (shown previously as Region 1), and channel 2 collected signal mainly produced from Cherenkov in the stem (Region 2). When trigger mode collection was used, channel 1 was automatically initiating and ending the measurement using the default threshold values of 0.4 pA (start) and 0.2 pA (stop). Manual collections were acquired by starting the collection right before the beam went on and stopping the collection after the beam went off and the dose rate indications for both channels of the electrometer returned to zero. Substantial leakage currents were occasionally noticed during the experiments. In order to minimise them, the detector was left to acclimatise for at least 10 minutes, pre-irradiated with a dose of approximately 10 Gy and the electrometer was subsequently corrected for background. The photodiode box was kept as far away from the beam as possible and shielded from scatter, as there is evidence to suggest that similar instrumentation is susceptible to noise from scatter radiation [193,194]. The irradiations were performed with an Elekta Versa HD linac and a Theratron (\(^{60}\)Co) unit at NPL, and a Varian Trilogy linac and Varian TrueBeam STx linac at Royal Surrey County Hospital (RSCH).

The work presented was undertaken with the PSD positioned in both perpendicular and parallel orientations to the beam. For perpendicular irradiations, the detector was calibrated in its calibration slab (Figure 5.1), using 30 x 30 cm blocks of WT1 (water-equivalent plastic material) for buildup and backscatter. The detector was placed at the isocentre with 5 cm buildup, 95 cm Source-to-Surface Distance (SSD), and at least 15 cm for backscatter, in a 40 x 40 cm field. Absolute dose measurements were performed in the same setup conditions but in a 10 x 10 cm field, using ionisation chambers with a calibration traceable to the NPL primary standard. For parallel irradiations, the PSD was placed in water, in a Blue Phantom plotting tank (IBA, Belgium).
PSD manufacturer recommends a different calibration procedure for this orientation [122] as the perpendicular calibration method using the calibration slab is not valid for these conditions. An alternative method was also proposed by Underwood et al and both of these methods were used for different tests in this work [130]. Different setup conditions than those stated above were used for some tests, which are described below in their respective sections. All uncertainties presented were calculated as one standard deviation of repeated measurements.

5.2.1 Dose response, collection mode and short-term repeatability

The response of the PSD to dose was checked in $^{60}$Co, 6 MV, 10 MV and 15 MV with respective TPR$_{20/10}$ Quality Indices (QI) of 0.682, 0.733 and 0.758 for the latter three beam qualities. For the linac beams, a PTW Semiflex ionisation chamber was placed 5 cm below the PSD to account for any variations in output. Measurements were performed using both triggered and manual collection modes to investigate their differences. The dose range investigated was from 0.1 Gy up to 40 Gy. The short-term repeatability of the detector was also assessed by performing repeated measurements in the same conditions in a $^{60}$Co beam.

5.2.2 Dose rate and dose-per-pulse

The dose rate dependence was investigated in air using a wooden clamp to ensure low scatter conditions and to position the detector parallel to the beam at the desired Source-to-Detector Distance (SDD). A cylindrical brass mini-phantom was fitted to the PSD to allow measurements beyond the depth of maximum dose (dmax). Irradiations were performed in a 3 x 3 cm field using the maximum dose rate available on the linac of 580 Monitor Units (MU) per minute. The SDD was varied from 70 to 130 cm and any deviation seen from the inverse square law was recorded as dose rate dependence. The 100 cm SDD setup was performed three times throughout the experiment to assess the uncertainty associated with the positional accuracy.

The dose-per-pulse dependence of the detector was also evaluated by varying the linac setting for dose rate from 100 MU/min up to 580 MU/min. For this test the detector was set up in a WT1 phantom and irradiated with a 10 x 10 cm beam at SSD 95 cm
and 5 cm deep.

### 5.2.3 Angular dependence

The purpose of the tests described below was to check for dependence in the angle of irradiation along both the symmetry and polar axis of the PSD. The gantry angle was kept at 0° for all tests. The irradiations were performed in a 3 x 3 cm field in order to minimise Cherenkov emissions from the stem.

For the symmetry axis angular dependence test, a cylindrical plastic sleeve was drilled for the PSD and marked along its circumference for 0°, 30°, 90° and 150° rotations in the clockwise and anticlockwise direction. The detector was fixed inside the sleeve, which was placed in a plastic phantom. The sleeve was rotated using the indicated angle marks. The signal collected at each rotation was normalised to the one at the reference angle 0° and the deviations were recorded as angular dependence along the symmetry axis.

For the polar axis angular dependence test, the PSD was initially positioned with its stem parallel to the beam in a water tank (reference detector angle 0°). The alignment cap provided by the manufacturer was used to align the sensitive volume of the detector with the water surface and the central axis of the beam (SSD = 95 cm). In order to evaluate the effect of the beam profile shape on the long axis of the detector, measurements were performed at three different depths: 1.5 cm, 5 cm and 10 cm. The PSD was rotated by 30° steps from 0° to 90°, where the PSD’s stem becomes perpendicular to the beam axis. Measurements were performed in clockwise and anticlockwise directions. A goniometer was used to visually verify the angle of rotation. The signal collected at each angle was compared to the one at the reference angle (0°) and the deviations were recorded as angular dependence along the polar axis. The reference irradiation was repeated three times throughout the experiment to determine positional uncertainty.

An additional test was performed to evaluate the effect of the detector orientation on dose measurements. For this tests, the detector was irradiated at a depth of 5 cm (95 cm SSD) in a 10 x 10 cm field with a 6 MV beam. The calibration was performed.
5.2. Characterisation of the Exradin W1 - Methods

following the manufacturer-recommended method for irradiations in a water tank with the PSD’s stem parallel to the beam axis [122]. This calibration procedure requires measurements with approximately 10 cm of the fibre exposed for the minimum setup condition and 20 cm - 30 cm for the maximum fibre setup condition. As recommended, the fibre bend radius was kept approximately the same for both conditions. The Gain and CLR factors from this calibration were then applied to measurements performed with the PSD’s stem perpendicular to the beam axis. Additionally, calibration factors were acquired with the PSD’s stem perpendicular to the beam axis, following the standard calibration procedure [168, 169]. Similarly, these calibration factors were applied to measurements performed with the PSD’s stem parallel to the beam axis. Five independent measurements were performed with both approaches and the averages were compared to the absolute dose measured with an NPL 2611 ionisation chamber. The deviation seen from absolute dose was recorded as dependence to orientation-specific calibration factor determination.

5.2.4 Temperature dependence

A water bath with a temperature calibration certificate traceable to NPL was used to maintain a temperature-controlled environment for the PSD. The detector was secured in the water bath held by a perspex stand, parallel to the beam at 5 cm deep (95 cm SSD). After allowing for a few minutes for the bath to achieve a homogeneous temperature, measurements with a thermistor placed close to the PSD were performed. To account for the temperature variations likely to be met in different linac bunkers, during a multi-centre audit, the range investigated was from 18 °C to 24 °C. Standard room temperature (20 °C) was used as a reference for comparison of the detector response.

5.2.5 Energy dependence

It is known that PSD CLR factors are energy dependent [124, 125]. The energy dependence was therefore investigated by calibrating the detector through a wide energy range ($^{60}$Co, 6 MV, 10 MV and 15 MV). Subsequently, measurements in all beams were performed and converted to dose, using calibration factors determined for all beam energies. The average of five measurements was recorded for each combination of beam
and calibration factors. Each of these combinations produced a dose measurement. The measured doses were compared to the absolute dose measured using a calibrated PTW Semiflex ionisation chamber. Any deviations from the absolute dose were recorded as energy dependence. The whole procedure was repeated a second time and the average values from both sets of measurements were calculated.

As the CLR factor shows energy dependence, the accuracy of the measurement could be compromised by the spectral change between large fields (calibration fields) and small fields (SRS fields). In order to investigate this effect, a MATLAB (MathWorks, Inc) model was developed. The model was based on the theoretical Cherenkov electron energy cut-off and refractive index of the fibre. A theoretical CLR was then analytically formalised using the Cherenkov emission spectrum and the detection efficiency functions for channels 1 and 2 of the detector. MC simulations of several photon energies were performed, and the spectra were then used to generate theoretical CLRs. These were compared with the previously experimentally measured CLRs to validate the model. The results shown in Section 5.3.5 were presented at ESTRO 35 in Turin, Italy [195].

5.2.6 Long-term stability

The degradation of the scintillator over its irradiation history needs to be evaluated. It is expressed as the loss of scintillation signal and is evident in the CLR factor change over accumulated dose. Studies with other PSDs suggest large losses of signal due to this effect [59]. However, recently it has been shown that acceptable levels of degradation can be achieved [121]. The manufacturer suggests that the detector should be re-calibrated every 1 kGy of exposures to account for this effect [169]. The long term stability of the detector was evaluated by periodically repeating the CLR calibration process in the same 6 MV beam over approximately 25 kGy of irradiation history. The decrease of CLR in relation to the initial measured value was recorded as the loss of sensitivity due to fibre degradation from accumulated dose.

---

1 This work was performed at NPL in collaboration with Hugo Bouchard who developed the MATLAB model.
5.2.7 Evaluation of detector response in small fields

The performance of the PSD was evaluated in the determination of small field output factors. Following the same approach as Underwood et al [130] the Exradin W1 was used as a class detector to determine detector-specific correction factors for an IBA SFD diode, a PTW E diode 60012 and an Exradin A26 micro-chamber. Measurements were performed in water for beam energies of 6 MV, 10 MV, 6 MV FFF, and 10 MV FFF and field sizes from 4 cm$^2$ down to 0.5 cm$^2$ at a depth of 10 cm and SSD 90 cm. Care was taken to determine the central axis (CAX) positions by measuring beam profiles in the inline and crossline directions through the various field sizes with all detectors. This work was presented at the 7$^{th}$ Latin-American medical physics congress in Cordoba, Argentina [196].

5.2.8 SuperMAX electrometer characterisation

As the manufacturer issued certificate for the SuperMAX electrometer did not fully cover the range of interest, tests were performed at NPL in the pC range for both channels. A calibrated current source was applied to each channel and 15 second collections for different currents ranging from 10 pA to 500 pA were acquired. Any deviations observed in range and linearity were included into the uncertainty budget.

---

$^{1}$This work was performed at NPL in collaboration with Ileana Silvestre Patallo who performed the analysis of the measurements.
5.3 Characterisation of the Exradin W1 - Results

5.3.1 Dose response, collection mode and short-term repeatability

The measurements showed linear response ($R^2 = 1.000$) in both pulsed and continuous radiation throughout the dose range investigated. There were no significant differences between triggered and manual collections for doses above 1 Gy, where variations of up to ±0.1% were observed. For doses between 0.1 Gy to 1 Gy, manual collection readings were noticeably higher than triggered collection readings, with the differences reaching up to 2.5% for 0.1 Gy. The detector’s short-term stability was found to be within ±0.3%.

5.3.2 Dose rate and dose-per-pulse

The deviation from the inverse square law in the dose rate measurements was found to be within ±0.5%. Repeated measurements of the 100 cm SDD setup showed repeatability within ±0.3% and therefore indicating reasonable levels of setup accuracy. The dose-per-pulse dependency was found to be slightly larger, reaching a maximum relative difference of 0.8% from the reference dose rate.

5.3.3 Angular response

The angular dependence along both axes of rotation of the PSD is shown in Figure 5.2. The dependence around the symmetry axis is very small as expected, and does not exceed 0.3% from the response at the reference angle of 0°. Along the polar axis, larger differences were expected as the sensitive volume is cylindrical and its length is three times the size of its diameter. The largest differences are seen when the detector was at a depth of 5 cm and at 60° - 90° rotations. The detector is showing over-response when rotated from the parallel orientation (0°) towards the perpendicular orientation (90°). The dependence shown is close to 1% of the reference irradiation at angle 0°. Repeated measurements at angle 0° showed a setup uncertainty of ±0.3%, similar to the detector’s short-term stability, and therefore not indicative of additional sources of uncertainty.
When parallel-specific calibration factors were applied to perpendicular measurements, the detector over-responded by 0.5%. When perpendicular-specific calibration factors were applied to parallel measurements, the detector under-responded by 0.7%. This effect appears to be greater than the detector reproducibility and indicates that a large amount of the dependence seen in the polar axis test may be attributed to calibration factor determination.

Figure 5.2: Angular dependence of the Exradin W1 plastic scintillation detector.
5.3.4 Temperature dependence

The temperature dependence of the PSD is shown in Figure 5.3. The signal produced decreases at a rate of approximately 0.25% per °C throughout the range investigated. Using the results presented, it is possible to apply temperature corrections to the measurements performed during an audit, in cases where a change in environmental conditions takes place.

![Temperature dependence graph](image)

Figure 5.3: Temperature dependence of the Exradin W1 plastic scintillation detector.

5.3.5 Energy dependence

The results of the measurements performed with all calibration factors are shown in Table 5.1. The deviations from absolute dose become noticeable when the calibration factors from a different beam are applied. These deviations become bigger for bigger energy differences between correction factors and beams. The maximum difference is seen when 15 MV correction factors are used in a $^{60}$Co beam (1.99%) and when $^{60}$Co correction factors are used in a 15MV beam (-1.62%). Reasonable agreement is seen with smaller difference in beam quality: 15 MV factors used in a 10 MV beam yield dose differences of 0.22%. When the correct factors were used for all energies the difference in absolute dose between the two detectors becomes negligible.
5.3. Characterisation of the Exradin W1 - Results

<table>
<thead>
<tr>
<th>Beam Energy</th>
<th>$^{60}$Co</th>
<th>6MV</th>
<th>10MV</th>
<th>15MV</th>
</tr>
</thead>
<tbody>
<tr>
<td>Quality Index</td>
<td>0.568</td>
<td>0.682</td>
<td>0.733</td>
<td>0.788</td>
</tr>
<tr>
<td>Calibration factors</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>$^{60}$Co</td>
<td>0.04%</td>
<td>-0.83%</td>
<td>-1.37%</td>
<td>-1.62%</td>
</tr>
<tr>
<td>6MV</td>
<td>1.22%</td>
<td>0.02%</td>
<td>-0.58%</td>
<td>-0.99%</td>
</tr>
<tr>
<td>10MV</td>
<td>1.70%</td>
<td>0.42%</td>
<td>-0.01%</td>
<td>-0.55%</td>
</tr>
<tr>
<td>15MV</td>
<td>1.99%</td>
<td>0.64%</td>
<td>0.22%</td>
<td>-0.04%</td>
</tr>
</tbody>
</table>

Table 5.1: Deviations from absolute dose for each factor-energy combination with the Exradin W1. Underlined results show where the correct factors were used.

The theoretical model-generated CLR factors are compared against experimentally measured CLR factors in Figure 5.4. They are plotted against the beam quality index of the different beams used. The comparison between experiments and the model, shows that the model reproduces the behaviour of the CLR energy dependence. However, the model under predicts the magnitude of the effect. For the energy range of $^{60}$Co to 18MV, CLR variation was measured to be about 1.8\% whereas the model predicts variations around 0.5\%.

![Figure 5.4: Experimentally measured and simulated energy dependence of the Cherenkov light ratio.](image-url)
5.3.6 Long-term stability

The loss of sensitivity of the PSD over 25 kGy of exposures is plotted in Figure 5.5. A rapid drop of about 8% was observed in the first 5 kGy we recorded (1.6% per kGy). In the following 20 kGy of exposures, the rate of loss of sensitivity decreased to about 0.2% per kGy.

![Figure 5.5: Long term stability of the Exradin W1 plastic scintillation detector.](image)

5.3.7 Evaluation of detector response in small fields

The small field output factors measured by all detectors in a 10 MV FFF beam are shown in Figure 5.6. In the larger fields measured, where the micro-ionisation chamber is considered a class detector, very good agreement was seen with the PSD for fields above 2 cm$^2$. In the two smaller fields, it is expected that the ionisation chamber will under-respond and the diode will over-respond due to the effects described in Section 2.1. In these fields the PSD is considered a class detector [130] and was used to determine detector specific correction factors for the other detectors used.
5.4. Characterisation of the Exradin W1 - Discussion

The detector showed good levels of reproducibility in measurements, similar to the levels reported by the other two studies [124, 125]. In order to achieve these levels of stability, the detector requires careful handling to ensure low leakage currents. Leakage appeared to become negligible when the photodiode box was shielded from scatter radiation and the optical fibre and cables were positioned with minimal strain applied to them. Since this system comprises of sensitive electronics and produces signals that are many orders of magnitude smaller than other dosimetry systems, leakage currents can produce large errors in measurement. However, it was possible to achieve high levels of reproducibility using the methodology described.

Figure 5.6: Small field output factors measured for a 10 MV FFF beam using various detectors including the Exradin W1 plastic scintillator.

5.3.8 SuperMAX electrometer characterisation

The linearity and range correction of the electrometer were found to be accurate within ±0.1% in the range of interest.

5.4 Characterisation of the Exradin W1 - Discussion
Manual and triggered modes did not show significant differences in the collected readings for doses above 1 Gy but the differences increased by an order of magnitude for doses below 1 Gy. The differences seen must be related to the amount of phosphorescence (delayed luminescence) collected at the end of a reading. Triggered mode stops the collection as soon as the signal passes the predefined threshold whereas manual collection, at least using the method described, always had a longer acquisition time. As the dose becomes larger, the contribution from phosphorescence is proportionally smaller and becomes negligible above 1 Gy. Carrasco et al reported similar differences when using manual collection mode at low doses but they did not investigate doses above 1 Gy [125]. Also, the authors do not describe how the manual collection was acquired so it is possible that a different methodology was used. Using the method described in this study, it was possible to achieve agreement within ±0.1% between two collection modes. In SRS plan verification measurements, the detector will be exposed to multiple beams and/or heavily modulated beams where a large amount of the dose collected will be delivered from penumbrae. The triggered mode is unreliable in ensuring appropriate start and stop time points for this collection, whereas the manual mode allows for this to be controlled by the user. The findings demonstrated that using manual mode and the consistent collection method described, accurate measurements are acquired in the dose range of interest, making this method suitable for the proposed audit.

The detector exhibits negligible angular dependence along the symmetry axis. Along its polar axis, a dependence of about 1% was observed, which is relatively small in comparison to diamond detectors that exhibit dependencies of up to 3% [52]. This dependency has not been previously reported and is important to include within the uncertainty budget for non-coplanar deliveries, like SRS. The determination of calibration factors, to be applied to SRS dose measurements, can only be performed using one of the two orientations. The standard perpendicular method is more reliable and practical for the purposes of an audit. As shown in the results, a dose difference of up to 0.7% was observed when calibration factors were acquired at a different orientation from the measurement. It should be noted, that the scenarios and uncertainties calculated from these tests are generous as the conditions simulated are unlikely to be met in
a clinical delivery. The tests accounted for situations where all of the dose is delivered from a single plane but in reality the delivery will be occur over multiple planes and the effects shown should be significantly smaller.

The results show a spread of 1.8% in the CLR factor determination over the range of energies investigated. This was expressed as a maximum difference of 2% in the dose measured. This difference diminishes when the factors are from a similar beam quality and disappear when the factors are acquired at the same beam energy. As demonstrated, the detector has an energy dependence which can be eliminated by performing the calibration in the same beam where the measurements are going to be performed. It should be mentioned that the calibration process most practical for an audit, requires irradiations in a 40 x 40 cm field. These conditions are not possible for all SRS delivery machines. However, by acquiring calibration factors in a beam with a quality index as close as possible to the measurement beam, the dose difference observed becomes very small.

The MATLAB model presented assumes isotropic Cherenkov emission while the angular distribution of the light varies with the electron kinetic energy and the optical fibre only guides light emitted at a specific angular range. Further improvements modelling Cherenkov light transport explicitly should confirm this. A potential application for the model would be to quantify the effect of small field spectral changes on the CLR.

The results shown for the long term stability of the detector are comparable with published studies. Beierholm et al show a rate of decrease in sensitivity of 2% per kGy over 1.5 kGy of exposures, comparable to the initial loss seen in this study (1.6% per kGy) [126]. Carrasco et al show the same trend in sensitivity loss but at different rates. However, the initial rate of loss seen in that study is similar to the latter rate seen in this study [125]. The results presented are supportive to the speculations that such differences in detector characteristics could be related to different pre-irradiation exposures by the manufacturer to overcome the need for frequent calibration [127]. The results suggest that the detector’s long term stability is sufficient for use in an audit. If a calibration is performed per kGy of exposures, the uncertainty contributed will be of the range of 0.2%.
The output factor measurements performed confirm that the PSD is a reliable detector for fields as small as 2 x 2 cm as the agreement with micro-ionisation chamber is very good. The detector-specific correction factors determined by the PSD are in good agreement with published correction factors, confirming the reliability of the detector in small fields [130].

5.4.1 Uncertainty budget

The findings presented in this chapter allow for the calculation of an uncertainty budget for any measurements performed for dose verification in SRS. The analysis of uncertainty follows the JCGM Guide to the Expression of Uncertainty in Measurement [191]. Uncertainties evaluated by statistical analysis are grouped as type A and the rest are grouped as type B. These are added in quadrature to give a combined standard uncertainty with coverage factor $k = 1$. Table 5.2 shows the calculated uncertainty budget using the manual collection mode for doses above 1 Gy, ranked from biggest to smallest.

<table>
<thead>
<tr>
<th>Source</th>
<th>Standard uncertainty (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Type A</td>
</tr>
<tr>
<td>Gain determination</td>
<td>±1.2</td>
</tr>
<tr>
<td>Angular dependence</td>
<td>±1.0</td>
</tr>
<tr>
<td>Dose rate dependence</td>
<td>±0.8</td>
</tr>
<tr>
<td>Ion chamber measurement uncertainty</td>
<td>±0.8</td>
</tr>
<tr>
<td>CLR determination</td>
<td>±0.7</td>
</tr>
<tr>
<td>Setup accuracy</td>
<td>±0.5</td>
</tr>
<tr>
<td>Long-term stability</td>
<td>±0.2</td>
</tr>
<tr>
<td>Manual collection mode</td>
<td>±0.1</td>
</tr>
<tr>
<td>Electrometer</td>
<td>±0.1</td>
</tr>
<tr>
<td>Combined</td>
<td>±2.1</td>
</tr>
</tbody>
</table>

Table 5.2: Uncertainty budget for the Exradin W1 plastic scintillation detector.
5.5 Characterisation of the Exradin W1 - Conclusions

The Exradin W1 plastic scintillation detector was tested in various conditions to characterise its behaviour. The results show good agreement with published data and the dependencies to all factors investigated are relatively small. The PSD used in the manual collection method, with careful calibration and handling constitutes a suitable dosimeter which can be employed in a multi-centre dosimetry audit. The measurement uncertainty calculated is generous and accounts for the worst case scenario for measured doses above 1 Gy.
Chapter 6

Adaptation and validation of a phantom for radiosurgery audit

6.1 Phantom adaptation - Introduction

This chapter presents the adaptation of a commercial anthropomorphic phantom for the novel use of three simultaneous detectors, with the purpose of employing it for a radiosurgery dosimetry audit. The methodology utilised was guided by the current practice for cranial SRS in the UK presented in Chapter 3 [179]. This indicated that single brain metastatic lesions of volumes ranging from 1 to 20 cm$^3$ are the most commonly treated indications. The survey also highlighted that quality assurance programs should provide confidence not only in the dose delivered, but also the location and shape of the dose distribution delivered, as SRS is often prescribed to irregularly shaped targets.

The goal of the adaptation was to combine a close representation of a typical radiosurgical patient case, with a design capable of simultaneous measurements with a number of previously characterised detectors, suitable for SRS dose measurement. These were the new commercially available radiochromic film EBT-XD presented in Chapter 4, the new commercially available Exradin W1 PSD presented in Chapter 5 and alanine pellets from the NPL chemical dosimetry service [113]. The validation of this phantom and detectors combination included verification that the different systems did not adversely affect one another when making simultaneous measurements.
The work presented in this chapter has been submitted to Biomedical Physics and Engineering Express (under review at the time of submission of the thesis).

6.2 Phantom adaptation - Methods

STE$_2$EV is a commercially available anthropomorphic phantom (CIRS, Norfolk VI, USA) which has been designed with a range of materials to simulate tissue electron densities. The phantom contains bone and soft tissue structures, as well as teeth and air gaps to reflect realistic anatomy. The phantom design allows the insertion of interchangeable cuboid inserts in the centre of the brain and the insertion of radiation detectors through two parallel cylindrical access cavities that run superior to inferior through the phantom connecting the neck to the brain. The two cylindrical access cavities are 3 cm apart, centre to centre. The anterior cavity is aligned with the trachea and the posterior cavity is aligned with the spinal cord.

Figure 6.1: The STE$_2$EV anthropomorphic phantom, bespoke inserts and detector sleeves developed.
Two interchangeable bespoke cuboid inserts were developed for the brain cavity of the phantom. The first, the target insert, was modelled on real anatomical structures from clinical CT to simulate a patient as closely as possible. The insert is a 3D-printed cube made of proprietary resin that contains two liquid-fillable structures; one irregularly shaped “target” structure (Planning Target Volume – PTV) designed to simulate a centrally located brain metastasis of typical size ($\approx 8 \text{ cm}^3$) and another “organ at risk” (OAR) structure in the shape and size of the brainstem and thalamus of the brain (Figure 6.2a). The two structures are aligned and centred with the cylindrical access cavities. The posterior surface of the PTV structure is approximately 1 cm away from the anterior surface of the OAR.

![Figure 6.2: CT scan sections of the STE$_2$EV phantom through axial, coronal and sagittal planes with the target insert (a) and dosimetry insert (b) placed inside the cavity.](image)

The second, the dosimetry insert, is made of the same material as the surrounding brain equivalent material, obtained from the manufacturer, and was designed for multiple simultaneous dose measurements (Figure 6.2b). It is comprised of three cubes which, when joined together, have the same dimensions as the target insert (Figure 6.2a). The dosimetry insert was engineered to have two planar indentations of 280 microns depth to be loaded with film, one in the sagittal and one in the axial plane. Sections through
these are shown in Figure 6.2b. The axial film is positioned such that it bisects the target structure in the superior to inferior direction. The sagittal film is positioned superior and perpendicular to the axial film, such that it bisects the superior half of the target structure.

Figure 6.3: Successive pictures of the different parts of the dosimetry insert being put together. The last two images show the dosimetry insert fitted into the phantom and the film in the insert after irradiation.
Three small asymmetric fiducial markers (1 mm diameter) have been built into each film plane to allow accurate registration of the film when doing the analysis.

The lower part of the insert, which sits under the axial film, allows access from the inferior direction through the cylindrical cavities of the phantom, for the placement of other radiation detectors. Two bespoke detector holders have also been manufactured to hold the PSD and a stack of four alanine pellets. They are interchangeable so that they can be used to make measurements in both cylindrical cavities. The holders were designed so that the geometric centres of the top alanine pellet and the PSD are aligned as shown in Figure 6.4. The goal was to have the two measurement systems placed in the same position and enable two different geometries to produce comparable measurements for the mean dose absorbed in their volume.

![Figure 6.4: Schematic representation of the sagittal plane through the middle of the STE\textsubscript{2}EV phantom showing all detectors on the Planning Target Volume (PTV) and Organ at Risk (OAR). The scintillator positions are superimposed on the interchangeable alanine pellet positions. Both are cylindrical detectors - dimensions of scintillator: 3 mm length x 1 mm diameter, dimensions of pellets: 2.5 mm length x 5 mm diameter).](image-url)
The methodology was designed to enable accurate point measurements as well as accurate dose distribution measurements. Alanine \cite{111} and radiochromic film \cite{118, 119} have been chosen due to their good performance in dose measurement and successful use in other previous radiotherapy dosimetry audits. EBT-XD film was preferred to the most commonly used EBT-3 film as it was previously demonstrated (Chapter 4) that it is more suitable for small field high dose applications, mainly due to its extended dose range and reduced lateral scanner effects. Both alanine and Gafchromic film are near-water equivalent detectors with small sensitive volumes (high spatial resolution), characteristics considered ideal for SRS dosimetry. The Exradin W1 PSD is a novel detector with attractive characteristics that are suitable for SRS-type small field measurements. The chosen methodology enables us to assess the performance of this detector in SRS against the TPS calculated dose but also against alanine which is an established dosimeter. All detectors chosen, have small sensitive volumes compared to the PTV and field sizes typically used for radiosurgery. Assuming a relatively homogeneous dose distribution in the target, these detectors are expected to generate reliable measurements. For measurements in the OAR, larger deviations are expected due to lower absolute doses and steeper dose gradients.

The Exradin W1 PSD was calibrated for its stem effect (Cherenkov Light Ratio – CLR) and dose-to-water (GAIN) correction factors following the manufacturer’s recommendations \cite{168} in a 40 x 40 cm and 10 x 10 cm field respectively, with the later done against a PTW Semiflex chamber with a traceable calibration to the National Physical Laboratory, Teddington, Middlesex, UK.

6.2.1 Positioning accuracy of inserts

Multiple CT scans of 0.625 mm slice thickness were performed with each insert inside the phantom cavity. The inserts were removed and replaced between each scan to assess the reproducibility of positioning the inserts. The CT-origin was centred for all scans using the phantom’s built-in fiducials and the CT coordinates of two opposite corners of the cube insert were recorded for each scan. The maximum vector distance between the coordinates of the corners was calculated and defined as the positional reproducibility of the insert. CT scans were also taken with the PSD and alanine pellets in both cavities.
6.2. Phantom adaptation - Methods

6.2.2 Basic plan dose verification with ionisation chamber

A basic treatment plan was generated on the phantom, to be tested for plan dose verification, in order to establish a baseline of the dosimetric accuracy of the phantom. The treatment plan was a 3-field conformal radiotherapy plan with one anterior beam at gantry angle 0° and two lateral beams at 90° and 270°. All three fields had primary jaw collimation fixed at 3 x 3 cm and MLC collimation in their four corners to produce approximately circular 3 cm fields. The plan isocentre was placed in the middle of the anterior phantom cavity, to match the position of a calibrated PTW Semiflex ionisation chamber. The plan was delivered to the phantom and the measured dose for each beam was compared to the TPS predicted mean dose for a contour structure created to match the ionisation chamber sensitive volume.

6.2.3 Radiochromic film measurements

Film measurements were performed three times to assess the repeatability of the measurements. A measurement was conducted with the film only inside the phantom and it was repeated twice with the other detectors present, in order to assess the effect of any perturbations caused by the presence of the alanine and PSD on the film. In the absence of the film and alanine pellets, water equivalent rods were used to fill the cylindrical cavities. The methodology used for the film analysis in this chapter was similar to the one previously presented in Chapter 4. The films were scanned on an EPSON Expression 10000XL scanner at 96 dpi (0.265 mm resolution) in transmission mode with no corrections applied, and analysed on FilmQA Pro software (Ashland ISP Advanced Materials, NJ, USA). They were scanned consistently in the landscape orientation, 3 days after exposure to allow for post-irradiation darkening to occur. The films were always kept together in a controlled environment and were handled with latex gloves. A glass compression plate was used during the scans to keep the films flat on the scanner. High resolution (1 mm grid) predicted dose planes were exported from the TPS.
for comparison with the 48bit images of digitised films. A selection of gamma passing rates [186] suitable for SRS plan analysis were chosen. These include both local gamma (LG) and global gamma (GG) criteria to highlight the impact of agreement in the low dose regions of the dose map distributions. The distance-to-agreement (DTA) parameter of the gamma criteria used was always kept below 2 mm as distances above that are considered unacceptable tolerances for SRS delivery. Similarly the dose-difference (DD) parameter was varied between 2-5% to be representative of acceptable tolerances in SRS whilst accounting for the steep dose gradients present. The gamma passing rates used were collected using the red colour channel with triple channel dosimetry correction (TCC) [149] for the regions shown in Figure 6.5 (50 x 60 mm axial and 70 x 40 mm for sagittal), with a cut-off threshold for doses below 2 Gy to remove areas of low signal and high noise from the analysis.

6.2.4 End-to-end validation test

The phantom, inserts and detectors were tested for their suitability in performing an end-to-end dosimetry test for SRS. The phantom was immobilised in a thermoplastic mask used for radiotherapy treatment and CT scanned with both inserts in the cavity. The two scans were then imported into iPlan (BrainLAB AG, Feldkirchen, Germany) where they were fused, the target and detectors were contoured and a 7-field IMRT plan was generated using the local protocol’s prescription practice. The plan was delivered on a TrueBeam STx Linac (Varian, Palo Alto, CA, USA) with ExacTrac (BrainLAB AG, Feldkirchen, Germany). Measured point doses and dose planes were compared to Treatment Planning System (TPS) predicted doses and planes.
6.3 Phantom adaptation - Results

6.3.1 Positioning accuracy of inserts

The vector distances from the origin for the two points of the dosimetry insert were found to range from 53.51 - 53.79 mm for point 1 and from 74.18 - 74.61 mm for point 2. For the target insert the two distances ranged from 53.47 - 53.87 mm and 74.06 – 74.46 mm respectively. The maximum deviation in position was less than 0.5 mm for both inserts and the positional agreement between the two inserts was found to be better than 0.6 mm. The detector holders showed reproducible placement of the PSD and alanine detectors and confirmed the intended relative positions were as designed and shown in Figure 6.4. This analysis was limited to a qualitative assessment as any quantification of the PSD position was not possible due to the small and water equivalent sensitive volume of the detector that was difficult to visualise on the CT scans.

6.3.2 Basic plan dose verification with ionisation chamber

Table 6.1 shows the ionisation chamber measurements for the plan dose verification of the basic 3-field plan. The percentage difference for each beam and the overall difference were within 0.5%, suggesting an acceptable phantom accuracy.

<table>
<thead>
<tr>
<th>Beam Name</th>
<th>Measured dose (in cGy)</th>
<th>Predicted dose (in cGy)</th>
<th>Percentage Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Left Lateral</td>
<td>50.7</td>
<td>50.5</td>
<td>0.4%</td>
</tr>
<tr>
<td>Anterior</td>
<td>101.2</td>
<td>100.7</td>
<td>0.5%</td>
</tr>
<tr>
<td>Right Lateral</td>
<td>49.8</td>
<td>49.6</td>
<td>0.5%</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td><strong>201.7</strong></td>
<td><strong>200.8</strong></td>
<td><strong>0.5%</strong></td>
</tr>
</tbody>
</table>

Table 6.1: Measurements of a basic 3-field plan dose verification using the STE2EV phantom and a Semiflex ionisation chamber. The calculated k=1 uncertainty of the chamber is ±0.7%. 
6.3.3 Radiochromic film measurements

High levels of agreement were found between the film and TPS, whether the alanine or PSD were in place or not. All GG criteria used (Table 6.2) were in all cases agreeing with more than 93.2% of pixels passing. Any differences seen between LG and GG had the disagreements mainly in the low dose areas. However, these disagreements are not significant as the results for 5%-2mm LG were always above 96.1%.

The variation in gamma passing rates between the three pieces of film was very similar for all the criteria investigated. The maximum percentage difference was seen with the strictest criteria of 5%-1mm LG and 5%-1mm GG, which was 3.8% and 3.7% respectively.

<table>
<thead>
<tr>
<th>Detectors used</th>
<th>Local Gamma</th>
<th></th>
<th></th>
<th></th>
<th></th>
<th>Global Gamma</th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>5%-2mm</td>
<td>3%-2mm</td>
<td>2%-2mm</td>
<td>5%-1mm</td>
<td>5%-2mm</td>
<td>3%-2mm</td>
<td>2%-2mm</td>
<td>5%-1mm</td>
<td></td>
</tr>
<tr>
<td>Film only</td>
<td>97.1</td>
<td>87.1</td>
<td>73.3</td>
<td>73.6</td>
<td>99.8</td>
<td>96.4</td>
<td>95.9</td>
<td>93.2</td>
<td></td>
</tr>
<tr>
<td>Film &amp; PSD</td>
<td>96.1</td>
<td>88.0</td>
<td>76.0</td>
<td>70.4</td>
<td>100</td>
<td>99.8</td>
<td>98.0</td>
<td>96.9</td>
<td></td>
</tr>
<tr>
<td>Film &amp; alanine</td>
<td>96.9</td>
<td>87.9</td>
<td>75.1</td>
<td>69.8</td>
<td>100</td>
<td>97.4</td>
<td>96.2</td>
<td>95.4</td>
<td></td>
</tr>
</tbody>
</table>

Table 6.2: Gamma passing rates for EBT-XD axial film measurements with and without the PSD/alanine abutting the film plane.
6.3.4 End-to-end validation test

Table 6.3 shows results for the PSD and alanine measurements. The difference between measured and TPS dose was within 0.4% for the PTV and the agreement between the PSD and pellet 1 was also within 0.4%. For the OAR the difference between the PSD and pellet 1 was 1.2%. The film dose plane measurements are shown in Figure 6.5.

<table>
<thead>
<tr>
<th>Detector location</th>
<th>Detector used</th>
<th>Measured dose (in cGy)</th>
<th>Predicted dose (in cGy)</th>
<th>Percentage Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>PTV</td>
<td>PSD</td>
<td>2605.7</td>
<td>2598.0</td>
<td>0.3%</td>
</tr>
<tr>
<td></td>
<td>Pellet 1</td>
<td>2595.6</td>
<td>2598.0</td>
<td>-0.1%</td>
</tr>
<tr>
<td></td>
<td>Pellet 2</td>
<td>2598.4</td>
<td>2595.0</td>
<td>0.1%</td>
</tr>
<tr>
<td></td>
<td>Pellet 3</td>
<td>2604.4</td>
<td>2595.0</td>
<td>0.4%</td>
</tr>
<tr>
<td></td>
<td>Pellet 4</td>
<td>2613.6</td>
<td>2604.0</td>
<td>0.4%</td>
</tr>
<tr>
<td>OAR</td>
<td>PSD</td>
<td>51.5</td>
<td>43.0</td>
<td>19.8%</td>
</tr>
<tr>
<td></td>
<td>Pellet 1</td>
<td>51.0</td>
<td>43.0</td>
<td>18.6%</td>
</tr>
</tbody>
</table>

Table 6.3: Dose measurements in the target and OAR compared to TPS predicted doses. The calculated k=1 uncertainties of the PSD and alanine pellets are ±2.1% and ±2.0% respectively.
Figure 6.5: Dose distribution comparisons between the film-measured doses (thin lines) and the TPS-calculated doses (thick lines) for the axial and the sagittal films used in the end-to-end test.
6.4 Phantom adaptation - Discussion

As the phantom and inserts are intended for performing end-to-end tests in SRS it is imperative that the positional uncertainty of the inserts is as low as possible. In the survey conducted in Chapter 3 it was found that the majority of SRS centres in the UK reported positional accuracies less than 1 mm for SRS delivery. Considering the engineering tolerances allowed to facilitate insertion and removal of the two inserts inside the phantom, the sub-millimetre variations found are considered acceptable.

The agreement between film and TPS for the three films used in this study showed mean passing rates of 96.7% for 2%-2mm global gamma. This shows equivalent agreement with the measurements previously performed in a simple homogeneous phantom in Chapter 4. The results presented in this study also demonstrate similar levels of agreement with the TPS to other studies that also used anthropomorphic phantoms [97, 119]. Most importantly, the film measurements show a high degree of repeatability which is essential if the methodology is to be employed for audit purposes. The high agreement to the TPS and repeatability in measurement suggest that this is an appropriate phantom design and film methodology to be used for an end-to-end audit.

The PSD showed very good agreement with the TPS and the first alanine pellet. The agreement between the two detectors in the target was within 0.4% demonstrating the potential of positioning two detectors of different geometries in the same area to compare their measurements. The results also suggest that the Exradin W1 PSD could be a suitable detector for end-to-end audits.

The measurements performed in the brainstem show larger differences from TPS calculated doses than the measurements in the target. This is due to the measurement being performed in a region with a high dose gradient, where a small positional deviation results in a large dose difference. The detector signal measured is also much lower than the detector signal measured in the target which hence produces larger uncertainties in the relative measurement. Despite the above, the absolute difference between the measured and predicted doses is at low levels of 8-9 cGy. Also, the difference between the alanine pellet in the brainstem and the PSD is within 1% giving more confidence to the measurement. Moreover, the two film measurements performed that sit medi-
ately superior to the two detectors also show that the dose measured in the brainstem is slightly higher than the TPS predicted dose and therefore confirms that the measurements of the three independent systems provide consistent results, even in regions where the TPS calculation algorithm may be less accurate.

The perturbation of the alanine and PSD on the sagittal film plane was expected to be very low due to the near-water equivalent density of the two detectors. Also, the small volume of the two detectors, relative to the large area of the film, would only potentially affect a small number of pixels, which would be reflected in a small change in gamma passing rates. The gamma passing rate differences observed between the three exposures, are mostly attributed to dose differences in the low dose regions. These must therefore be related to film measurement variations or treatment delivery variations, but not detector perturbation. The results confirm that the differences seen are very low as hypothesised.

The phantom and inserts used enabled a thorough test of all aspects of the treatment protocol in place for SRS. The immobilisation system, CT scan and scanning protocol, import of images into TPS, fusion of two sets of scans, TPS accuracy, export of images on the treatment platform, pre-treatment imaging for precise positioning and finally dosimetric and geometric accuracy of the treatment delivery were all tested.
6.5 Phantom adaptation - Conclusions

This chapter presented the adaptation of a commercial anthropomorphic phantom with suitably designed inserts to image and irradiate in order to be suitable for an end-to-end dosimetry audit of SRS. It was demonstrated that the inserts produced have reproducible positioning inside the phantom and they can be utilised for end-to-end tests to provide accurate and repeatable measurements. The adopted methodology was tested for dose verification and it is possible to achieve high agreement and repeatability between TPS and the three detectors used in the phantom (EBT-XD film, alanine pellets and the Exradin W1 PSD). The use of all detectors simultaneously inside the phantom does not produce any significant perturbations and is therefore a suitable and time-saving methodology. The work presented has demonstrated a novel combination of three detectors for simultaneous measurement in an anthropomorphic phantom which create a suitable phantom-detector system and methodology for an end-to-end SRS dosimetry audit.
Chapter 7

Methodology of the national audit for stereotactic radiosurgery

7.1 National SRS Audit - Introduction

Chapters 4 and 5 presented the characterisation of appropriate detectors for SRS dose verification and showed the development of suitable measurement methods. Chapter 6 presented the use of these detectors and alanine in an anthropomorphic phantom, demonstrating their suitability for end-to-end radiosurgery audit. In this chapter, the methods developed were incorporated into the procedure used for conducting an end-to-end audit for SRS.

The work presented in this chapter was endorsed by the RadioTherapy Trials and Quality Assurance (RTTQA) team, and was used as the credentialing QA exercise that was completed by centres participating in two clinical trials. Additionally, NHS England procured the commissioning of SRS services in early 2016 and used the results of this audit as the part of the credentialing exercises completed by centres that competed in the selection process [197].
7.1.1 Initial investigations of a pilot audit study

Initial investigations for compatibility of the end-to-end audit methodology presented in Section 6.2.4 with all treatment delivery platforms were performed. As shown previously in Section 3.3.3 most centres use MRI scans for treatment planning. Therefore, the initial tests incorporated MR imaging of the phantom. This confirmed the limited visibility of the phantom on MRI scans and the difficulty of booking MR time for non-clinical work. As a pragmatic approach, it was therefore decided that the audit procedure would require centres to only perform CT scans of the phantom, due to the good visibility of the phantom and detectors and the availability of scanning slots.

During these initial tests it was also noticed that inserting and removing the water-filled target insert could be problematic. It could lead to damaging both the phantom and the insert, but also had the potential to compromise the measurements performed with the (hygroscopic) alanine pellets, due to water drops left in the phantom after removal of the insert. Moreover, it was noticed that during the filling process, small air bubbles were left in the insert. If dose calculations were performed on this insert, algorithms with inhomogeneity corrections would be taking into consideration the lower density regions and predicting slightly different dose distributions. Due to the above reasons, it was decided not to scan the phantom with this insert during the audits but rather to provide participating centres with a CT image set of the phantom with the insert in place. This image set was then used only for outlining the target and OAR, but not for dose calculations. After contouring, the provided image set could be fused with a scan taken on the day, with the dosimetry insert placed inside the phantom. The centre then proceeded to treatment planning and dose calculations. This process avoided dosimetric inaccuracies contributed from density differences between the two inserts. It also streamlined the audit as the CT image set was sent to the centre in advance and by the day of the visit the centre had the opportunity to prepare a suitable plan.

The film analysis methodology presented previously in Section 6.2 was used in these initial investigations to assess its compatibility with all SRS platforms. Some of the exported dose planes from the TPSs tested were incompatible with the film analysis
software, but it was possible to overcome these by editing the files using third party software prior to the analysis. It was also noticed that the coordinate systems used by the TPSs tested were different. Due to these differences, in some cases it was impossible to use the fiducial markers for co-registering the film with the exported dose plane. Where the use of fiducial markers was possible, it added significant time to the analysis and proved to be an impractical method for use in a national audit. In order to overcome these problems, it was decided not to use the fiducial markers registration function, but instead utilise the optimisation function of FilmQA Pro. This function attempts multiple registrations between the two dose distributions to find a best fit, where the gamma passing rate is as high as possible. The film was initially positioned manually and one optimisation cycle was subsequently applied. The positional shifts applied by the optimiser to the initial position were monitored to ensure that they were within reasonable limits (±1 mm and ±1° rotations).

7.2 National SRS Audit - Methods

7.2.1 Schedule of audits

Audit visits were performed between January 2016 and July 2016 in 26 participating centres. During these visits, 28 treatment plans were assessed as two centres participated with two different treatment modalities. Two auditors were always present during each audit, where at least one was a registered clinical scientist from NPL and/or RT-TQA. The author was present in 27 audits, Russell Thomas of NPL in 11, Jonathan Lee of RTTQA in 6, David Eaton of RTTQA in 5, Rushil Patel of RTTQA in 4, Ileana Silvestre Patallo of NPL in 2 and Rada Zotova of RTTQA in 1. The geographical locations and equipment groups of the UK centres visited are shown in Figure 7.1. Sixteen treatment plans were delivered by Linacs, seven by GKs, four by CKs and one by TT.
Figure 7.1: Map of the UK showing the geographical locations and equipment groups of the centres visited.

7.2.2 Audit image set

A high resolution CT scan (1 mm slice thickness) of the phantom was performed with the target insert in place (see Figure 6.2a). The insert was carefully filled with water following the manufacturer’s recommendations for minimising air bubbles to improve image quality. The image set was saved in Digital Imaging and Communications in Medicine (DICOM) format and sent to participating centres along with instructions on how to outline the target and OAR. This scan was the secondary data set and was used instead of the MRI scan most commonly used in SRS patient pathways. After outlining the two structures, the centres were able to use the scenario presented to them in deciding on an appropriate treatment plan.
7.2.3 Immobilisation and CT scan

On the day of the audit, centres were asked to “immobilise” the phantom following the local protocol for a patient with a single brain metastasis. The phantom was loaded with the dosimetry insert, axial and sagittal films, four alanine pellets in the posterior cavity and the PSD in the anterior cavity. A laser alignment system was used to position the phantom parallel to the couch. Where a laser system was not available, handheld levelling devices were used instead. Figure 7.2 shows some of the immobilisation systems used by different centres.

![Immobilisation systems used by different participating centres in the audit: a) BrainLab thermoplastic mask, b) U-frame thermoplastic mask, c) White orfit thermoplastic mask, d) Leksell Gamma Knife stereotactic frame.](image)

To ensure a reliable setup with the Leksell stereotactic frame without damaging the phantom, hard rubber caps were placed on the stereotactic pins. The frame was firmly attached so that the pins were pushing the rubber caps against the surface of the phantom. Initial tests showed reliable fixation with this method and it was considered suitable for use in the audit.
After immobilisation, a CT scan (primary data set) was performed following the centre’s protocol for the scenario presented. Figure 7.3 shows an example of a CT scan acquired during an audit visit, with the phantom immobilised and all detectors in place.

Figure 7.3: Example of a primary data set CT scan acquired during an audit visit with the dosimetry insert and all detectors placed inside the phantom.

7.2.4 Image fusion and contouring

The primary data set was then imported into the TPS and co-registered (fused) with the secondary data set, which was provided prior to the visit. The local planner verified that the fusion between the two scans was correct and proceeded to outline all structures
required, following the instructions provided by the auditors (Appendix C).

As shown in Figure 7.3, the primary data set was acquired with the PSD in the target and alanine pellets in the OAR. As these structures were visible on the scan, the planners were able to contour them directly. In order to allow dose comparisons with both detector systems in both phantom cavities, the contoured structures were copied and positioned concentrically to the other cavity. As both detectors are near water-equivalent, density perturbations were considered negligible. The copied contours are shown in Figure 7.4. The four alanine pellet structures were copied and centred into the anterior cavity position, and the PSD structure was copied and centred into the posterior cavity position.

![Figure 7.4](image)

Figure 7.4: Sagittal section through the middle of the phantom showing copied and repositioned contours.

Figures 7.5a and 7.5b, show sagittal sections of the two data sets fused together. The blue-framed pane tool shown, displays the primary data set position relative to the secondary data set during the fusion procedure. Figure 7.5c, shows an example of the primary data set with all required contours displayed.

Up to this stage, the procedure described tested the patient pathway for immobilisation, CT scanning, image export, image registration, image fusion and contouring.
Figure 7.5: Sagittal sections of the phantom showing fusion of the primary and secondary CT data sets (a & b) and contours for all structures (c).
7.2.5 Treatment planning

Following contour delineation, audited centres proceeded to treatment planning. They were asked to follow their local prescription protocol and tolerance doses. Treatment plans acceptable to local practice were generated, checked and exported to the delivery platform. Plan exports were collected by the auditors for retrospective analysis. Sagittal and axial dose planes calculated with a 1 mm grid were also collected by the auditors for comparison to film planes. The maximum, minimum and mean doses as calculated by the TPS were recorded for all contoured structures. Figure 7.6 shows an example of a treatment plan generated on iPlan (BrainLab AG, Feldkirchen, Germany) that was planned with four non-coplanar dynamic conformal arcs.

![Figure 7.6: Example of a audit treatment plan: a) beam orientation, b) axial view through film plane, c) sagittal view through film plane, d) frontal view through anterior phantom cavity.](image-url)
7.2.6 Treatment delivery

After completion of the treatment plan, the auditors continued with a measurement session on the delivery platform. Output measurements were performed, followed by calibration of the scintillation detector and a reference film measurement. Subsequently, the anthropomorphic phantom was setup with all detectors in place. The phantom position was verified using on-board imaging, where available, and the phantom was repositioned if necessary following the local protocol. When on-board imaging was used, a small imaging dose was inevitably delivered to the film and alanine pellets. This is, however, negligible in comparison to the treatment dose. Following phantom positioning and verification imaging, the treatment plan was delivered twice. Between the two treatment deliveries, films and alanine pellets were replaced. The position of the PSD was also swapped with the new set of alanine pellets for the second measurement.

7.2.7 Output measurement

A PTW Semiflex ionisation chamber, traceable to the graphite calorimeter primary standard at NPL, was used in all centres to perform output measurements. The chamber was placed in WT1, using the centre’s reference conditions, to measure the output in the machine specific reference field \([80, 198]\). For linacs, CKs and Tomotherapy the beam quality index (\(\text{TPR}_{20\text{cm}}^{10\text{cm}}\)) was measured to derive a beam-specific dose-to-water correction factor for the chamber. Ion recombination correction factors were measured using a two-voltage method \([199]\). For GKs, output measurements were performed in spherical phantoms provided by the centre. A \(\text{TPR}_{10\text{cm}}^{20\text{cm}}\) of 0.568 for \(^{60}\text{Co}\) was assumed and ion recombination was considered negligible. All measurements were corrected for temperature and pressure. The measurements were analysed with respect to the TPSs reference conditions and the deviations from agreement were expressed as percentage differences.

7.2.8 EBT-XD film measurements

Reference film irradiations were performed in WT1 blocks at the centre, using the local reference conditions. Where this was not possible, the measurement was performed at NPL on the same day. Alternatively a film reference measurement from another
centre was used, ensuring always appropriate post-exposure scanning times following the manufacturer's recommendations [153]. The film was irradiated with a dose as close as possible to the prescription dose level. The output measurement in the same conditions was used to calculate the exact dose delivered to the film.

During treatment plan deliveries, four test films (two sagittal and two axial) were irradiated at each centre. All films were returned to NPL and scanned at least 72 hours after exposure following the protocol (Appendix B) developed in Chapter 4 and using the scanner setup shown in Figure 7.7.

![Figure 7.7: Setup used for scanning audit test films.](image)

All films were handled following the methods described in Chapter 4. The film analysis was performed collectively on FilmQAPro at Queen Alexandra Hospital in Portsmouth after all audit visits were completed. One axial and one sagittal film was analysed from each centre using the methods described in Section 6.2 with the modifications described in Section 7.1.1. The regions of interest used for the analysis were a 6 x 5 cm
rectangle for the axial films and a 7 x 4 cm rectangle for the sagittal films. Passing rates were collected for the same criteria used in Section 6.2 for both red and green channels with TCC. The passing rates were also recorded with and without the film-dose linear scaling corrections applied.

### 7.2.9 Exradin W1 measurements

The Exradin W1 was positioned in its calibration slab and WT1 was used for build-up and back-scatter. Measurements were performed after the Semiflex output measurement, using the same setup conditions. The PSD was initially irradiated with approximately 10 Gy to encourage stability. Where a 40 x 40 cm beam was achievable by the delivery machine, the PSD was calibrated for CLR using the method described in Section 2.5.2. For CKs, GKS and TT where these conditions were not possible, the CLR factor was acquired in a beam of similar quality index either at NPL or another audited centre. The CLR factor was then applied to PSD measurements in reference conditions, in order to calibrate the detector for absolute dose. Equation 2.2 and the ionisation chamber dose measurement were used for defining the cross-calibration (Gain) factor. As different spherical phantoms were used in GK centres, it was not possible to perform measurements with the PSD in the reference conditions due to the lack of detector holders compatible with all phantoms. The PSD was therefore calibrated in the NPL Theratron 60Co beam to acquire suitable calibration factors, which were then applied to the dose verification measurements performed during the audits.

Two dose verification measurements were performed with the PSD in each audit: one in the brainstem and one in the target. The raw readings from the two electrometer channels were converted to dose using the calibration factors and Equation 2.3. Temperature corrections were applied when differences of more than 1 °C from calibration temperatures were observed. The dose measured by the PSD was compared to the mean dose calculated by the TPS in the PSD contour. The dose measured with the PSD was also compared to the first alanine pellet that was placed in the same relative position.
7.2.10 Alanine measurements

Two sets of four alanine pellets were irradiated in each centre: one in the target and one in the brainstem. The pellets were kept in their holders at all times during the audits. The pellets were only handled by the Chemical Dosimetry group at NPL during preparation and processing of the samples. The phantom temperature before and after each measurement was recorded and the mean temperature was used to apply the correction factors. All pellets were returned to NPL and were scanned within 1 month of the audit visit, ensuring negligible fading as the recommended time frame for scanning is less than three months post-irradiation. The measured doses were compared to the mean dose calculated by the TPS for each pellet structure. Alanine was considered the class detector for the audit measurements, as it is a reliable dosimeter which has proven its efficacy in small fields, provided by an established dosimetry service based at NPL.
Chapter 8

Results of the national audit for stereotactic radiosurgery

8.1 National SRS Audit - Results

The sixteen linac plans assessed adopted different methods for planning and delivery. With regards to the beam energy chosen, eight plans used 6 MV, three used 6 MV-SRS mode, three used 10 MV FFF and two used 6 MV FFF. A range of delivery techniques was also used: six centres used 3 Non-Coplanar Dynamic Conformal Arcs (NC-DCA), three centres used 8-9 Non-Coplanar Static Conformal Fields (NC-SCF), three centres used 3-5 Non-Coplanar VMAT (NC-VMAT), three centres used 1-2 Coplanar VMAT (C-VMAT) and one centre used 4 Non-Coplanar Circular Collimator Arcs (NC-CCA). Four different TPSs with six different calculation algorithms were used, all of which employed density heterogeneity corrections. Pencil Beam was the most commonly used algorithm followed by Collapsed Cone Convolution, Analytical Anisotropy and Monte Carlo.

In the GK group, five centres were equipped with a Perfexion model and two with an Icon, all of which use 192 $^{60}$Co sources. All centres used GammaPlan Version 10 or 11 with the TMR10 algorithm for dose calculation that does not apply density inhomogeneity corrections. Four centres used CT to generate skin contours and three used depth helmet measurements. The shot diameters used were comparable for all centres with the number of isocentres ranging from 16 to 22.

All CK centres were equipped with variations of the CK model 5. They all used the Ray Tracing dose calculation algorithm with heterogeneity corrections. The same nominal
beam energy was used by all CKs (6 MV FFF).

The TT centre used a HiART unit to deliver a helical treatment plan. The nominal beam energy used was 6 MV FFF and the dose was calculated using a Non-Voxel Broad Beam algorithm.

The prescription doses used across all delivery platforms ranged significantly, from 16 to 21 Gy. The same trend was also observed with the prescription isodose used. Prescription isodoses can be difficult to compare as some TPSs normalise to the isocentre whereas others normalise to the maximum dose, or the prescription dose. In order to enable comparisons between all treatment plans, the prescription isodoses described below are relative to the plan maximum dose. Prescription isodoses ranged from 43-90%, translated to a range of maximum doses inside the target of 19.9 - 41.7 Gy. The prescription isodoses used by the different equipment groups were: for GKs 43-50%, for 3 out of 4 CKs 65-70% (1 centre prescribed at 53%), for 15 out of 16 linacs 76-90% (1 centre prescribed at 52%), and for the one TT plan at 87%.

As there was only one centre that used TT in the audit, the findings may not be representative of a typical TT delivery. Therefore, in order to avoid misrepresentation of the modality but also to avoid identification of the centre, the TT plan was included in the linac group for the results that follow and for the rest of this thesis.

Table 8.1 summarises the equipment, techniques and prescription practices of the audit participants. The centres were grouped by platform in random order, different to the order shown in the results.
<table>
<thead>
<tr>
<th>No.</th>
<th>Platform</th>
<th>Energy</th>
<th>Technique</th>
<th>Coplanar/Non-Coplanar</th>
<th>Algorithm</th>
<th>Heterogeneity Corr.</th>
<th>Prescription Dose (Gy)</th>
<th>Prescription Isodose rel. to Dmax (%)</th>
<th>Dmax (Gy)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>LB Varian</td>
<td>10 MV FFF</td>
<td>Four DCA</td>
<td>NC</td>
<td>Pencil Beam</td>
<td>Yes</td>
<td>21</td>
<td>80</td>
<td>26.4</td>
</tr>
<tr>
<td>2</td>
<td>LB Novalis</td>
<td>6 MV</td>
<td>Nine SCF</td>
<td>NC</td>
<td>Pencil Beam</td>
<td>Yes</td>
<td>17.5</td>
<td>77</td>
<td>22.7</td>
</tr>
<tr>
<td>3</td>
<td>LB Varian</td>
<td>6 MV</td>
<td>Four CCA</td>
<td>NC</td>
<td>Circular Cone</td>
<td>Yes</td>
<td>18</td>
<td>76</td>
<td>23.6</td>
</tr>
<tr>
<td>4</td>
<td>LB Elekta</td>
<td>6 MV</td>
<td>Five VMAT</td>
<td>NC</td>
<td>Convolution</td>
<td>Yes</td>
<td>18</td>
<td>79</td>
<td>22.9</td>
</tr>
<tr>
<td>5</td>
<td>LB Novalis</td>
<td>6 MV SRS</td>
<td>Eight SCF</td>
<td>NC</td>
<td>Convolution</td>
<td>Yes</td>
<td>18</td>
<td>80</td>
<td>22.4</td>
</tr>
<tr>
<td>6</td>
<td>LB Elekta</td>
<td>6 MV</td>
<td>Three DCA</td>
<td>NC</td>
<td>Monte Carlo</td>
<td>Yes</td>
<td>18</td>
<td>90</td>
<td>19.9</td>
</tr>
<tr>
<td>7</td>
<td>LB Novalis</td>
<td>6 MV SRS</td>
<td>Four DCA</td>
<td>NC</td>
<td>Pencil Beam</td>
<td>Yes</td>
<td>18</td>
<td>78</td>
<td>23.1</td>
</tr>
<tr>
<td>8</td>
<td>LB Varian</td>
<td>6 MV</td>
<td>Four DCA</td>
<td>NC</td>
<td>Pencil Beam</td>
<td>Yes</td>
<td>18</td>
<td>79</td>
<td>22.8</td>
</tr>
<tr>
<td>9</td>
<td>LB Elekta</td>
<td>6 MV</td>
<td>Five VMAT</td>
<td>NC</td>
<td>Monte Carlo</td>
<td>Yes</td>
<td>21</td>
<td>80</td>
<td>26.3</td>
</tr>
<tr>
<td>10</td>
<td>LB Varian</td>
<td>10 MV FFF</td>
<td>Two VMAT</td>
<td>C</td>
<td>AAA</td>
<td>Yes</td>
<td>20</td>
<td>86</td>
<td>23.3</td>
</tr>
<tr>
<td>11</td>
<td>LB Novalis</td>
<td>6 MV SRS</td>
<td>Four DCA</td>
<td>NC</td>
<td>Pencil Beam</td>
<td>Yes</td>
<td>18</td>
<td>78</td>
<td>23.2</td>
</tr>
<tr>
<td>12</td>
<td>LB Varian</td>
<td>6 MV FFF</td>
<td>One VMAT</td>
<td>C</td>
<td>AAA</td>
<td>Yes</td>
<td>16</td>
<td>80</td>
<td>19.9</td>
</tr>
<tr>
<td>13</td>
<td>LB Varian</td>
<td>10 MV FFF</td>
<td>One VMAT</td>
<td>C</td>
<td>AAA</td>
<td>Yes</td>
<td>16</td>
<td>79</td>
<td>20.3</td>
</tr>
<tr>
<td>14</td>
<td>LB Elekta</td>
<td>6 MV</td>
<td>Eight SCF</td>
<td>NC</td>
<td>Convolution</td>
<td>Yes</td>
<td>21</td>
<td>89</td>
<td>23.5</td>
</tr>
<tr>
<td>15</td>
<td>LB Varian</td>
<td>6 MV</td>
<td>Five DCA</td>
<td>NC</td>
<td>Pencil Beam</td>
<td>Yes</td>
<td>18</td>
<td>79</td>
<td>22.7</td>
</tr>
<tr>
<td>16</td>
<td>LB Elekta</td>
<td>6 MV FFF</td>
<td>Three VMAT</td>
<td>NC</td>
<td>Monte Carlo</td>
<td>Yes</td>
<td>18</td>
<td>52</td>
<td>34.8</td>
</tr>
<tr>
<td>17</td>
<td>TT</td>
<td>6 MV FFF</td>
<td>Helical Tomo</td>
<td>C</td>
<td>NVBB</td>
<td>Yes</td>
<td>18</td>
<td>87</td>
<td>20.7</td>
</tr>
<tr>
<td>18</td>
<td>GK PFX</td>
<td>60Co</td>
<td>21 shots</td>
<td>NC</td>
<td>TMR10</td>
<td>No</td>
<td>18</td>
<td>43</td>
<td>41.6</td>
</tr>
<tr>
<td>19</td>
<td>GK PFX</td>
<td>60Co</td>
<td>17 shots</td>
<td>NC</td>
<td>TMR10</td>
<td>No</td>
<td>20</td>
<td>50</td>
<td>40.3</td>
</tr>
<tr>
<td>20</td>
<td>GK Icon</td>
<td>60Co</td>
<td>20 shots</td>
<td>NC</td>
<td>TMR10</td>
<td>No</td>
<td>18</td>
<td>49</td>
<td>36.7</td>
</tr>
<tr>
<td>21</td>
<td>GK Icon</td>
<td>60Co</td>
<td>16 shots</td>
<td>NC</td>
<td>TMR10</td>
<td>No</td>
<td>18</td>
<td>50</td>
<td>36.0</td>
</tr>
<tr>
<td>22</td>
<td>GK PFX</td>
<td>60Co</td>
<td>16 shots</td>
<td>NC</td>
<td>TMR10</td>
<td>No</td>
<td>18</td>
<td>46</td>
<td>39.1</td>
</tr>
<tr>
<td>23</td>
<td>GK PFX</td>
<td>60Co</td>
<td>18 shots</td>
<td>NC</td>
<td>TMR10</td>
<td>No</td>
<td>18</td>
<td>43</td>
<td>41.7</td>
</tr>
<tr>
<td>24</td>
<td>GK PFX</td>
<td>60Co</td>
<td>22 shots</td>
<td>NC</td>
<td>TMR10</td>
<td>No</td>
<td>18</td>
<td>43</td>
<td>41.4</td>
</tr>
<tr>
<td>25</td>
<td>CK</td>
<td>6 MV FFF</td>
<td>138 beams</td>
<td>NC</td>
<td>RayTracing</td>
<td>Yes</td>
<td>21</td>
<td>65</td>
<td>32.3</td>
</tr>
<tr>
<td>26</td>
<td>CK</td>
<td>6 MV FFF</td>
<td>123 beams</td>
<td>NC</td>
<td>RayTracing</td>
<td>Yes</td>
<td>18</td>
<td>70</td>
<td>25.7</td>
</tr>
<tr>
<td>27</td>
<td>CK</td>
<td>6 MV FFF</td>
<td>139 beams</td>
<td>NC</td>
<td>RayTracing</td>
<td>Yes</td>
<td>18</td>
<td>53</td>
<td>34.0</td>
</tr>
<tr>
<td>28</td>
<td>CK</td>
<td>6 MV FFF</td>
<td>109 beams</td>
<td>NC</td>
<td>RayTracing</td>
<td>Yes</td>
<td>20</td>
<td>65</td>
<td>30.8</td>
</tr>
</tbody>
</table>

Table 8.1: Equipment, techniques and prescription practices of centres that participated in the audit.
8.1.1 Output measurements

The output measurements performed by the centres and the auditors are presented in Figure 8.1. All measurements from both the auditors and the centres where within ±2.4%, ranging from -1% to 2.4%. The difference between centres and auditors was 0.5% on average with a maximum difference of 1.2%. In all cases the differences seen between auditor and centre measurements, were within the measurement uncertainties of the ionisation chambers used.

The linac group had the largest spread in the output measurements performed (3.2%) ranging from -0.8 to 2.4%. CKs had a smaller spread (1.8%) ranging from -1.0 to 0.8% and GKs had the smallest spread (1.6%) ranging from -0.9 to 0.7%.

Figure 8.1: Output measurements in local reference conditions for the 28 platforms that participated in the audit. An uncertainty of ±0.7% (k=1) on the auditors' measurement is indicated by the error bars. The “acceptable” tolerances of ±2% are indicated.

8.1.2 EBT-XD film measurements

The passing rates for the criteria used in Section 6.3.3 were recorded, for the green and red channels (with TCC), with and without dose-linear scaling. A section of the spreadsheet developed to record and analyse passing rates is shown in Appendix D with the results for three centres. Due to the large amount of data generated and the
restricted time for analysis, it was decided that for the purpose of this thesis and for the audit reports sent to the centres, only results for the conventionally used red channel (with TCC) would be analysed. The passing rates for two criteria (one LG and one GG), without dose-linear scaling applied, are shown in this section.

Figure 8.2 and Figure 8.3 show the passing rates for all centres for the criteria of 3% - 2 mm LG and 5% - 1 mm GG respectively.

Figure 8.2: Axial and Sagittal film passing rates for 3% - 2 mm Local Gamma.
The results showed comparable passing rates between axial and sagittal films. As expected, higher passing rates for all films were observed for GG criteria. For the 3% - 2 mm LG criterion, all but two films showed passing rates above 75%. For the 5% - 1 mm GG criterion, all but 2 films showed passing rates above 90%.

When the regions of interest used for the gamma analysis were reduced to smaller areas to include the target region only, passing rates improved significantly for all centres, showing very good agreement between TPS-predicted and delivered dose distributions. The majority of failed pixels for all films analysed were found to be outside the target, between the 2 Gy (threshold level) and the 12 Gy isodose line. This is demonstrated in Figure 8.4, which shows Film-TPS dose difference maps for the axial and sagittal films irradiated at Centre 1. In this example, the film is measuring higher doses than predicted by the TPS.
8.1. National SRS Audit - Results

8.1.3 Exradin W1 measurements

The measurements with the Exradin W1 PSD were compared against the alanine pellet placed in the same position but were also compared directly to the TPS-predicted dose. One linac centre was excluded from the analysis due to poorly controlled environmental conditions in the treatment room which resulted in substantial noise and leakage currents on the PSD measurement.

When compared to the alanine, PSD measurements in the target were within ±1.9% for linacs, CKs and TT. The spread of the differences observed was 3.4% (-1.5% to 1.9%). Linac and CK measurements showed spreads of 2.4% and 2.7% respectively. Much larger percentage differences were observed for target measurements on GKs with the PSD consistently under-responding in all measurements. The spread of these differences was 9.5% (-12% to -2.5%). Target dose measurements compared to the alanine are shown in Figure 8.5.
When PSD measurements were compared directly to the TPS-predicted dose, the differences observed doubled in magnitude and were within ±3.8% for linacs, TT and CKs. The spread of the differences observed for linacs was significantly larger in this comparison reaching 5.1% (-1.3% to 3.8%). CK and GK measurements showed similar spreads to those seen in the comparison with alanine, of 2.2% and 9.9% respectively. PSD measurements in GKs showed the same under-response observed in the comparison with alanine. Target dose measurements compared to the TPS-predicted doses are shown in Figure 8.6.
The measurements performed with the PSD in the OAR showed much larger disagreements in comparison to target measurements. These large disagreements were observed for both comparisons performed (to the dose measured by alanine and the dose predicted by the TPS). When compared to the alanine the deviations observed ranged from -20.7% to 57% and when compared to the TPS they ranged from -16.5% to 70.1%. CK measurements showed a consistent over-response of 5% to 57% when compared to the alanine and 12.1% to 70.1% when compared to the TPS. As observed with GK measurements in the target, under-response was seen again in GK measurements in the OAR with deviations ranging from -17.5% to -3% when compared to alanine, and -11% to -2.3% when compared to the TPS.

### 8.1.4 Alanine Measurements

Due to steep dose gradients in the distributions delivered in SRS, individual pellet measurements in the target showed differences of up to 14% when compared to the TPS-predicted mean dose in their individual contoured pellet structure. These deviations decreased substantially when the measured mean dose for the alanine pellet stack was compared to the mean dose predicted by the TPS for the whole stack of pellets.
It was therefore considered appropriate to compare mean doses for all pellets used in each stack, to account for positional uncertainties that can lead to large percentage differences between measured and predicted doses. Figure 8.7 shows the percentage differences found between the mean dose measured by the 4 pellets in the target compared to the mean dose predicted for these pellets by the TPS.

Figure 8.7: Alanine pellet measurements performed in the target during the audit. Platform groups are indicated in the legend. The mean for all centres was 0.7%. The dotted lines represent one standard deviation of the mean (±1.4%) and the dashed lines represent two standard deviations of the mean (±2.8%).

Linacs showed the largest spread in percentage difference of 5.2% (from -1.3% to 3.9%) with a mean of 0.5%. CKs measurements had a spread of 2.6% (from 1.4% to 4%), much smaller than linacs, with the highest mean difference in comparison to the other platforms (2.5%). GKs showed the smallest spread at 2.4% (from -0.8% to 1.5%) comparable to that of CKs, with the smallest mean percentage difference (0.4%) comparable to that of linacs.

TPS-predicted doses for the alanine pellets in the OAR ranged from 30 cGy up to 750 cGy. As OAR measurement were performed along a steep dose gradient, any positional uncertainties may be expressed as large dose differences. Moreover, due to the low doses delivered by some centres, which were below the calibration threshold of alanine, measurements had a higher uncertainty due to lower SNRs in the alanine readout. In order to overcome these problems whilst providing useful information to the audited
centres, it was decided to normalise the percentage difference between alanine and TPS to the 12 Gy dose level, a nominal brainstem tolerance dose used by many centres. This analysis enabled a dosimetric comparison whilst assessing the plan quality in terms of overdose to the OAR. Figure 8.8 shows the percentage difference between the mean dose measured by OAR alanine pellets the the TPS-predicted mean dose normalised to 12 Gy.

![Figure 8.8: Alanine pellet measurements performed in the organ at risk during the audit. Platform groups are indicated in the legend. The mean for all centres was 1.0%. The dotted lines represent one standard deviation of the mean (±1.2%) and the dashed lines represent two standard deviations of the mean (±2.4%).](image)

Similar trends were observed in the comparison of OAR alanine pellet measurements to those observed in target alanine pellet measurements. Linac measurements showed the largest spread at 4.6% ranging from -1% to 3.6% with a mean of 1.3%. CK measurements ranged between 0% to 1.9% with a mean of 0.9%. GKs had a similar spread to CKs (2%) ranging from -1.1% to 0.9% but with a lower mean of 0.1%.
8.2 National SRS Audit - Discussion

During the audit, 28 treatment plans were generated, all for the same realistic patient scenario of a single metastatic lesion located anterior to the brainstem. The approaches adopted by the UK SRS practitioners to treat the presented indication differed on many aspects. Some of these differences, with respect to the equipment, software and delivery techniques used, were previously identified [179]. Apart from these, probably the most influential and clinically relevant variation observed in the protocols assessed, was found in prescription practices. The linac group was found to have the most heterogeneous practices, compared to some small variations seen in CK practices, and more consistent practices seen in the GK group. All of the SRS approaches followed in the UK were assessed in this end-to-end audit.

The measurements performed with alanine pellets and Gafchromic EBT-XD film were considered the class detectors used in this work. Their use in the 28 SRS plans assessed enabled the analysis and comparison of all participating centres in terms of the accuracy achieved during the delivery. It also led to the production of audit reports that were sent to all participating centres. An example of one of the reports produced is shown in Appendix E.

The results also enabled the assessment of the PSD system’s performance for dose verification, via comparison with alanine pellet measurements and direct comparison with the TPS-predicted dose.

8.2.1 Output measurements

The majority of output measurements were within ±2%, a basic output acceptance range used by most of the centres visited, as recommended in IPEM 81 [200]. A few linac centres were operating using a higher acceptance range of ±3% and some GK centres were operating using a lower acceptance range of ±1%. Three measurements in total, two performed by the auditor and one by a centre, were above the 2% tolerance but less than 2.5%. In all three cases, the comparative measurement performed in the same conditions was within the 2% tolerance. Since the differences between comparative measurements of the auditors and the centres were within measurement
uncertainties, all platforms tested were considered acceptable with regards to absolute
dose in reference conditions.

On average, the auditors’ measurements were 0.5% higher than the measurements per-
formed by the centres. This is partly attributed to the fact that the auditors used a
PTW Semiflex chamber for all output measurements whereas the majority of centres
were using larger ionisation chambers. This could result in different levels of volume
averaging effects that may lead to a higher output measured by the smaller chamber,
especially in FFF beams where the dose profile presents a small peak in the centre of
the reference field. Evidence for this effect was seen when calculating the mean per-
centage difference between the two sets of measurements, separately for the 15 centres
using flattened beams and the 13 centres using FFF/SRS mode. In the 15 centres with
flattened beams, the mean percentage difference was 0.3%, with mean outputs of 1.001
and 1.004 for centres and auditors respectively. In the 13 centres with FFF/SRS mode
beams, the mean percentage difference doubled to 0.6%, with mean outputs of 1.002
and 1.008 for centres and auditors respectively.

Linac platforms showed a larger spread in output compared to the other platforms. The
reasons contributing to this may be related to the various reference conditions (SSD
and depth) used between linac centres, whereas CKs and GKs use the same reference
conditions within their equipment groups. Other factors contributing to this may be
related to the fact that most linacs tested were versatile, multi-modality platforms,
with more moving components and perhaps more prone to dosimetric uncertainties, as
opposed to single energy platforms like the CK or platforms with stable decaying sources
like the GK. The larger spread may also be partly attributed to the larger linac sample
group participating in the audit. However, it should be noted that the well-documented
Cobalt-60 source output of GKs and their simpler design are probably responsible for
the smaller output deviations seen (better than ±0.9%) in that equipment group.

8.2.2 EBT-XD film measurements

Film measurements were performed in all plans and analysed using red and green TCC
gamma analysis methods. The differences seen in passing rates between red and green
colour channels were on average 2.3%. As both methods showed similar passing rates it
was decided to adopt the more commonly used red colour channel with TCC. However, the red channel appeared to be affected by dose-linear scaling corrections more than the green channel. On the other hand, dose linear scaling did not have a significant impact on the majority of films assessed and it was therefore decided not to apply this correction.

The passing rates recorded showed good agreement with the predicted dose distributions for both sagittal and axial films. All treatment modalities showed comparable variations in passing rates between the centres assessed and the passing rates alone do not suggest significant differences between the different equipment groups. It is important to note that gamma index analysis is not ideal for direct comparison between the SRS deliveries assessed due to differences in dose gradient and maximum dose in the measured distributions. LG criteria may favour linac centres in which the dose gradient could be less steep than GK and CK, and GG criteria may favour GK centres in which the maximum (normalisation) dose is higher. Whilst the methodology for film analysis was designed to diminish sensitivities to different dose distributions it is impossible to achieve this with gamma index analysis. Gamma passing rates are also sensitive to the position of the film relative to the dose plan, the position of the region of interest used for the analysis and the 2 Gy threshold levels applied. Further work needs to be conducted to develop an analysis method that is less sensitive to these dose distribution differences, enabling a more reliable direct comparison between competing plans.

Despite its pitfalls, the analysis method used enabled quantification of the dose shaping abilities of all SRS platforms active in the UK. The results showed good performance by all platforms with noticeable dosimetric differences outside the target volume (See Figure 8.4). These dosimetric inaccuracies, seen in the majority of centres, are related to the TPSs limitation in simulating out of field doses and in most cases resulted to underestimation of doses to the OAR. Nevertheless, the findings of the film results suggest clinically acceptable dosimetric and geometric accuracies. However, as films were optimally matched to the TPS dose planes, it may be argued that the results are not representative of positional uncertainties in the delivery. Due to the steep dose gradients present in SRS, even small positional deviations can result in high dosimetric differences that could be detrimental in clinical deliveries. Further analysis with the
film measurements acquired during the audit will enable assessment of the positional 
deviations of all plan deliveries.

8.2.3 Exradin W1 measurements

The Exradin W1 PSD was used for dose verification measurements in all 28 SRS plans. 
Two measurements were performed in each plan, one in the target region and one in 
the OAR. The performance of the PSD was evaluated by comparison to the alanine 
measurements in the same region and by direct comparison to the TPS-predicted dose. 
The PSD demonstrated good agreement with both alanine and TPS in target dose 
measurements for Linacs and CKs, but showed poor agreement for GKs. The consistent 
under-response shown by the detector in GKs is attributed to the different calibration 
procedure performed for these centres. It is perceived that the difference in source 
geometry between GKs and the Theratron unit used for calibrating the PSD is the 
major contributing factor to the under-response observed. These differences will be 
expressed in the Gain factor and further work involving measurements with the PSD 
and Semiflex chamber in the GK reference field is needed to confirm this. Another, 
smaller, contributing factor in this under-response for GK measurements, is the longer 
measurement acquisition time that may contribute to larger (compared to linac and 
CK) leakage currents occurring during the course of the measurement. In linacs and 
CKs it was possible to calibrate the detector in the treatment beam and acquire accurate 
Gain factors for these deliveries. The results demonstrate that the PSD is a suitable 
tool for SRS dose verification, within the target volume, for Linac and CK deliveries.

Unexpectedly, much larger disagreements were observed for out-of-field measurements, 
when the PSD was placed in the OAR. These are partly attributed to the steep dose 
gradient present at the point of measurement and the lower SNRs observed in these 
measurements. Despite these, the large disagreements seen cannot be fully attributed to 
these factors. The perceived contributing factor is related to the out-of-field stem effect 
and the CLR correction applied for its removal. The electron energy cut-off threshold 
for Cherenkov production in PMMA is approximately 150 keV. It is anticipated that 
depending on the location of the PSD in an SRS delivery, during the course of the 
measurement, the fibre will be irradiated by varying electron spectra, both in terms
of angular distribution and energy. This in turn, may cause variations in Cherenkov production as the proportion of electrons above the 150 keV threshold can vary significantly during the course of the measurement. It is therefore hypothesised that as the fibre is moved further away from the target, into the OAR, the Cherenkov production in the PMMA optical fibre (stem of PSD) will be significantly different and difficult to predict. Considering the above, it is clear that a constant CLR factor for the removal of stem signal is inadequate for acquiring accurate out-of-field measurements using the chromatic removal method. Further work focusing on the development of improved instrumentation for PSDs will enable the reduction or complete removal of Cherenkov “noise” contributions.

8.2.4 Alanine measurements

The mean dose measured by each set of alanine pellets was compared to the TPS-predicted dose in the target and OAR. Good overall agreement with the TPS was observed with three centres falling outside (above) two standard deviations of the mean (two centres in the target dose measurements and one in the OAR measurements).

Linacs showed the largest spread in comparison to GKs and CKs. This is not surprising considering that the CK and GK centres visited had almost identical equipment, software and prescription practices within their sub-groups. On the other hand, the linac centres visited used a range of beam energies, delivery techniques, TPSs calculation algorithms and prescription practices. Linac practices appear to be influenced by the equipment and software available, clinician preferences and influences from the local radiotherapy practices.

CK measurements in the target, showed that the TPS (Multiplan) under-estimated the dose in all four centres visited. This finding is in agreement with another study utilising the same alanine service for target dose measurements in CK plans [111]. Interestingly, one CK centre (centre number 8) prescribed at a much lower prescription isodose (53%) than the other CK centres (65-70%), due to the clinician’s preference who had previous GK experience. The alanine measurements in the target were found to have a percentage difference to the TPS that was 2% higher than the mean of the other 3 CK centres. This may be related to MultiPlan’s limitations in calculating the
dose when lower prescription isodoses are chosen, but more investigations are required to verify this.

Alanine measurements in GK centres showed good agreement with the TPS (Gamma-Plan). This was somewhat unexpected as the GammaPlan algorithm (TMR10) does not take into account density inhomogeneities and assumes water density within the skull contour. Moreover, the patient (skull) contour can be less accurate in comparison to other SRS practices as it may be generated by interpolating between multiple depth helmet measurements. A recent study [201] investigated the GammaPlan convolution algorithm (employs density heterogeneity corrections) for 50 clinical plans by assessing the change in beam-on time. Their results showed a 6% difference between the two algorithms, with the TMR10 having lower beam-on times. This study also found that about 1.5% was attributed to depth helmet measurements. On the contrary, the results of this work suggest that these factors do not contribute significantly to dosimetric inaccuracies. Further work investigating the GammaPlan convolution algorithm using the audit methodology can provide conclusive evidence to understand the discrepancies seen between these two studies.

Despite the lower SNRs observed in the OAR measurements the results showed good agreement with the TPS. All pellets measured doses below 7.5 Gy and 86% of pellets measured doses below 5 Gy. The majority of pellets were therefore exposed to dose levels below the acceptable level previously operated by the NPL Chemical dosimetry service. The Chemical dosimetry group performed additional measurements at lower doses to extend the calibration range to lower doses. The results of the audit suggest that reliable alanine measurements can be performed in clinical plans at out-of-field dose levels below 5 Gy.

Finally, it should be noted that the mean and standard deviations indicated in Figures 8.7 and 8.8 are statistical tools used to describe and compare the distribution of the data and should not be interpreted as clinically acceptable tolerances.
8.3 National SRS Audit - Conclusions

This national SRS audit has provided a means of comparison between all UK SRS practices for the same realistic patient scenario whilst enabling assessment of three detector systems in SRS dosimetry. The results provide a useful record of detector performance in SRS that can be utilised to optimise these systems further for plan dose verification. As far as the audit results are concerned, the dosimetric performance of all centres is considered clinically acceptable, despite the tendency for poorer agreement observed in the low dose regions. However, concerns are raised regarding the variations seen in clinical practice. These variations are more prominent in the linac subgroup which raises the need for steps to be taken towards standardisation and consensus.
Chapter 9

Conclusions and Further Work

9.1 Conclusions

With reference to the initial objectives defined in Section 1.6, this thesis has shown the investigations conducted to identify and understand current SRS practices in the UK. These investigations helped to develop a better understanding of the complexities involved in performing accurate dosimetry in clinical SRS plans. Subsequently, novel detector systems were characterised and tested for suitability in SRS dosimetry. These were then incorporated into the development of a novel methodology for the end-to-end assessment of stereotactic radiosurgery. The methods adopted showed consistency and reliability to a large extent, but areas of improvement and optimisation in their performance were also identified. The methods were used to evaluate SRS practices in the UK and have provided a snapshot of their dosimetric and geometric accuracy. Areas of standardisation in SRS have been identified through the course of this work and they are outlined in this chapter.

9.1.1 Dosimetric accuracy

Dosimetric differences of up to 4% of the TPS-predicted dose have been observed in the target region and even larger differences have been observed outside the target. These results have been corrected for output variations of the delivery unit, but if these corrections are not applied the deviations would, in some instances, be slightly larger. The need for stringent quality assurance procedures to ensure consistency in clinical deliveries is therefore highlighted. However, on a national scale, the clinical impact of the dosimetric differences observed in the audit is more than likely negligible.
Considering the variations seen in prescription dose and prescription isodose, the dose delivered to patient is subject to much larger differences depending on the protocol used for treatment. As mentioned previously in Section 8.1, prescription doses for the presented indication varied from 16 to 21 Gy and maximum doses inside the target varied from 19.9 to 41.7 Gy. This range is translated to differences of up to 24% in prescription dose and up to 52% in maximum dose. Therefore, the need for more consistent prescription practices is also highlighted.

### 9.1.2 Geometric accuracy

The dose distribution shaping abilities of all SRS systems in the UK were also tested and their performance has been quantified. Geometric accuracy is imperative in SRS, and arguably more important than dosimetric accuracy. Due to the steep dose gradients inherent to radiosurgical deliveries, a small geometric inaccuracy may result in significant under-dose to the target or significant over-dose to an OAR. This is extremely important in clinical cases where OARs are abutting or even surrounding the target volume. The results of this work showed that all centres were able to achieve acceptable agreement between predicted and delivered dose distributions. Nevertheless, the positional accuracy of the dose distributions measured was not thoroughly assessed in this work and would benefit from further analysis. Moreover, the evaluation methods that were available for this analysis are sensitive to the dosimetric differences between the deliveries. Therefore, a direct comparison of the geometric accuracy of competing platforms was not possible.

### 9.1.3 Treatment Plan Quality

The collection of treatment plans generated by the centres that participated in the audit was assessed by various plan metrics\(^1\). Another similar study was performed by the RTTQA group, in which centres were asked to submit a treatment plan for six different clinical scenarios. Both these studies showed that there were large variations in conformity and dose fall-off outside the target for the treatment plans assessed. The differences observed are related to the techniques and prescription practices used.

\(^1\)This work was performed by Shaik M Usman Ghouse Mohiuddin as part of his MSc dissertation.
There are currently no universal guidelines for acceptable treatment plan quality in radiosurgery and their importance is emphasised by the large variations observed.

Another major variation in radiosurgery planning practice is the use of a margin around the treatment volume to account for treatment uncertainties. This practice is typically not followed by GK radiosurgery centres but is common in linac-based radiosurgery. In this work, the use of margins was not assessed, in an attempt to produce comparable treatment plans by all participating centres. However, these differences should be investigated further with the aim of achieving standardised practise.

9.1.4 Biologically Effective Dose

The concept of Biologically Effective Dose (BED) was proposed in 1989 and has since then been extensively used in fractionated radiotherapy studies [202]. This concept was quickly adopted in radiosurgery [203] but has not received as much attention since then. BEDs in radiosurgery are susceptible to different influences, including the technique used and the length of the delivery [204].

In the scenario presented in the audit, both the techniques and length of treatments varied significantly. The quickest delivery was performed using a single VMAT arc that was delivered in less than 2 minutes and the longest treatment was a 12 shot GK plan that took over 2 hours to deliver. The differences in BED between these two treatments are not fully appreciated and they are currently not considered at all in clinical practice. BED variations should receive more attention to allow their introduction into the clinic in order to optimise treatments and achieve better outcomes.
9.2 Further work

9.2.1 Dosimetric accuracy

It is essential that future trials steer their focus on developing more consistent prescription practices for the management of brain metastasis using stereotactic radiosurgery. This should develop a cohesive national approach that would make it possible to compare patient outcomes and enable optimisation of SRS treatments. This, in combination with reliable clinical deliveries, could guarantee improvements in tumour control and complications for patients receiving SRS.

9.2.2 Geometric accuracy

Due to inherent differences in the coordinate systems between treatment platforms and compatibility issues with some TPSs it was not possible to assess positional deviations in the delivered isodoses. Further work in the development of a stand-alone software could enable this assessment and provide valuable comparisons between the different delivery platforms. This work may also provide an assessment of the accuracy of various immobilisation devices. As these film measurements were performed in a phantom, patient variations can be factored out, and the results will be representative of the accuracy associated with the immobilisation and delivery equipment.

9.2.3 Dosimetry systems

This thesis has presented the use of three different dosimetry systems for dose verification. The detectors demonstrated suitability for SRS measurements and were used to perform semi-3D plan dose verification. Furthermore, two detectors of different geometries, alanine and scintillator, were used interchangeably to allow comparisons of the absolute dose measured in clinical plan deliveries.

Alanine dosimetry was found to be a suitable reference dosimeter for audit purposes with reliable performance, not only for measurements above 10 Gy, but also for measurements in the region of 2 - 10 Gy. With further optimisation and automation in the labour intensive read-out process, alanine will become even closer to the ideal dosimetry system.
EBT-XD film was found to be an excellent dosimeter for the assessment of high-dose and steep-gradient dose distributions seen in SRS. The less commonly used green channel produced results comparable to the red channel method. However, the green channel appeared to be less susceptible to post-irradiation darkening and should be investigated further as it may present a favourable method for audit or postal dosimetry where consistent scanning times are not always possible.

The film analysis revealed weaknesses in the local and global (maximum) gamma index method for comparing the performance of competing plans with different dose-gradients and maximum doses. Alternative analysis methods may allow more meaningful comparisons. For example, global gamma analysis normalised to the prescription dose should be less susceptible to differences in maximum dose. Also, as mentioned previously, the positional accuracy of dose distributions was not fully assessed with the analysis method in place. Further work should be conducted to develop a stand-alone method purposed to assess the positional deviation of an SRS delivery.

The Exradin W1 plastic scintillator performed reliable measurements inside the target volume but requires refinement to improve its performance both inside and outside the target. Key objectives in its refinement should be focused on developing less sensitive electronics and improving the Cherenkov removal methods. The main limitation of the chromatic removal method used in this work is that it assumes constant CLR and Gain factors during a clinical delivery, which is not the case in most SRS deliveries. There are however, some promising approaches for removing Cherenkov noise more efficiently. One of these relies on using red emitting scintillators to further separate the two spectra and achieve a more efficient chromatic removal at the expense of detector water-equivalence [205]. The development of water-equivalent red emitting scintillators may have an important role to play in scintillation dosimetry. Other promising approaches in removing Cherenkov noise rely on multiple photodiodes, CCDs or spectrometers to perform multi-spectral and hyper-spectral analysis of the light emitted by scintillation systems [206]. Such approaches will produce more accurate measurements in clinical deliveries where varying levels of Cherenkov and scintillation are emitted in the course of a measurement.
During the audits conducted, TLD measurements of the dose to the eyes of the phantom were also performed. These were performed using a 3D-printed plastic goggle-shaped insert. The insert was manufactured to accommodate small plastic TLD holders with three types of TLDs: Ge-doped optical fibres, glass beads and Lithium Fluoride (TL-100). The goal of this study is to compare the performance these TLDs in the measurement of typical lens doses from CT scans and SRS treatments. Further work in analysing these TLD measurements will provide evidence of the typical doses received during these procedures.

9.2.4 Standardisation in SRS

SRS practice in the UK, at its current state, would benefit from standardisation. The initial focus of this standardisation should be aimed at prescription doses and prescription isodoses. These need to be more cohesive in order to facilitate clinical trials for producing consistent data, allowing the investigation of treatment effectiveness through patient outcomes.

As practice variations are more prominent in the linac subgroup, it would be beneficial if standardisation was introduced in this group first. In the audit conducted, there have been examples of CK and linac centres following GK inspired practices. This is indicative of other equipment subgroups acknowledging value in GK planning practices. Out of 28 centres, only 4 used a prescription isodose above 80%. It is now widely acceptable that dose homogeneity within the target is not essential for radiosurgery and there is no evidence to suggest that it is unsafe. Therefore, a reasonable and effective point of standardisation, ideally introduced immediately, would be to limit centres in only using prescription isodoses between 40-80% of the maximum dose. This should be followed up closely and subject to further revision, possibly in narrowing the range to 40-60%. The recommendations should also allow for deviations from this practice when a target volume is small (below 0.5 cm³) where a steep dose gradient is not achievable or even important. Such an introduction should show improvements in normal tissue toxicity whist steering towards consistent prescription practices. Subsequently, prescription doses can be revised to produce recommendations for appropriate clinical protocols.
9.2. Further work

Treatment plan quality is another area of SRS that would benefit from standardisation. This is mainly assessed by target conformity and dose gradient outside the target, both of which can be scored using various indices [85, 86, 198, 207, 208]. There are also a number of treatment planning parameters (volume of 12 Gy and 10 Gy) that have shown correlation with patient complications and can be used as prognostic factors or thresholds for improving treatment plan quality [209, 210]. It was noted that only 4 out of 28 centres that participated in the audit used coplanar delivery techniques. It is anticipated that centres are utilising non-coplanar deliveries to benefit from improvements in conformity, gradient and reduction in the volume of 10 and 12 Gy. Therefore another appropriate step for standardisation would be to restrict the use of coplanar SRS. There are currently no SRS-specific guidelines in the UK recommending minimum acceptable plan quality. Following the recent commissioning of SRS provision in England, it is now an appropriate time to develop such documentation to ensure high quality treatments throughout the country. This effort would also be supported with the development of plan quality indices that are intuitive and practical for all subgroups to calculate.

Treatment margins have not been assessed in this study and the results would have been more difficult to interpret if this variable was included. Their clinical impact on treatment effectiveness and treatment complications in SRS is still questioned and should be examined further. However, this assessment will not be possible if the variations outlined above are not tackled first.

Finally, another variable in SRS practice is treatment technique and the length of treatments, which both have a significant impact on the BED. Further work should be undertaken in radio-biological experiments and modelling to understand how these factors impact treatment effectiveness. Moreover, once treatment practices become more consistent, patient data could be analysed retrospectively to determine correlations between BED and patient outcomes. The lessons learned from these studies should then be translated to the clinical environment for further optimisation of SRS treatments.
9.3 Closing remarks

Improvements in dosimetry systems, dose verification methods, audit and QA will increase the confidence of the medical physics community in delivering highly focused radiation treatments. These efforts will be in vain if they are not accompanied by efforts for achieving consistent prescription practices, consistent planning philosophies, consistent treatment plan quality, consensus with regards to treatment margins and BED-optimised treatments. However, if these efforts are synergistic, the next human generation will witness extraordinary cure rates with negligible complications, not only for malignancy but also for benign tumours, functional disorders using radiosurgical lesioning or thalamotomy, even mental disorders using psycho-radiosurgery.

‘ἐν οἴδα ὅτι οὐδὲν οἴδα’

Σωκράτης
Bibliography


[88] L. Olsson, J. Arndt, A. Fransson, and B. Nordell, “Three-dimensional dose map-


[159] A. Beddar, T. Mackie, and F. Attix, “Water-equivalent plastic scintillation detec-


[192] Y. Kamio and H. Bouchard, “Correction-less dosimetry of nonstandard photon


Appendix A

Questions included in the survey questionnaire (Section 3.2)

• 1. What is your centre’s experience with Cranial Stereotactic Radiosurgery?
• 2. Estimate how many Cranial SRS patients your centre has treated
• 3. State the number of machines in your centre used clinically for SRS
• 4. What make and model are the machines that you use for Cranial SRS?
• 5. Estimate on average how many new Cranial SRS patients your centre treats each month
• 6. Does your centre wish to expand its current SRS program in the future?
• 7. Does your centre limit the number of new Intracranial SRS cases starting each week/month?
• 8. Which technique for Cranial SRS does your centre perform? (multiple answers permitted)
• 9. Which photon energies are you using for Cranial SRS?
• 10. Do you use flattening filter free beams for SRS clinically at present?
• 11. Are you using any of the following beam collimating systems? (list of options available)
• 12. On average, how long does it take to complete the delivery of a Cranial SRS patient treatment? (Time from the point the patient lies on the treatment couch to the point he/she sits up)
• 13. For the pathologies listed below, please select the number of cases you treat per month for each indication
• 14. Which imaging modalities are used for Outlining?
• 15. Which imaging modalities are used for Planning?
• 16. Which software package do you use for volume delineation / outlining?
• 17. Which tissues are normally outlined?
• 18. On average, how long does it take to delineate all the necessary structures for planning?
• 19. Which Treatment Planning System (TPS) and which algorithms do you use for SRS planning?
• 20. On average, how long does it take to complete a plan for SRS?
• 21. What is the most common isodose distribution that you prescribe to?
• 22. How often are patient-specific QA measurements performed for SRS?
• 23. If patient-specific QA measurements are not performed regularly, please state primary reasons for this
• 24. If patient-specific QA measurements were reduced after being performed for every plan, after how many plans did you decide to reduce the workload?
• 25. Do you use any phantom(s) for measurement-based verification/QA of Cranial SRS?
• 26. Do you use any detector(s) for Cranial SRS verification?
• 27. On average, how long does it take to complete your measurement-based verification/QA procedure?
• 28. Which immobilisation systems do you have in place for Cranial SRS?
• 29. Do you use any pre-treatment imaging for localisation?
• 30. Do you use any imaging during treatment (between fields) for localisation?
• 31. What is the action level below which setup accuracy would be considered acceptable?
Appendix B

Protocol followed for film handling and scanning

1. Prepare a clean surface and wear latex gloves. It is important to keep films clean, dust-free, scratch-free and avoid bending it. Keep films in a controlled environment at room temperatures and low moisture.

2. Prepare a black light-tight envelope for placing the film inside after cutting it. It is important to minimise exposure to ambient light.

3. Take a sheet (25.4 x 20.3 cm) of EBT-XD film out of the packet and measure the piece you would like to cut. Ensure that the maximum width of the film is less than 6 cm. (Note: the orientation used for calibration, had the long side of the film sheet parallel to the scanning direction - ensure that all film samples follow this orientation too).

4. Using a thin permanent marker, mark dotted lines along the edges of the sample that will be cut and use a guillotine or sharp scissors to cut the film in order to avoid delamination along the edges.

5. Avoid placing films on top of each other as they can cause scratches. If more than one samples are placed in an envelope, use a piece of paper to separating them and minimise friction between them.

6. When setting up films for irradiation avoid friction between films and phantoms. Any film not intended for irradiation must be kept outside the bunker and away from scatter radiation.

7. After irradiation, place films in the light-tight envelope and store in a safe location at standard room temperatures. Allow 72 hours for post-irradiation darkening to occur.

8. Before scanning the film, switch on the scanner and allow at least ten minutes for it to acclimatise. Perform 5 warm-up scans at 96dpi for the full length of the scanner. Scan the films within a few minutes of the warm-up scans. If the scanner has been idle for a long time perform warm-up scans again.
9. Place the positioning jig on the scanner bed and align it with the markings on the scanner axis.

10. Thoroughly clean the scanner bed to remove any dust. Clean the test films and glass compression plate too.

11. Position test films within the frame of the positioning jig and in the centre of the scanner. Rest the compression plate above the film and check that it has not moved and that it is flat against the scanner.

12. Close the scanner lid and perform a preview scan. Select a rectangular scanning area aligned with the edges of the film.

13. The scanner should be in transmission mode, the resolution should be 96 dpi, colour and sharpness corrections should be deactivated and the image should be saved in 48 bit tiff format.
Appendix C

Contouring for SRS audit

Figure C.1: Diagram with schematic representation of contoured structures required for audit
Appendix D

Film analysis results for three centres

### Results with dose linear scaling

<table>
<thead>
<tr>
<th>CENTRE</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>Mean</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>Mean</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>Mean</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>12.3</td>
<td>12.4</td>
<td>12.5</td>
<td>12.6</td>
<td>12.6</td>
<td>12.7</td>
<td>12.8</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
</tr>
<tr>
<td></td>
<td>13.0</td>
<td>13.1</td>
<td>13.2</td>
<td>13.3</td>
<td>13.3</td>
<td>13.4</td>
<td>13.5</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
</tr>
<tr>
<td></td>
<td>14.7</td>
<td>14.8</td>
<td>14.9</td>
<td>15.0</td>
<td>15.0</td>
<td>15.1</td>
<td>15.2</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
</tr>
<tr>
<td></td>
<td>16.4</td>
<td>16.5</td>
<td>16.6</td>
<td>16.7</td>
<td>16.7</td>
<td>16.8</td>
<td>16.9</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>CENTRE</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>12.6</td>
<td>12.7</td>
<td>12.8</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
</tr>
<tr>
<td></td>
<td>13.3</td>
<td>13.4</td>
<td>13.5</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
</tr>
<tr>
<td></td>
<td>15.0</td>
<td>15.1</td>
<td>15.2</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
</tr>
<tr>
<td></td>
<td>17.0</td>
<td>17.1</td>
<td>17.2</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
</tr>
</tbody>
</table>

### Results without dose linear scaling

<table>
<thead>
<tr>
<th>CENTRE</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>Mean</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>Mean</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>Mean</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>D2 DAY</th>
<th>Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>12.3</td>
<td>12.4</td>
<td>12.5</td>
<td>12.6</td>
<td>12.6</td>
<td>12.7</td>
<td>12.8</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
</tr>
<tr>
<td></td>
<td>13.0</td>
<td>13.1</td>
<td>13.2</td>
<td>13.3</td>
<td>13.3</td>
<td>13.4</td>
<td>13.5</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
</tr>
<tr>
<td></td>
<td>14.7</td>
<td>14.8</td>
<td>14.9</td>
<td>15.0</td>
<td>15.0</td>
<td>15.1</td>
<td>15.2</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
</tr>
<tr>
<td></td>
<td>16.4</td>
<td>16.5</td>
<td>16.6</td>
<td>16.7</td>
<td>16.7</td>
<td>16.8</td>
<td>16.9</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
<td>17.0</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>CENTRE</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
<th>Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>12.6</td>
<td>12.7</td>
<td>12.8</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
<td>12.9</td>
</tr>
<tr>
<td></td>
<td>13.3</td>
<td>13.4</td>
<td>13.5</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
<td>13.6</td>
</tr>
<tr>
<td></td>
<td>15.0</td>
<td>15.1</td>
<td>15.2</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
<td>15.3</td>
</tr>
<tr>
<td></td>
<td>17.0</td>
<td>17.1</td>
<td>17.2</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
<td>17.3</td>
</tr>
</tbody>
</table>

---

Figure D.1: Gamma passing rates recorded for three centres
Appendix E

Example of an Audit Report

Report on National Cranial Stereotactic RadioSurgery Dosimetry Audit

A collaboration between:

Royal Surrey County Hospital NHS
Portsmouth Hospitals NHS Trust
NPL RTTOA
National Physical Laboratory

Centre: [Redacted]
Delivery Platform / Planning systems: Gamma Knife / GammaPlan
Technique / Energy: Non-coplanar static fields / Cobalt

Date of visit: 20th April 2016
Local Hospital Staff: [Redacted]
Auditors: Alexis Dimitriadis & Rushil Patel
Alamene pellets processed and checked by: Clare Gouldstone & David Crossley – NPL
Film results processed and checked by: Alexis Dimitriadis & Antony Palmer
Report compiled by: Alexis Dimitriadis
Report checked by: Catharine Clark

MPE signature: [Signature]
MPE name: Catharine Clark
Date: 17th August 2016

Figure E.1: Example of an Audit report - Page 1 of 8.
Description of audit procedure:

**Immobilisation:** A visit was conducted to perform an end-to-end audit for intracranial stereotactic radiosurgery. A STEEV phantom (CIRS, Norfolk, VI, USA) was provided which had to be immobilised following the local protocol for a single brain metastasis treatment. The phantom was scanned with a series of detectors in situ. An additional scan of the phantom with the target and organ at risk volumes was provided in DICOM format and the two scans were imported into the Treatment Planning System (TPS).

**Planning:** The two scans of the phantom were co-registered in the TPS. The detectors were contoured on the local scan and the target volume and brainstem were contoured on the scan provided (the target was an irregularly-shaped, ~9-9 cc single brain metastasis, located centrally in the brain and 1 cm anterior to the brainstem). The plan was generated following the local protocol and exported to the treatment delivery platform.

**Treatment delivery:** Basic output checks were performed by the centre, according to local practice, and independently by the auditors. Dosimetric assessment of the delivery was performed with alanine pellets and Gafchromic EBT-XD film (results below). Measurements were also performed for research purposes with an Exradin W1 plastic scintillation detector and thermoluminescent dosimeters which are not included in this report.

**Analysis:** The alanine pellets irradiated during the audit were scanned using Electron Paramagnetic Resonance by the NPL Chemical Dosimetry group and a report was provided to the auditors with the results. The dose difference between TPS predicted and measured doses in the volume of each pellet was calculated. The dose to the pellets in the brainstem are reported relative to a 12Gy nominal tolerance dose for the brainstem.

The EBT-XD Gafchromic films irradiated during the audit were scanned on an EPSON Expression 10000XL scanner at NPL and analysed on FilmQA Pro software (Ashland ISP Advanced Materials, NJ, USA) at Queen Alexandra Hospital, Portsmouth. They were compared to high resolution (1 mm grid) predicted dose planes exported from the TPS. A selection of gamma passing rates suitable for SRS plan analysis were chosen. The gamma passing rates presented were collected using the red colour channel with triple channel dosimetry correction for the regions analysed (50 x 50 mm axial and 70 x 40 mm sagittal), with a cut-off threshold for doses below 200 cGy.

![Figure 1: Annotated photograph of the STEEV phantom with the inserts and detectors used in the audit.](image)

Figure E.2: Example of an Audit report - Page 2 of 8.
Figure 2: Sections of the phantom through axial, coronal and sagittal planes with the target insert (2a) and the dosimetry insert (2b) inside the cavity. The axial and sagittal sections shown in 2b are the planes where the films were placed.

Figure 3: Schematic representation of the sagittal plane through the middle of the phantom showing all detectors on the PTV and OAR. The scintillation detector positions are superimposed on the interchangeable alanine pellet positions (both are cylindrical detectors - dimensions of scintillator: 3mm length x 1mm diameter, dimensions of pellets: 2.5mm length x 5mm diameter).

Figure E.3: Example of an Audit report - Page 3 of 8.
RESULTS

Basic Output results:

Output measured by centre: 219.7 cGy/min – 1.000 (normalised output)
Output measured by auditors: 219.7 cGy/min – 1.000 (normalised output)
Difference: 0.0%

Alanine Pellet results:

<table>
<thead>
<tr>
<th>Location</th>
<th>Pellet No.*</th>
<th>Measured (cGy)</th>
<th>Output Corrected (cGy)</th>
<th>Predicted (cGy)</th>
<th>Relative difference with output corrected (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Target</td>
<td>2861.1</td>
<td>3076.7</td>
<td>3076.7</td>
<td>3040.0</td>
<td>1.2%</td>
</tr>
<tr>
<td></td>
<td>2862.2</td>
<td>3284.5</td>
<td>3284.5</td>
<td>3230.0</td>
<td>1.7%</td>
</tr>
<tr>
<td></td>
<td>2863.3</td>
<td>3247.6</td>
<td>3247.6</td>
<td>3210.0</td>
<td>1.2%</td>
</tr>
<tr>
<td></td>
<td>2864.4</td>
<td>3181.5</td>
<td>3181.5</td>
<td>3000.0</td>
<td>-8.6%</td>
</tr>
<tr>
<td>Mean of all pellets</td>
<td>3197.5</td>
<td>3197.6</td>
<td>3197.6</td>
<td>3195.0</td>
<td>0.1%</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Location</th>
<th>Pellet No.*</th>
<th>Measured (cGy)</th>
<th>Output Corrected (cGy)</th>
<th>Predicted (cGy)</th>
<th>Absolute Difference with output corrected (cGy)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Brainstem</td>
<td>2860.1</td>
<td>334.9</td>
<td>334.9</td>
<td>330.0</td>
<td>4.9</td>
</tr>
<tr>
<td></td>
<td>2860.2</td>
<td>303.9</td>
<td>303.9</td>
<td>300.0</td>
<td>3.9</td>
</tr>
<tr>
<td></td>
<td>2860.3</td>
<td>291.5</td>
<td>291.5</td>
<td>290.0</td>
<td>1.5</td>
</tr>
<tr>
<td></td>
<td>2860.4</td>
<td>279.6</td>
<td>279.6</td>
<td>270.0</td>
<td>9.6</td>
</tr>
<tr>
<td>Mean of all pellets</td>
<td>302.5</td>
<td>302.5</td>
<td>302.5</td>
<td>297.5</td>
<td>5.0</td>
</tr>
</tbody>
</table>

*pellet 1 at the superior end of the holder; pellet 4 at the inferior end of the holder.

Gafchromic EBT-XD film results:

<table>
<thead>
<tr>
<th>Film Plane measured</th>
<th>Local gamma passing rate</th>
<th>Global gamma passing rate</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>5%-2mm</td>
<td>5%-1mm</td>
</tr>
</tbody>
</table>
Figure E.5: Example of an Audit report - Page 5 of 8.
RESULTS for alanine pellets – National results

You are centre number: 23

Solid line = mean, Dotted lines = 1 standard deviation, Dashed lines = 2 standard deviations

Figure E.6: Example of an Audit report - Page 6 of 8.
RESULTS for GaChronic EBT-XD film – National results

You are centre number: 23

Figure E.7: Example of an Audit report - Page 7 of 8.
Statement:

Our observations and analysis performed have not indicated any major concerns regarding the local practices for the specific aspects of dosimetry for cranial stereotactic radiosurgery.

Disclaimer: The results reported above are only valid on the day of the measurement. The responsibility for accuracy of clinical treatment delivery remains with the local centre. Medical physics expertise should be used in interpreting these results.
Appendix F

Published abstracts

F.1 Journal publications


Objective: To investigate and benchmark the current clinical and dosimetric practices in stereotactic radiosurgery (SRS) in the UK.

Methods: A detailed questionnaire was sent to 70 radiotherapy centres in the UK. 97% (68/70) of centres replied between June and December 2014.

Results: 21 centres stated that they are practising SRS and a further 12 centres plan to start SRS by the end of 2016. The most commonly treated indications are brain metastases and acoustic neuromas. A large range of prescription isodoses that range from 45% to 100% between different radiotherapy centres was seen. Ionisation chambers and solid water phantoms are used by the majority of centres for patient-specific quality assurance and thermoplastic masks for patient immobilisation are more commonly used than fixed stereotactic frames. The majority of centres perform orthogonal kV X-rays for localisation before and during delivery. The acceptable set-up accuracy reported ranges from 0.1 - 2 mm with a mean of 0.8 mm.

Conclusion: SRS has been increasing in use in the UK and will continue to increase in the next two years. There is no current consensus between SRS centres as a whole, or even between SRS centres with the same equipment, on the practices followed. This indicates the need for benchmarking and standardisation in SRS practices within the UK.

Advances in knowledge: This article outlines the current practices in SRS and provides a benchmark for reference and comparison with future research in this technique.

Abstract: There is renewed interest in film dosimetry for the verification of dose delivery of complex treatments, particularly small fields, compared to treatment planning system calculations. A new radiochromic film, Gafchromic EBT-XD, is available for high-dose treatment verification and we present the first published evaluation of its use. We evaluate the new film for MV photon dosimetry, including calibration curves, performance with single- and triple-channel dosimetry, and comparison to existing EBT3 film. In the verification of a typical 25 Gy stereotactic radiotherapy (SRS) treatment, compared to TPS planned dose distribution, excellent agreement was seen with EBT-XD using triple-channel dosimetry, in isodose overlay, maximum 1.0 mm difference over 200–2400 cGy, and gamma evaluation, mean passing rate 97% at 3% locally-normalised, 1.5 mm criteria. In comparison to EBT3, EBT-XD gave improved evaluation results for the SRS-plan, had improved calibration curve gradients at high doses, and had reduced lateral scanner effect. The dimensions of the two films are identical. The optical density of EBT-XD is lower than EBT3 for the same dose. The effective atomic number for both may be considered water-equivalent in MV radiotherapy. We have validated the use of EBT-XD for high-dose, small-field radiotherapy, for routine QC and a forthcoming multi-centre SRS dosimetry intercomparison.


Abstract: Anthropomorphic phantoms are considered ideal for performing quality assurance procedures in radiotherapy as they are more representative of patient-like conditions. In this work we present the adaptation and validation of a commercial anthropomorphic head phantom to be used for cranial radiosurgery audit. Two bespoke inserts were designed and produced for the phantom: one for providing the target and organ at risk for delineation and one for performing dose measurements. Each insert showed reproducible positioning within 0.5 mm and the positional agreement between the two inserts was within 0.6 mm. An initial basic treatment plan dose verification with a PTW Semiflex ionisation chamber showed agreement to the TPS within 0.5%. The phantom and inserts were then used to perform dose verification measurements of a 7-field IMRT linac radiosurgery plan, delivered by a Varian TrueBeam with 10FFF beam energy. The dose was measured with alanine pellets, EBT-XD Gafchromic film and the Exradin W1 plastic scintillation detector (PSD). Repeated film measurements showed consistent results for all gamma passing criteria investigated with very good agreement to the treatment planning system (TPS). For 2%-2mm global gamma the mean passing rate was 96.7% and the variation in passing rates did not exceed 2.1%. The alanine pellets and PSD showed good agreement with TPS predicted doses (-0.1 and 0.3% dose difference in the target). Good agreement was also observed between the
two dose detection systems (within 1%). The results demonstrated that the presence of the alanine and PSD does not affect the film measurement significantly, enabling simultaneous use of all three detectors and allowing a more efficient audit method. The developed methods presented in this study provide a thorough end-to-end test for stereotactic radiosurgery, with capability to incorporate all steps of the clinical pathway in a time-efficient and reproducible manner, making it suitable for a national audit.


Abstract: Scintillation detectors are considered ideal for dosimetric measurement of small fields in radiotherapy due to their near-tissue equivalence and their small sizes. A commercially available detector, the Exradin W1 (Standard Imaging, Middleton, USA), was previously characterised by two independent studies (Beierholm et al., 2014; Carrasco et al., 2015) and the results from these publications differed in some aspects (e.g. energy dependence, long term stability). The authors highlighted the need for more studies to be published (Beierholm et al., 2015; Carrasco et al., 2015). In this work, the Exradin W1 was characterised in terms of response to dose, dependence on dose rate, energy, temperature, angle of irradiation and long-term stability. The dose linearity, short-term repeatability, temperature dependence and energy dependence observed were in good agreement with published data, and corrections should be applied where possible in order to achieve low-uncertainty measurements. The angular dependence was characterised along the symmetry and polar axis of the detector for the first time and little dependence was observed, up to 1%. The long-term stability of the detector was observed to decrease at a rate of approximately 1.6% kGy-1 of exposure for the first 5 kGy recorded, but improved to a more manageable rate of 0.2% kGy-1 in the subsequent 20 kGy. The main goal of this dosimetric characterisation, was to assess the suitability of the Exradin W1 for its application to dose verification measurements in stereotactic radiosurgery. The results confirm that the detector is suitable for use in such situations. With the application of appropriate correction factors, the detector is now utilized in a multi-centre stereotactic radiosurgery dosimetric audit.

Abstract: Small field dosimetry is a developing area of research with a substantial number of studies published so far. Corrections to direct output factors (OF) measurements could be considered machine-specific for each beam quality and detector combination. Several groups are currently working on dosimetric protocols for small fields. It is, therefore however, possible for individual institutions to establish their own procedures, based on their experience and the equipment available. As part of Pinnacle commissioning, a practical approach for OF determination for small fields (from 3 x 3 down to 0.5 x 0.5 cm²) using the Daisy chain was investigated. Field size and detector specific correction factors were also experimentally determined. The Exradin W1 plastic scintillator was selected as class detector. The two methodologies were compared. NPL-Elekta linac is a Versa-HD with an Agility MLC, enabling delivery of flattened (FF) and unflattened (FFF) beams. Measurements were performed for 6 MV FF, 10 MV FF, 6 MV FFF, and 10 MV FFF. Field sizes from 0.5 x 0.5 to 40 x 40 cm² were organized in a Stored Beam Sequence. Three solid-state detectors (IBA SFD diode, PTW E diode 60012 and Exradin W1 scintillator) together with four ionization chambers (PTW 30013 Farmer, 31010 Semiflex, IBA CC13 and Exradin A26 micro chamber) and EBT-XD films were employed in this study. IBA Blue phantom with OmniPro-Accept was used to position the detectors precisely at the centre of the radiation field. Films were analysed using OmniPro-IMRT and Verisoft. The detectors were used with a Dose 1 electrometer except for the W1 detector, where a SuperMax electrometer was employed. CAX analysis of 6 MV FFF profiles for small fields (for four different detectors) showed a consistent averaged crossline deviation of 1.15 mm. Two different approaches of profile analysis for FFF beams in OmniPro-Accept ((i) field width at 50% and (ii) at maximum slope), were compared with film based measured Effective Field Sizes (FSeff). Average difference between FSeff was larger using maximum slope definition by -0.81 - 0.07 mm (across all fields, both diodes and FFF beams). Daisy chained OFs with the two diodes were on average 1.3% larger in comparison to the OFs calculated using the specific correction factors. The use of correction factors reduces difference in OFs obtained for the two diodes (from -0.61% to -0.1%). An overall uncertainty of 0.9% was determined for diodes’ specific correction factors, based on W1 as class detector. OFs for small fields were characterised with two available diodes and a micro ionisation chamber. Two experimental approaches were compared. The study provided a comprehensive set of OFs, which are used as baseline for linac quality assurance (QA) and Pinnacle commissioning. Additionally, the measured OFs are of high importance as a guidance for the beam data acquisition service and for comparison with measurements carried out for other linacs in similar conditions. Further investigation is needed to evaluate the influence of small fields OFs in Pinnacle models.
F.2 Conference presentations


**Purpose:** To examine the UK’s current practices in CSRS

**Methods:** A questionnaire, designed to include Gamma Knife (GK), Cyberknife (CK), Linac-Based (LB) and TomoTherapy (TT), was sent to 70 radiotherapy and radiosurgery centres in the UK between June and November 2014.

**Results:** 87.1% (61/70) of centres responded. Of these, 32.8% (20/61) were performing CSRS, 8.2% (5/61) are in the process of implementing CSRS and are planning to be clinical by August 2015 and 8.2% (5/61) are planning to implement CSRS by October 2016. The remaining 50.8% (31/61) are not performing CSRS and do not plan to implement it before October 2016. 25% (5/20) treat up to 4 patients per month, 45% (9/20) treat 5-15 patients per month and 30% (6/20) treat more than 16 patients per month. There are 29 machines used for CSRS in the country (14 LB, 6 CK, 7 GK and 1 TT) but they are not all dedicated to CSRS. The most commonly used techniques are non-coplanar static fields (used by 85% of centres), non-coplanar dynamic conformal arcs (20%) and circular collimator arcs (20%). 70% are using 6MV photons and 30% using Cobalt-60 (≈1.25MV). A range of imaging modalities is used for outlining: Fused CT&MR (70%), MR (60%), CT (50%), Angiogram (45%), PET (20%) and Fused CT&PET (10%). A large range of answers were given for the most common prescription isodose. Two peaks were seen: 20% (4/20) usually prescribe to the 45-50% isodose, 20% (4/20) to the 80-85% isodose, with the remaining centres prescribing between these 2 groups and up to the 95-100% isodose. Patient specific QA measurements are performed on every plan by 35% (7/20) and 65% (13/20) decreased the measurements taken after 10-25 plans. The results show a wide range of detectors and phantoms being used for QA measurements. The most common treatment sites are solitary and multiple brain metastases, followed by acoustic neuromas, meningiomas and AVMs. The majority of centres (70%) stated that treatment delivery usually takes less than 1 hour. The results show that pre-treatment and during-treatment imaging is used in the majority of CK and LB treatments but not used at all in GK. When asked for a figure of acceptable setup accuracy, 50% stated sub-millimetre accuracies with the remaining ranging from 1-2mm.

**Conclusions:** The number of centres delivering CSRS is increasing and will continue to increase in the next 2-3 years. This is particularly the case with LB radiosurgery. Most centres are aiming to expand their service to treat more indications and more patients. There is a wide variety of planning procedures, QA methods, prescription protocols and delivery practices despite the fact that the indications treated by all centres are comparable.

**Purpose/objectives:** Plastic scintillation detectors (PSD) are highly valuable for a variety of dosimetry applications, since their atomic composition and volume size produce small perturbation effects. A commercial PSD provided by Standard Imaging Inc (Exradin W1) is available and its Cherenkov light correction is based on the method proposed by Guillot et al. However, recent studies showed that the Cherenkov light ratio (CLR) is energy dependent, which could compromise its performance in clinical photon beams. The goal of this work is to investigate a theoretical model to characterise the energy dependence of the CLR and evaluate its effect on photon beam measurements.

**Materials/Methods:** The electron energy cut-off at which Cherenkov light is produced varies with the wavelength-dependent refractive index. Based on this rationale, the theoretical CLR, describing the relative amount of blue to green light, is formalised analytically using the Cherenkov emission distribution and the detection efficiency functions of the blue and green channels. As the analytic expression depends on the electron spectrum, Monte Carlo simulations of several photon beam qualities are performed to evaluate the spectrum. This allows predicting the theoretical CLR as a function of the TPR2010 quality index (QI), which includes cobalt-60 and mega-voltage (MV) beams. Experiments are performed to evaluate CLR over a wide range of QI in cobalt-60 and clinical MV beams.

**Results:** Comparison between experiments and theory show that the model reproduces the behaviour of the CLR energy dependence. However, the model under predicts the magnitude of the effect. For clinical MV beams, the variation of the theoretical CLR is about 0.5% while it is found to be about 1.8% with experiments. For cobalt-60 beam, the theoretical CLR is found to be about 1.005 of the value at the reference QI while the experiment reports a value of 1.017. Discrepancies between experiments suggest that other effects play a role in the energy dependence. More specifically, the model implicitly assumes isotropic Cherenkov emission, while the angular distribution of the light varies with the electron kinetic energy and the optical fibre only guides light emitted at a specific angular range. Further improvements modelling Cherenkov light transport explicitly should confirm these hypotheses.

**Conclusions:** The theoretical model proposed in this work is promising to evaluate the energy dependence of the Cherenkov correction in commercial PSD. Potential applications of this work could allow determining the energy dependence of PSD measurements using the CLR technique in small photon fields.

Purpose: The validation of radiotherapy treatments by dosimetric measurement is essential for the introduction of new techniques, pre-treatment verification and dosimetry audit. Film dosimetry has the advantage of high spatial resolution, low energy dependence and water equivalence. A new film (EBT-XD) has been assessed for its suitability for the dosimetry of stereotactic radiosurgery (SRS) applications.

Methods: Calibration curves for red, green and blue channels were created in the range of 0-4000 cGy for EBT-XD and its predecessor EBT3. Ten film pieces were irradiated in a nominal 6MV linac. The film was scanned using an EPSON Expression 11000XL scanner and the analysis was performed in FilmQA Pro software (Ashland ISP Inc, NJ, USA). Film dosimetry uncertainties were assessed for typical SRS fields, including lateral scanner effect at high doses. Both EBT-XD and EBT3 films were used in-phantom for treatment dose verification of typical Linac-based and Gamma Knife (GK) stereotactic radiosurgery within the STE2EV anthropomorphic phantom (CIRS, VA, USA). The dosimetry methodology for a forthcoming UK dosimetry audit of SRS treatment was utilised.

Results: EBT-XD film has lower optical density than EBT-3 throughout the dose range tested. EBT-XD was more suitable for high-dose applications because of a lower lateral scanner uncertainty. For the width of the film sizes that will be used in the SRS audit (50 mm) and the typical doses measured, the lateral scanner effect was estimated to be of the range of 0.5% for EBT-XD and 3% for EBT-3. Higher agreement between TPS and film dose distributions was seen for EBT-XD using both single and triple channel dosimetry at 2% (local normalisation), 1 mm gamma index analysis criteria, with the recommended triple channel used for EBT-XD having a 95.5% passing rate, compared to conventional single channel EBT3 having only 89.1%. Single channel EBT-XD had 89.7% passing rates and triple channel EBT-3 38.9%. An example is shown in figure 1, of EBT-XD showing a 98.3% gamma passing rate for a GK radiosurgery plan at 3% (local), 1.5 mm criteria.

Conclusions: We have evaluated the use of a new film, EBT-XD, for SRS dosimetry verification and demonstrated its suitability for a forthcoming audit of radiosurgery services in the UK. EBT-XD is less susceptible to lateral scanner effects and shows better agreement to TPS dose distributions than EBT-3 in linac-based radiosurgery dose verifications. EBT-XD also showed excellent agreement with TPS dose distributions in GK radiosurgery.

**Purpose:** To assess the geometric and dosimetric accuracy of stereotactic radiosurgery (SRS) in the UK for linac-based (LB), TomoTherapy (TT), Cyberknife (CK) and Gamma Knife (GK) radiosurgery.

**Methods:** 26 SRS centres were visited and 28 treatment plans were assessed (16 LB, 7 GK, 4 CK, 1 TT). The audit methodology employed an anthropomorphic head phantom with realistic tissue densities with one irregularly-shaped target (PTV), modelled on a brain metastasis, located centrally in the brain and in close proximity to the brainstem (OAR). The phantom was immobilised, scanned, planned and treated following the local protocol. Previously characterised near-water equivalent dosimeters were placed inside the phantom (EBT-XD film and alanine pellets) to measure absolute dose, both inside the PTV and OAR (Figure 1), and compare with TPS predictions. Film measurements were digitised with triple-channel-correction and compared to TPS dose planes on FilmQA Pro using gamma-analysis for a range of global and local criteria.

**Results:** Figure 2 shows the alanine measurements inside the PTV. LB showed the largest range in percentage difference to the TPS of 5.2% (-1.3% to +3.9%) with a mean of +0.5%. CK had a range of 2.6% (+1.4% to +4%), with the highest mean difference in comparison to the other platforms (+2.5%). GK showed the smallest range at 2.4% (-0.8% to +1.5%) being comparable to that of CK, with the smallest mean percentage difference (+0.4%) comparable to that of LB. Similar trends were observed in the OAR with alanine measurements showing a range from -1% to +3.6% (mean= +1.3%), 0% to +1.9% (mean= +0.9%) and -1.1% to +0.9% (mean= +0.1%), for LB, CK and GK respectively. The film measurements showed comparable passing rates between axial and sagittal films, regardless of the platform used. As expected, higher passing rates were observed for Global-gamma criteria. For 3%-2 mm Local-gamma, all except 2 films showed passing rates above 75%. For 5%-1 mm Global-gamma, all except 2 films showed passing rates above 90%.

**Conclusion:** This audit enabled the comparison of all participating centres in terms of the accuracy achieved during the delivery. The techniques used differed in many aspects. The LB group showed the largest variations in agreement to the TPS, related to more heterogeneous practices within the group, compared to small variations seen in CK, and more consistent practices seen in GK. Good overall agreement with the TPS was observed with only 3 centres falling above two standard deviations of the mean (2 centres in the target measurements and 1 in the OAR). Film measurements showed comparable gamma-passing rates for all centres assessed with small differences between platform groups. The results suggest that good agreement with the predicted dose distributions is achievable by all treatment modalities but highlight the need for standardisation in SRS practices.