

1 **Cone beam CT doses in radiotherapy patient positioning in Finland – prostate treatments**

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3 **Short title: Cone beam CT doses in patient positioning**

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5 T. Siiskonen^{1,2,*}, S. Alenius¹, T. Seppälä³, J. Tikkanen², M. Nadhum^{4,5}, J. Ojala^{4,5,6}

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7 1 University of Helsinki, Department of Physics, P.O. Box 64 (Gustaf Hällströmin katu 2) FI-00014

8 University of Helsinki, Finland

9 2 STUK - Radiation and Nuclear Safety Authority, Jokiniemenkuja 1, FI-01370 Vantaa, Finland

10 3 Comprehensive Cancer Center, Helsinki University Hospital and University of Helsinki, PL180, 00290

11 Helsinki, Finland

12 4 Tampere University Hospital, Department of Medical Physics, FI-33521 Tampere, Finland

13 5 Tampere University, Faculty of Medicine and Health Technology, FI-33720 Tampere, Finland

14 6 Tampere University Hospital, Department of Oncology, FI-33521 Tampere, Finland

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16 * Corresponding author, email teemu.siiskonen@stuk.fi, tel. +358 9 75988318

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26

27 *Abstract*

28

29 Imaging parameters, frequencies and resulting patient organ doses in treatments of prostate cancer
30 were assessed in Finnish radiotherapy centres. Based on a questionnaire to the clinics, Monte Carlo
31 method was used to estimate organ doses in ICRP standard phantom for prostate, bladder, rectum and
32 femoral head. The results show that doses from CBCT imaging have reduced compared to earlier
33 studies and are between 3.6 and 34.5 mGy per image for the above-mentioned organs and for normal
34 sized patients. There still is room for further optimization of the patient exposure, as many centres use
35 the default imaging parameters, and the length of the imaged region may not be optimal for the purpose.

36

37 *Key words:* Image guided radiotherapy; Dosimetry; Imaging; Monte Carlo; CBCT; Optimization

38

39 INTRODUCTION

40

41 Imaging using computed tomography (CT) has a central role in planning and delivery of radiotherapy.
42 Precise delivery of radiation to the intended treatment region requires precise daily knowledge of location
43 of the tumour and tissues surrounding it. Today, cone beam CT (CBCT) is one of the most used imaging
44 techniques for that purpose⁽¹⁾.

45

46 While improving the treatment safety and efficacy, the CBCT imaging does not come without potential
47 drawbacks. Frequent imaging during the treatment process can yield cumulative doses to the imaged
48 region that may exceed 1 Gy^(2, 3). In addition, for producing additional dose to the planning target volume
49 and to the tumour itself that is of the order of 1 – 2 % of the therapeutic dose, it produces additional
50 doses to the surrounding healthy tissues, increasing the risk of late harmful effects⁽⁴⁾. Therefore, the
51 optimization of the imaging parameters and practices is needed. The need for optimization has been
52 recognized by International Commission on Radiological Protection (ICRP), who has set up a specific
53 task group TG116 to increase awareness of radiological protection and optimization in radiotherapy
54 imaging (www.icrp.org). They conclude that more data on dose levels are needed for optimization
55 studies⁽⁵⁾. In Europe, European Radiation Dosimetry Group EURADOS and its working groups 6, 9 and
56 12 started a project to promote optimization and patient protection in radiotherapy imaging. In this
57 project, imaging practices in European radiotherapy centres are assessed and, based on the imaging
58 parameters collected, the patient (organ) doses are estimated with Monte Carlo simulations. The organ
59 doses, dose variations and their connections to imaging parameters are needed for producing good
60 practice guides to promote patient dose optimization, one of the main goals of the EURADOS project.

61

62 In this work, a summary of current CBCT imaging practices and resulting patient doses in Finland is
63 presented. All Finnish radiotherapy centres were sent a questionnaire on their imaging parameters and
64 practices in prostate treatments. Based on the answers, organ absorbed doses and dose volume
65 histograms were calculated for critical organs using the modified ICRP standard male phantom⁽⁶⁾ and
66 ImpactMC Monte Carlo code (CT Imaging©. GMBH, Erlangen, Germany)⁽⁷⁾. Status of the radiotherapy
67 CBCT imaging in Finland with regard of dose optimisation is discussed. This work forms a part of the
68 abovementioned EURADOS project.

69

70 MATERIALS AND METHODS

71

72 Information on imaging parameters and practices in prostate treatments were assessed by email
73 questionnaire that was sent to all Finnish radiotherapy centres. Answers from 9 out of 13 radiotherapy
74 centres were obtained (70 % of the Finnish centres). Out of the nine centres four were at university
75 hospitals and five were part of central hospitals. In questionnaire the imaging parameters needed for the

76 Monte Carlo calculations (x-ray tube voltage and tube current-time product Q, filtration, dose conversion
77 coefficient, i.e. the ratio of CTDI_w (or CBDI_w) to Q (mGy/mAs) and geometry factors) and the imaging
78 frequencies and practises (status of optimization) were asked for.

79

80 The Monte Carlo model needs the information on the x-ray beam shaped filter (so-called bowtie filter) of
81 the imaging device. The properties of these filters were determined experimentally. The method for
82 shaped filter thickness profile determination consisted of dose profile measurements with semiconductor
83 detector (Radcal Accu-Gold, California, USA) and radiochromic films (GafChromic XR-QA2 or XR-SP2,
84 Ashland, Wilmington, USA) with the x-ray tube positioned at zero degree position (upmost position) and
85 iterative calculation of the equivalent aluminium thickness of the filter from the relative dose compared to
86 the beam centre. The iterative calculations used a simulated spectrum, calculated with SpekPy
87 software⁽⁸⁾. The detectors were placed on a polystyrene sheet outside the edge of the patient support
88 table to minimize the unwanted backscattering. The detector and films were calibrated with IEC RQT-8
89 radiation beam quality⁽⁹⁾ in a secondary standard dosimetry laboratory at STUK. The shaped filter
90 profiles and inherent filtration were determined for OBI imaging devices in three Varian TrueBeam linear
91 accelerators (Varian Medical Systems (VMS) Inc. – A Siemens Healthineers Company, Palo Alto, CA,
92 USA), and the profiles agreed within the measurement uncertainties. Therefore, the same filter profile
93 was used for all Varian devices in simulations. Shaped filter of XVI imaging device in Elekta Infinity linear
94 accelerator (Elekta AB, Stockholm, Sweden) used in calculations was determined using a similar
95 approach as for Varian devices.

96

97 The x-ray spectra in Monte Carlo calculations were based on the parameters (tube voltage) obtained
98 from the radiotherapy centres, the inherent filter thicknesses obtained as described above and the anode
99 angle was 14°. In SpekPy, 0.5 keV energy bins were used for the x-ray spectra.

100

101 The organ doses were calculated with the ImpactMC Monte Carlo code⁽⁷⁾ using the standard ICRP male
102 voxel phantom⁽⁶⁾ with the arms removed to resemble the realistic exposure geometry in CBCT imaging
103 where hands are positioned out of the x-ray field. The effect of the shaped filter is taken into account in
104 ImpactMC simulations. In simulations, $2 \cdot 10^{10}$ photons were emitted from the x-ray source and it was
105 assumed that the rotation isocentre was in the middle of the prostate. The resulting three-dimensional
106 dose distributions were analysed with ImageJ program⁽¹⁰⁾. The organs (and organ doses) were
107 segmented manually from the ICRP male phantom⁽⁶⁾, and the organ doses were analysed slice by slice
108 using the organ position data of ICRP⁽⁶⁾ and visual inspection of the phantom. The organ doses were
109 determined for the prostate, bladder, rectum and femoral head containing the red bone marrow. As the
110 ICRP standard phantom was used, the simulation model did not include the patient support table.

111

112 The CTDI_w values of Table 1 reported by the radiotherapy centers were obtained according to definition
113 with measurements at peripheral positions and in the middle of a 32 cm diameter CTDI phantom
114 consisting of PMMA. ImpactMC takes as an input air kerma at the rotation isocentre, determined free-in-
115 air. To obtain the free-in-air kerma values, the attenuating effect of the 32 cm phantom was determined
116 with ImpactMC and the CTDI_w values of Table 1 were corrected accordingly to free-in-air kerma values.
117 However, there is some controversy of how CTDI should be measured in case of wide x-ray beams of
118 CBCT^(11, 12). Some Finnish centres do not measure CTDI routinely after the device initial acceptance
119 tests have been performed. Those centres that measure the CTDI follow the methodology of Buckley et
120 al⁽¹¹⁾ with 10 cm wide x-ray beam and 10 cm pencil chamber.

121

122 RESULTS

123

124 The results for the imaging parameters are shown in Table 1. In all cases with half fan x-ray beam, one
125 whole rotation (360 degrees) of the x-ray tube was used, whereas for the full fan x-ray beam the rotation
126 was partial, 200 degrees. In case of hospital C, the x-ray tube rotates either above or below the patient.
127 In all other cases, the tube rotates at the side of the patient. In all half fan cases the gantry rotation
128 started from the top position (0 degrees) of the gantry, whereas the full fan partial rotations started at a
129 slightly tilted angle, 10 degrees from the top position, except for the hospital C. For hospital C the
130 starting angle was 70 degrees (tube below, positive gantry rotation direction) or 110 degrees (tube
131 above, negative gantry rotation direction). Hospitals A – D are university hospitals, E – I are central
132 hospitals.

133

134 The results from the questionnaire on the imaging practices are presented in Table 2. Over half of the
135 radiotherapy centres use the default imaging parameters. The most common optimization method is the
136 reduction of the x-ray tube current, resulting in lower Q. The length of the imaged region is decreased in
137 some centres.

138

139 The organ doses (mean absorbed doses) for prostate, rectum, bladder, and femoral head calculated with
140 ImpactMC and parameters from Table 1 are shown in Table 3. Hospitals whose imaging parameters
141 were close to each other are grouped together since it is assumed that the resulting organ doses do not
142 differ significantly between these cases.

143

144 If imaging is done at every fraction, the cumulative mean doses (ignoring the large patient protocol in
145 hospital H) vary from 184 mGy to 530 mGy for prostate, from 107 mGy to 218 mGy for rectum, from 72
146 mGy to 233 mGy for bladder, and from 250 mGy to 690 mGy for femoral head, if imaging is done in
147 treatment consisting of 20 fractions. In case of 39 fractions the resulting doses are nearly doubled, e.g.,
148 the maximum dose for the prostate is 1034 mGy and 1346 mGy for the femoral head.

149

150 **Uncertainties**

151

152 The statistical uncertainty of the simulated organ doses is approximately 1 %, when organs in the
153 primary x-ray beam are considered. The organs considered here, prostate, rectum, bladder, and femoral
154 head, are all totally or mostly in the primary beam. Some uncertainty results from the manual
155 determination of the regions-of-interest in the dose calculation. However, the impact on the resulting
156 dose is small (< 1 % effect on organ doses). For the femoral head this uncertainty might be larger due to
157 incomplete knowledge of red bone marrow distribution within the femoral head in the phantom and
158 therefore uncertain delimitation of the region-of-interest. Largest uncertainty stems from the conversion
159 of in-phantom air kerma (or CTDI) values to the free-in-air values. This uncertainty is estimated to be 5
160 %.

161

162 The simulation model did not include the patient support table. Therefore, the doses especially at the
163 backside of the phantom might be overestimated by several percent or more⁽¹³⁾, since the attenuation
164 (tube below), and backscattering (tube above), caused by the table is absent. This effect is present both
165 in half fan and full fan x-ray beam geometries, as the x-ray tube moves both above and below the
166 patient, expect for hospital C where the tube rotates either above or below the patient. In this study, the
167 uncertainty related to table attenuation is taken to be 4 %. The total relative standard uncertainty of
168 organ doses is then 6.6 %.

169

170 The ICRP standard phantom⁽⁶⁾ used in this work was developed to represent a standard male weighing
171 73 kg and with a height of 176 cm. At the level of prostate, the phantom thickness is approximately 190
172 mm, and the width is 360 mm. Therefore, the calculated organ doses cannot be directly interpreted as
173 organ doses of actual patients with different sizes and anatomy. For that purpose, calculations based on
174 individual patient's CT images are needed or size-specific dose estimates could be used to scale the
175 organ mean absorbed doses corresponding to the patient size (for methodology, see e.g.⁽¹⁴⁾). In Finland
176 an average male has a height of 177 cm and weight of 87 kg, when ages over 30 years are
177 considered⁽¹⁵⁾. Therefore, the standard male ICRP phantom may underestimate the size (and thickness
178 at the abdomen) of an average Finnish male patient. Then, the calculated doses of this study may
179 overestimate the organ doses of a real patient, especially for organs deeper in the body.

180

181 The uncertainties are summarized in Table 4.

182

183 **DISCUSSION**

184

185 Many hospitals have optimized their imaging parameters and some clinics have set up their own imaging
186 protocols. Traditionally, the doses from imaging in radiotherapy process have not been a topic of great
187 interest⁽¹⁶⁾, and thus, the current trend of dose optimization can be seen as a welcomed addition to the
188 patient radiation protection repertoire. The protocols are optimized by the imaging device vendors, using
189 titanium filter to absorb the low energy x-rays, using partial rotation of the x-ray tube, development of
190 image reconstruction algorithms and, naturally, optimization of the imaging parameters at the clinics.

191

192 Gorges and Zylka⁽¹⁷⁾ analysed different dose saving techniques in CBCT. They concluded that e.g., the
193 use of bowtie filter, reducing the tube current and adjusting the field-of-view can result in substantial
194 dose savings. These are the techniques that are used by the participant hospitals in the current study.

195

196 In general, the doses are in general smaller than those obtained in earlier studies (e.g. ^(2, 18)) with
197 imaging devices from the same vendors than in this study. Alaei and Spezi⁽²⁾ reported per fraction doses
198 that ranged from 10 to 50 mGy for internal organs and up to 70 mGy for the skin. Kawahara et al⁽¹⁸⁾
199 measured prostate doses from 17 mGy to 26 mGy and rectum doses from 17 mGy to 25 mGy,
200 depending on the imaging device vendor.

201

202 Comparison of tube current-time product, $CTDI_w$ and prostate dose values from Tables 1 and 3 to the
203 values presented in the literature (e.g., ^(3, 11)) shows that the values for Q are similar. For example,
204 Ordóñez-Sanz et al⁽³⁾ reported Q-values 832 mAs (for TrueBeam) and 1014 mAs (for Clinac iX) for a
205 Pelvis M protocol. For the same Clinac iX Pelvis M protocol, a $CTDI_w$ value of 13 mGy is given, which is
206 in line with values of Table 1. In case of a large patient (Pelvis XL protocol), the $CTDI_w$ in Ordóñez-
207 Sanz et al⁽³⁾, 40 mGy, is higher than Hospital H large patient $CTDI_w$ (29.43 mGy).

208

209 Dose to the rectum doubles when the x-ray tube rotates below the table compared to case where it
210 rotates above (hospital C). However, the difference might be reduced by the patient support table, which
211 was not accounted for in the simulation. Moreover, the doses to other organs considered here are higher
212 in tube above-the-patient configuration. The organ doses with protocol for large patients (hospital H) are
213 not directly comparable to other results, since the standard ICRP male phantom was used in all
214 simulations. Hospital H did not provide criteria for a large patient.

215

216 The length of the imaged region varies from hospital to hospital. In some hospitals, only a smaller area
217 around the prostate is imaged, while in other hospitals, the entire pelvis is imaged (Table 1). On the
218 other hand, the information on bladder size and accordingly the position and shape of prostate is
219 valuable information that may not be obtained with too tight imaging margins. Even though this variation
220 has an effect to the mean doses of organs considered in this study, it does have even more profound
221 effect to doses further away from the isocentre. Unnecessary length of the imaged region produces

222 unwanted doses to regions where image guidance information is not needed. The needed imaging
223 length depends e.g. on positioning technique. Possibilities include bony-structure based or prostate-
224 based positioning or use of radio opaque markers inside the prostate gland⁽¹⁹⁾. The lower mean dose to
225 the bladder compared to the prostate is partly due to x-ray beam collimation, leaving parts of bladder
226 outside the primary beam.

227

228 The quality of the CBCT image, and subsequently the dose level, is chosen according to the intended
229 use of the images. If the actual delivered dose to the prostate and surrounding healthy tissues is
230 calculated from CBCT images of the day, then the device manufacturer's original pelvis protocol of the
231 entire pelvic area with a good image quality is often justified. However, if the CBCT images are used only
232 for image guidance of the prostate with or without seminal vesicles, the original image protocols defined
233 by the vendor can be adjusted and Q can be reduced by up to 30 % (Table 1), in addition, the field of
234 view and at the same time imaging time can be reduced by using a 200° full fan imaging instead of the
235 360° half fan imaging.

236

237 CONCLUSIONS

238

239 As a conclusion, the patient dose levels in radiotherapy patient positioning imaging in Finland are at the
240 same level or lower compared to international studies some years ago. Many hospitals have optimized
241 their imaging parameters, e.g., by reducing the x-ray tube current or making their own protocols.
242 However, there is room for further optimization such as reducing the lengths of the imaged regions.

243

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248

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250

251 Declarations of interest: none

252

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317

Hospital	Model	Dose conversion coefficient (mGy/100 mAs)	Beam	Tube voltage (kV)	Q (mAs)	CTDI _w (mGy)	Length of the imaged region (mm)	Field of view (cm)
A	Varian TrueBeam	1.64	Full fan	125	400	6.6	185	26.2
A	Varian TrueBeam	1.48	Half fan	125	720	10.66	175	46.5
B	Varian TrueBeam	1.48	Half fan	125	1080	15.98	160	45
C	Elekta Infinity XVI	1.84	Full fan	120	292.8	5.4	276.7	27.7
D	Varian TrueBeam STx	1.48	Half fan	125	1080	15.98	176	46.5
E	Varian TrueBeam	1.48	Half fan	125	1080	15.98	170	46
F	Varian TrueBeam	1.48	Half fan	125	720	10.66	176	46.5
G	Varian iX	2.6	Half fan	125	680	17.7	160	45
G	Varian TrueBeam	1.64	Full fan	125	750	12.3	186	26.3
H*	Varian TrueBeam	2.18	Half fan	140	1350	29.43	185	50
H	Varian TrueBeam	1.48	Half fan	125	720	10.66	185	50
I	Varian TrueBeam	1.48	Half fan	125	1080	15.98	155	46

319

320 Table 1. Imaging device model and imaging parameters for the nine radiotherapy clinics. * Protocol for
321 large patients.

322

323

Hospital	Imaging frequency	Level of optimization
A	At every fraction (20 or 39). Half fan: 25 fractions if brachytherapy is used, too. Golden fiducial markers in use.	Parameters have been adjusted
B	At every fraction, with SBRT at 5 fractions; With golden fiducial markers imaging with CBCT if 7 mm positioning tolerance is exceeded from kV planar images.	Default parameters
C	At every fraction (5, 20 or 25); Gold markers in use.	Parameters have been adjusted; Own protocol in use; Slight increase in exposure to have better soft tissue separation in image. Low dose protocol available for bony structure-based positioning.
D	At every fraction	Default parameters
E	At every fraction; Gold markers in use	Default parameters
F	At every fraction (20) or more often, if non-optimal positioning or organs have deformed	Parameters have been adjusted (tube current reduced)
G	At every fraction	Default parameters
H	At every fraction	Parameters have been adjusted; Protocol for large patients available.
I	At every fraction. Always both kV planar imaging and CBCT.	Default parameters

325

326 Table 2. Imaging frequencies and the level of optimization.

327

328

Hospital	Note	Dose to prostate (mGy)	Dose to rectum (mGy)	Dose to bladder (mGy)	Dose to femoral head (mGy)
A	Full fan	11.2	5.5	3.6	12.5
A, F, H	Half fan	16.6	6.2	4.6	14.2
B, D, E, I	Half fan	24.0	7.2	5.7	16.6
C	Tube rotates below	9.2	10.8	5.4	25.8
C	Tube rotates above	10.2	5.3	11.7	34.5
H	Large patient	43.3	18.2	13.4	41.7
G	Full fan	21.5	10.9	7.0	24.0
G	Half fan	26.5	8.3	6.3	18.4

329

330 Table 3. Mean organ doses (mGy) per CBCT image for prostate, rectum, bladder, and femoral head.

331

Uncertainty component	Relative standard uncertainty (%)
Statistical uncertainty of organ doses	1,0
Determination of region-of-interest in dose calculation	1,0
Dose conversion coefficient	5,0
Table attenuation	4,0
Total uncertainty	6,6

332