

Simulation approach for the evaluation of tracking accuracy in radiotherapy: A preliminary study

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journal or publication title	Journal of Radiation Research
volume	54
number	1
page range	146-151
year	2013-01-01
URL	http://hdl.handle.net/2297/33448

doi: 10.1093/jrr/rrs055

1 i) Title:
2 Simulation approach for the evaluation of tracking accuracy in radiotherapy; Preliminary study
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14 iv) Running title:
15 Simulation approach to evaluate tracking accuracy
16

17 v) Numbers of Text pages, Figures and Tables:
18 Total pages 11 (including this title page)
19 Figures 4
20 Tables 0
21

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39

40 **ABSTRACT**

41 Real-time tumor tracking in external radiotherapy can be achieved by diagnostic (kV) X-ray
42 imaging with a dynamic flat-panel detector (FPD). It is important to keep the patient dose as
43 low as possible while maintaining tracking accuracy. Simulation approach would be helpful to
44 optimize the imaging conditions. This study was performed to develop a computer simulation
45 platform based on a noise property of imaging system for the evaluation of tracking accuracy at
46 any noise level. Flat-field images were obtained using a direct-type dynamic FPD, and noise
47 power spectrum (NPS) analysis was performed. The relationship between incident quantum
48 number and pixel value was addressed, and a conversion function was created. The pixel values
49 were converted into a map of quantum number using the conversion function, and the map was
50 then input into the random number generator to simulate image noise. Simulation images were
51 provided at different noise levels by changing the incident quantum numbers. Subsequently, an
52 implanted marker was tracked automatically and the maximum tracking errors were calculated
53 at different noise levels. The results indicated that the maximum tracking error increased with
54 decreasing incident quantum number in flat-field images with an implanted marker. In addition,
55 the range of errors increased with decreasing incident quantum number. The present method
56 could be used to determine the relationship between image noise and tracking accuracy. The
57 results indicated that the simulation approach would aid in determining exposure dose
58 conditions according to the necessary tracking accuracy.

59

60 **Key words:** Noise simulation; Image noise; Flat-panel detector (FPD); Target tracking;
61 Radiotherapy

62

63 **INTRODUCTION**

64 Dynamic flat-panel detectors (FPDs) are commonly used in clinical practice. In external
65 radiotherapy, real-time tumor tracking can be achieved by diagnostic (kV) X-ray imaging with a
66 dynamic FPD [1-3]. There is concern regarding the relationship between image quality and
67 accuracy of target tracking, because low image quality is associated with the risk of increased
68 tracking errors, although it can reduce the patient dose. There are a number of factors that affect
69 image quality, such as X-ray tube voltage, radiation dose, and patient body shape. In particular,
70 reducing the radiation dose leads directly to an increase in image noise, and this may be one of
71 the major factors reducing the accuracy of target tracking.

72 Recently, several methods for measuring the temporal modulation transfer function

73 (MTF) and the detective quantum efficiency (DQE) have been proposed, and the properties of
74 FPDs used in dynamic imaging have been reported [4-7]. There are several reports on the
75 performance of electric portal imaging devices [8-9]. In a previous study, it was revealed that a
76 patient dose could be reduced by approximately 28% by optimal settings for the low-dose
77 acquisition mode with respect to image quality and dose [10]. However, there have been no
78 studies regarding the effects of image noise on tracking accuracy. It is necessary to address the
79 relationship between image noise and accuracy of target tracking to keep the patient dose as low
80 as possible while maintaining tracking accuracy.

81 In general, X-ray images have image noise due to statistical fluctuations in the number
82 of incident quanta entering a detector (q) [11,12]; a higher quantum number results in less image
83 noise. The value of q is determined as the reciprocal of the Wiener spectrum (WS) of the system
84 (*i.e.*, $q=1/WS$), which is measured in flat-field images. The relationship between q and pixel
85 value in an image is also determined from the average pixel value in the region of interest (ROI),
86 where the WS was measured. Furthermore, q follows a Poisson distribution. Thus, images with
87 various noise levels can be simulated by changes in the value of q and inputting them into a
88 Poisson random number generator. The simulation images would allow us to evaluate the
89 accuracy of target tracking at any noise level and to determine the appropriate exposure dose.
90 Our purpose was to develop a computer simulation method to determine imaging conditions
91 during target tracking in radiotherapy. Here, we developed a computer simulation platform
92 based on a noise property of imaging system and investigate the feasibility of the simulation
93 approach.

94

95 **MATERIALS AND METHODS**

96 *Measurement of noise power spectrum (NPS)*

97 (i) *Image data set*

98 A set of images was generated for determination of the NPS of a direct-type (a-Se/TFT) FPD
99 system (SONIALVISION Safire2; Shimadzu, Kyoto, Japan). An RQA5 X-ray spectrum was
100 used (HVL=7.1 mm Al, realized with 21 mm Al additional filtration at 70 kV) [13]. The matrix
101 size was 2048×2048 pixels, the pixel size was 0.123×0.123 mm, and the field of view was
102 25.4×25.4 cm. Image preprocessing consisted of offset and gain correction as well as
103 compensation for defective or nonlinear pixels, as applied in normal clinical use of the detector.
104 Pixel scaling was linear with respect to exposure, with a bit depth of 16 bits.

105

106 (ii) *NPS determination methods*

107 For determination of the NPS, three independent flat-field images were obtained at each of two
108 exposure levels (6 images in total); the exposure levels (air kerma) were 7.54 μ Gy and 15.7 μ Gy,

109 respectively, for the two series. The air kerma values were measured free-in-air in the detector
110 plane with an ionization chamber (AE-132a 2902209; Oyogiken Inc., Tokyo, Japan). The
111 source-to-target distance (SID) was limited to 1.0 m in the system evaluated. The ionization
112 chamber was placed 500 mm behind the detector, which was located approximately halfway
113 between the X-ray tube and the detector surface. The air kerma at the detector surface was
114 calculated by the inverse square distance law.

115 Regions of interest (ROIs), located manually near the detector center, were 256×256
116 pixels in size, with a pixel sampling pitch of 0.123 mm, in the same subarea of the full detector
117 area, for the two series. Average pixel values were measured by use of Image-J ver. 1.42
118 (<http://rsb.info.nih.gov/ij/>) in each ROI. The NPS was calculated according to IEC6220-1-1 [14].
119 For removing long-range background trends, a two-dimensional 2nd order polynomial was fitted
120 to each image and subtracted. The area of each image was divided into half-overlapping ROIs
121 for each image and the results were averaged. The 2-D NPS was then calculated by application
122 of the fast Fourier transform to each ROI. One-dimensional cuts through the 2-D NPS were
123 obtained by averaging of the central ± 7 lines (excluding the axis) around the horizontal and
124 vertical axes [15].

125

126 *Creation of conversion function from pixel value to quantum number*

127 The q is determined as the reciprocal of the WS [mm²] of the system as follows [11,12]:

$$128 \quad q = \frac{1}{WS} \quad (1)$$

129 In the present study, to determine q using Eq. 1, the averaged WS through all spatial frequencies
130 in two directions were used as WS in Eq. 1 for each exposure level. The quantum number per
131 pixel q' was then derived as follows:

$$132 \quad q' = q \times ps \times ps, \quad (2)$$

133 where ps is the pixel size, which was 0.123 mm in this study. The average pixel value (in digital
134 units) vs. the number of incident quanta (in count units) was fitted with a linear function,
135 $y=a+bx$.

136

137 *Simulation of image noise*

138 A tracking implanted marker with an acrylic plate 20 cm thick was located in clinical settings
139 during target tracking in radiotherapy and was imaged at 70 kV, 250 mA, 36 ms, and SID=1.0 m.
140 An averaging image was then created from ten images obtained at the above dose as a substitute
141 for the image with vanishingly low image noise obtained with a large dose.

142 Pixel values were converted to quantum number according to the conversion function.
143 Subsequently, the resulting image was weighted from 1.0 to 0.1 in increments of 0.1. Image

144 noise was induced by statistical fluctuation of the quantum incident to the detector, which
145 followed a Poisson distribution. Thus, to simulate image noise, the weighted images were input
146 into the Poisson random number generator in each pixel [11,12]. The output was the final
147 resulting image with image noise.

148

149 *Data analysis*

150 (i) *Target tracking*

151 The targets in the simulation images were tracked by a template-matching technique [16]. The
152 sum of differences in pixel value (R) between the search area in the next frame, $S(x + dx, y +$
153 $dy)$, and the template in the current frame, $T(x, y)$, was expressed as follows:

$$154 \quad R = \sum_{y=0}^N \sum_{x=0}^M |S(x + dx, y + dy) - T(x, y)| \quad (3)$$

$$155 \quad (0 < x < M, 0 < y < N, -10 < dx < 10, -10 < dy < 10)$$

156

157 M and N are the size of the template, and dx and dy are the search range. The smallest R value
158 was obtained when there were more similarities in the search area and template. The amount of
159 shift (dx, dy) in the search area was determined by minimizing of R , and the coordinates after
160 movement were expressed as $(x + dx, y + dy)$. In this study, the initial template was given as a
161 region into which the target was inserted in the first frame. After the second frame, the matching
162 region of interest in the previous frame was used as the new template. The size of the template
163 was 50×50 , the search range was ± 10 pixels, and thus the search area was 70×70 pixels,
164 determined to cover the displacements of implanted targets.

165

166 (ii) *Evaluation method*

167 Tracking accuracy was evaluated in images at ten different simulated noise levels. Implanted
168 marker was shifted in known amounts by image processing, ± 3 and ± 6 pixels in the
169 superior-inferior and right-left directions, respectively. A total of nine patterns were assessed of
170 each noise level, as shown in Fig. 1. The coordinates tracked were compared to the known shift
171 amounts, and the differences were calculated as tracking errors. The maximum tracking errors
172 were calculated and compared between simulation images at different levels of image noise.

173

Fig. 1

174 **RESULTS**

175 *NPS properties*

176 Figure 2 shows the WS in the horizontal and vertical directions for two exposure levels. The
177 results indicated that the present system had a stable WS through all spatial frequencies,

178 reflecting the noise property of a direct-type of FPDs. A high exposure level resulted in a higher
179 WS than a low exposure level at all of the spatial frequencies in both horizontal and vertical
180 directions.

181 Fig. 2

182 *Conversion function from pixel value to quantum number*

183 The average WS through all special frequencies of two directions in 7.54 μGy and 15.7 μGy
184 were $9.57 \times 10^{-6} \text{ mm}^2$ and $5.83 \times 10^{-6} \text{ mm}^2$, respectively. Thus, the quantum numbers q at each
185 exposure level were $1.05 \times 10^{-5} \text{ mm}^{-2}$ and $1.71 \times 10^{-5} \text{ mm}^{-2}$, respectively, according to Eq. (1). The
186 quantum numbers per pixel q' were $1.58 \times 10^{-3} \text{ pixel}^{-2}$ and $2.59 \times 10^{-3} \text{ pixel}^{-2}$, respectively,
187 according to Eq. (2). The fitted parameters values were $a=0.0585$ digital units and $b=541.35$
188 digital units per count. The quantum number per pixel q' was not zero even when the pixel value
189 was zero due to system noise caused by the electrical circuit and system.

190

191 *Effects of image noise on target tracking*

192 Figure 3 shows the simulation images at ten different noise levels. The results indicated that a
193 lower quantum number resulted in more image noise. It was difficult to recognize the location
194 of the marker at quantum numbers $<40\%$. Figure 4 shows the results regarding automatic
195 tracking of implanted markers. There was no error in the average images without noise. The
196 maximum tracking error increased with decreasing incident quantum number, as shown in
197 Figure 4. In particular, the tracking error tended to increase at less than half of the original
198 quantum number. Error bars show the standard deviation of nine data sets ($n=9$). The range of
199 errors also became larger in simulation images created with smaller incident quantum numbers.

200 Fig. 3 Fig. 4

201 **DISCUSSION**

202 Image noise has a big effect on visualization of an object with low contrast like a target in
203 radiotherapy. The present method was able to provide the relationship between image noise
204 levels and tracking accuracy. The maximum tracking error increased with increases in the image
205 noise. The range of errors also increased with increasing image noise. In this study, the tracking
206 error gradually increased about after half of the original quantum number. It was actually
207 difficult to identify implanted markers in images generated by less than half of the original
208 incident quantum number. Such information would be very useful for physicists to determine
209 the exposure dose according to the necessary tracking accuracy. The present method could be
210 applied to a different FPD system, requiring only the determination of the conversion function
211 for that system. These results indicated the feasibility of the simulation approach for
212 determination of the exposure dose during a real-time target tracking in radiotherapy.

213 However, there are several limitations that need to be solved in the present method.

214 For example, there are the other factors that affect image quality, such as X-ray tube voltage,
215 image lag, image blurring, and patient body thickness. In addition, it is necessary to consider the
216 other noise factors except quantum noise, such as electrical noise and structural noise. It should
217 be highlighted in this context that the simulation approach allows us to evaluate the accuracy of
218 target tracking at any noise levels. However, all of the noise factors are not involved in the
219 simulation image at the present time. For clinical implementation, further studies are required to
220 expand the system considering the other factors and to evaluate it in a real moving target in
221 clinical cases.

222

223 **CONCLUSION**

224 The present study was performed for development of a computer simulation method for
225 determining imaging conditions during target tracking in radiotherapy. Image noise was
226 simulated based on the noise property of that system, and the simulation was able to provide
227 the relationship between image noise levels and accuracy of target tracking, which the
228 maximum tracking error increased with decreases in the incident quantum number. These
229 results indicated the feasibility of the simulation approach for determination of the exposure
230 dose during a real-time target tracking in radiotherapy.

231

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Funding

This work was supported in part by a research grant from Japanese society of medical physics (JSMP).

Figure legends

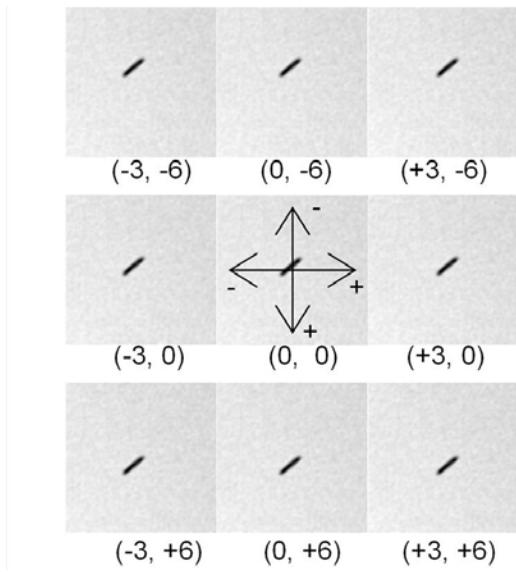


Fig. 1 Markers were shifted in nine combinations of ± 3 and ± 6 pixels in superior-inferior and right-left directions, respectively.

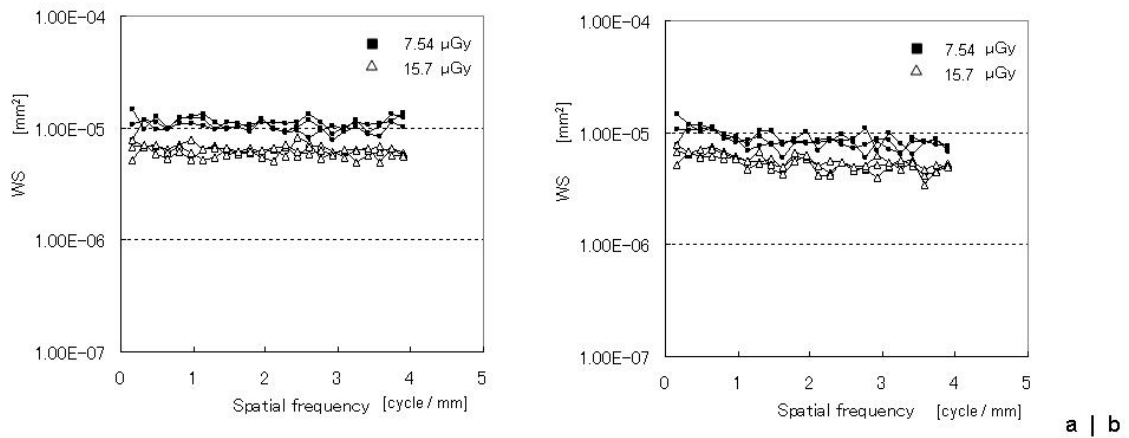


Fig. 2 Noise power spectra as determined for the set of flat-field images at two noise levels. (a) Horizontal direction. (b) Vertical direction.

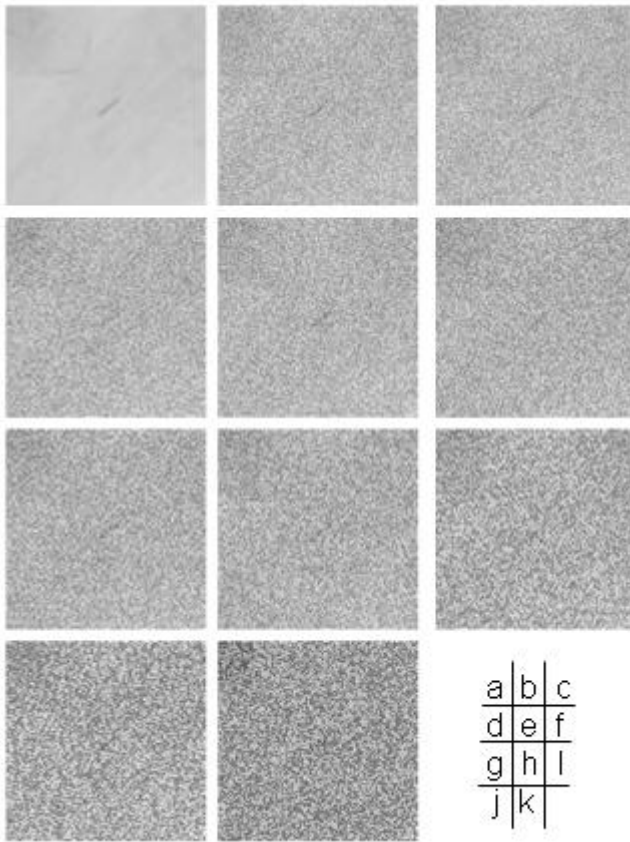


Fig. 3 Simulation images at ten different noise levels. (a) Averaging image (*i.e.*, image without noise). (b)–(k) Images with simulated noise by decreasing the number of incident quanta by 10%. Image noise increased as the incident quantum number decreased.

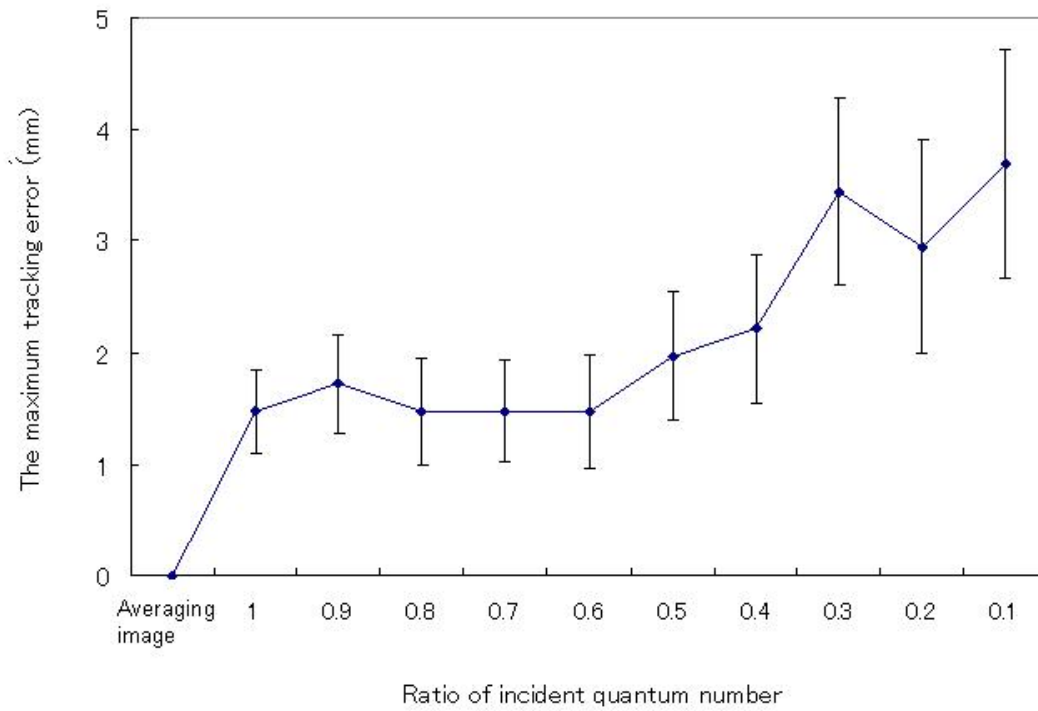


Fig. 4 Relationship between the maximum tracking error and ratio of incident quantum number to FPD (flat-field image). The average image without noise has no error, while, there are tracking errors in the images simulated in ratio of incident quantum number from 1.0 to 0.1. Error bars show \pm SD. (SD: standard deviation, n=9)