

1 Title: Practical method for determination of air kerma by use of an ionization chamber
2 toward construction of a secondary X-ray field to be used in clinical examination rooms
3

4 Authors:

5 Itsumi MAEHATA¹⁾, Hiroaki HAYASHI^{2,#)}, Natsumi KIMOTO¹⁾,

6 Kazuki TAKEGAMI³⁾, Hiroki OKINO³⁾, Yuki KANAZAWA²⁾, Masahide TOMINAGA²⁾
7

8 1) School of Health Sciences, Tokushima University

9 3-18-15 Kuramoto-Cho, Tokushima, Tokushima 770-8503, Japan

10 2) Institute of Biomedical Sciences, Tokushima University Graduate School

11 3-18-15 Kuramoto-Cho, Tokushima, Tokushima 770-8503, Japan

12 3) Graduate School of Health Sciences, Tokushima University

13 3-18-15 Kuramoto-Cho, Tokushima, Tokushima 770-8503, Japan
14

15 # Corresponding Author:

16 Hiroaki HAYASHI

17 Institute of Biomedical Sciences, Tokushima University Graduate School

18 3-18-5 Kuramoto-cho, Tokushima, Tokushima 770-8503, Japan

19 +81-88-633-9054

20 hayashi.hiroaki@tokushima-u.ac.jp

21

22 Keywords: ionization chamber, air kerma, contamination rate, scattered X-rays,

23 diagnostic X-ray equipment, secondary X-ray field

24

25

26 Abstract

27 We propose a new practical method for the construction of an accurate secondary
28 X-ray field by use of medical diagnostic X-ray equipment. For accurate measurement
29 of the air kerma of an X-ray field, it is important to reduce and evaluate the contamination
30 rate of scattered X-rays. In order to determine the rate quantitatively, we performed the
31 following studies. First, we developed a shield box in which an ionization chamber
32 could be set at an inner of the box to prevent detection of the X-rays scattered from the
33 air. In addition, we made collimator plates which were placed near the X-ray source for
34 estimation of the contamination rate by scattered X-rays from the movable diaphragm
35 which is a component of the X-ray equipment. Then, we measured the exposure dose
36 while changing the collimator plates, which had diameters of 25-90 mm ϕ . The ideal
37 value of the exposure dose was derived mathematically by extrapolation to 0 mm ϕ . Tube
38 voltages ranged from 40 kV to 130 kV. Under these irradiation conditions, we analyzed
39 the contamination rate by the scattered X-rays. We found that the contamination rates
40 were less than 1.7% and 2.3%, caused by air and the movable diaphragm, respectively.
41 The extrapolated value of the exposure dose has been determined to have an uncertainty
42 of 0.7%. The ionization chamber used in this study was calibrated with an accuracy of
43 5%. By using kind of this ionization chamber, we can construct a secondary X-ray field

44 with an uncertainty of 5%.

45 1. Introduction

46 Currently, X-ray examinations are widely used for diagnosis in the medical
47 field, and the risk of cancer in Japan caused by the diagnostic X-rays is estimated to have
48 the highest value in the world [1]. Radiologic technologists should make efforts to
49 reduce patient doses in addition to improving image quality [2]. In the diagnostic X-ray
50 region, reducing the entrance skin dose (ESD) [3] is important, in addition to optimizing
51 the exposure dose. Generally speaking, the ESD is estimated in terms of the air kerma
52 with a correction for the back-scatter factor (BSF). The original idea for this procedure
53 was reported previously [4,5], and recently Kato proposed a new method for calculating
54 the BSF [6]. Because the BSF is determined accurately, technologists need to measure
55 the air kerma with ionization chambers. Generally speaking, the ionization chambers
56 should be calibrated well with a standard X-ray field in which monoenergetic sources can
57 be provided within the special large room to reduce contamination by scattered X-rays.
58 Some institutions can provide calibration factors with accuracies of several percent, but
59 the calibrations are expensive and not convenient. If we can construct a secondary X-
60 ray field by using medical diagnostic X-ray equipment, inexpensive and convenient
61 calibrations will be available. As is generally known, the experimental environment by
62 means of medically-used X-ray equipment has many limitations; continuous X-rays with

63 contamination by scattered X-rays are generated. If these disadvantages caused by the
64 use of the diagnostic X-ray equipment are evaluated quantitatively, the secondary X-ray
65 field will become valuable under the limitation.

66 The diagnostic X-ray equipment used in clinics consists of an X-ray tube and a
67 movable diaphragm. It is well known that the movable diaphragm generates scattered
68 X-rays [7-9]. Therefore, the contributions of the scattered X-rays to the direct X-rays
69 should be estimated. Recently, Takegami *et al.* developed and suggested a new
70 collimator that has multiple-stage shields to reduce scattered X-rays coming from the
71 movable diaphragm [8], but the irradiation area formed by the equipment is limited to a
72 relatively small area [9]. For calibration of an ionization chamber without the
73 contamination of scattered X-rays, a relatively large irradiation area will be needed. We
74 propose here a new method for a practical calibration method used in the secondary-X-
75 ray field.

76 **Figure 1 (a)** illustrates the ideal situation in which we measure only direct X-
77 rays with an ionization chamber. In reality, scattered X-rays are additionally
78 superimposed on the direct X-rays, as shown in **Fig.1 (b)**; (A) and (B) indicate scattered
79 X-rays generated by air and by the movable diaphragm, respectively. **Figure 1 (c)**
80 shows a schematic drawing of the method we propose in this study. The ionization

81 chamber is located in a shield box, which was newly developed for the reduction of
82 scattered X-rays generated by air (indicated by (A) in **Fig. 1 (c)**). Also, a collimator
83 plate is placed in front of the movable diaphragm. In order to estimate the contamination
84 rate due to scattered X-rays (indicated by (B) in **Fig. 1 (c)**), we applied an extrapolation
85 method [10] in which experimental values associated with different collimator plates are
86 measured. In general, the exposure doses are analyzed based on the X-ray quality, which
87 is described by the half-value layers (HVLs) [11] of aluminum. Appropriate research
88 on the above-mentioned extrapolation method for deriving accurate half-value layers has
89 been performed [12,13]. We applied the extrapolation method to correct the exposure
90 dose measured with an ionization chamber.

91 In this paper, we propose a new method for constructing the secondary X-ray
92 field by using medical diagnostic X-ray equipment, and we developed a shield box for
93 the reduction of contamination from scattered X-rays. The rates of contamination by
94 scattered X-rays were determined, and we also evaluated the precision and accuracy of
95 the air kerma that was determined.

96

97 2. Materials and methods

98 2-1. Exposure dose measurements with ionization chambers

99 2-1-1. Development of apparatus

100 **Figure 2** shows a schematic drawing of the shield box which was newly
101 developed. We used commercially available materials to develop the apparatus. The
102 outer size of this shield box was 284 mm high, 334 mm wide, and 300 mm long. The
103 sides of the box were composed of 2 mm lead supported by 2 mm aluminum. We did
104 not add a shield at the back surface to prevent unnecessary scattered X-rays, which are
105 generated by the shield. The front surface was made of 2 mm aluminum and 2 mm lead,
106 and in addition to this, 2 mm of copper was used for reducing the characteristic X-rays
107 from lead [14]. The incident X-rays were limited by a shield-box-collimator placed at
108 the center of the front surface of the shield box. The size of this shield-box-collimator,
109 consisted of 2 mm aluminum and 2 mm lead, was 210 mm \times 165 mm, and had a
110 diameter of 100 mm ϕ . According to a well-known database [15], the mean range of
111 secondary electrons produced with X-rays having a tube voltage of 130 kV (effective
112 energy of 42 keV) was estimated to be 42 mm; therefore, the irradiation area formed by
113 the shield-box-collimator of 100 mm ϕ was sufficient for achieving secondary-electron
114 equilibration. The ionization chamber was held together by a clamp which was fixed to
115 the upper side of the shield box. At the rear of the shield box, a phosphor plate can be
116 set to confirm both the irradiation area and the position of the ionization chamber by use

117 of X-rays.

118 The collimator plates placed in front of the movable diaphragm (see **Fig. 1 (c)**)
119 were composed of lead and aluminum, each 210 mm high, 165 mm wide, and 2 mm thick.
120 A hole was bored through the center of the plate. The diameters of the holes were 25
121 mm ϕ , 30 mm ϕ , 40 mm ϕ , 50 mm ϕ , 60 mm ϕ , 70 mm ϕ , 80 mm ϕ , and 90 mm ϕ .

122

123 2-1-2. Experimental procedures

124 In order to measure exposure doses, we used diagnostic X-ray equipment
125 (MRAD-A 50S/70, Toshiba Medical Systems Corporation, Nasu, Japan), collimator
126 plates, a shield box, ionization chambers having a 3 cc detection volume (DC300, PTW,
127 Freiburg, Germany) and a 0.6 cc detection volume (30013 type, PTW, Freiburg,
128 Germany), a dosimeter (EMF521, EMF Japan Ltd., Osaka, Japan) for ionization
129 chambers. With help of the schematic drawing of **Fig. 1 (c)**, we explain the experiment.
130 **Figure 3** shows photographs of the experimental set up. Our experiments were
131 performed under the following four conditions: in setup A, the ionization chamber was
132 located in the shield box, and in setup B, the ionization chamber was placed in a free-air
133 condition (without shield box). For these conditions, ionization chambers having
134 different detection volumes were used; one had a detection volume of 0.6 cc and the other,

135 3 cc. By use of a commercially available standard X-ray field (Japan Quality Assurance
136 (JQA) organization, Japan), the calibration factors of the ionization chambers were
137 determined to be 13.91×10^5 (C/kg)/C for the 0.6 cc chamber and $2.83\text{-}2.99 \times 10^5$ (C/kg)/C
138 for the 3 cc chamber, with an uncertainty of 5%. The temperature and air pressure were
139 recorded, and the values measured with the ionization chambers were corrected so as to
140 agree with the standard temperature and pressure [16]. The collimator plates for
141 applying the extrapolation method were placed near the movable diaphragm (35 cm from
142 the X-ray source), as shown in the graph on the right in **Fig. 3**. An acrylic guide for the
143 collimator plates was set on a tripod for easy adjustment. The distances between the X-
144 ray source and the collimator-plate and ionization chamber were 35 cm and 250 cm,
145 respectively. Movable diaphragms was full open; the size of the irradiation area at the
146 end of an emission port is formed to be 13 cm \times 13 cm at the distance of 27 cm from
147 the X-ray source. Irradiation conditions were a current of 200 mA, an irradiation time
148 of 0.5 s, and tube voltages of 40 kV, 70 kV, 100 kV, and 130 kV. For each condition,
149 five measurements were performed for estimates of the statistical uncertainty [14].

150 Before measurements with the ionization chambers, we set a phosphor plate (RP-
151 4S, Konica Minolta Health Care Co., Ltd., Tokyo, Japan) at the rear of the shield box to
152 check the X-ray irradiation area and the position of the ionization chambers. In order to

153 check the exposure doses preliminary, the pixel value in the obtained image was analyzed
154 using a software ImageJ [17]. Then, based on the following mathematical formula
155 between digital value (DV) and dose (D), we estimated the doses from the pixel values
156 [18,19];

$$157 \quad D \propto \text{Exp}(0.00218 \times DV). \quad (1)$$

158 We used derived values to check the consistency of the measured values between the
159 ionization chambers and the phosphor plates.

160

161 2-1-3. Analysis

162 We describe the extrapolation method for estimating the contamination rate of
163 scattered X-rays measured with ionization chambers. According to that method [10],
164 the amount of scattered X-rays is considered to be proportional to the diameter of the
165 collimator plates which are set in front of the X-ray equipment. Here, the adopted value
166 corresponding to the ideal situation in **Fig. 1 (a)** can be obtained when we plot the
167 measured values as a function of the diameter of the collimator plates, 25 mm^φ to 90 mm^φ,
168 and the extrapolated values to 0 mm^φ. Note that the X axis is diameters of the collimator
169 plates, and not the diameters of the irradiation field. In our experiments, the detection
170 part of the ionization chamber was covered fully in the irradiation field even when the

171 collimator plate of 25 mm^ϕ was used. For the extrapolation, a linear function was used,
172 and the weighted least-squares method was applied. Simultaneously, we estimated the
173 uncertainty of the extrapolated value by consideration of the statistical uncertainty of the
174 measured values [C] of the ionization chambers. Then, the air kerma [J/kg] was
175 obtained by multiplying both the calibration factor [(C/kg)/C] and the “W-value divided
176 by the elemental charge e” of 33.97 [J/C] [15] to the measured value [C].

177

178 2-2. Exposure dose measurements using a CdTe detector

179 2-2-1. Experimental procedure

180 In order to check the effectiveness of the shield box based on a different
181 procedure, we also measured the X-ray spectra by using a CdTe detector (EMF123, EMF
182 Japan Ltd., Osaka, Japan) [20,21]. Setups C and D in **Fig. 3** show experimental setups
183 with use of the CdTe detector with and without the shield box, respectively; the CdTe
184 detector was set in the place by use of a camera platform. The irradiation conditions
185 were as follows: 70 kV, 200 mA, and 0.5 s. We applied the Compton scatter
186 spectroscopy method (scattering angle of 90 degrees) proposed by Maeda *et al.* [22]. A
187 carbon scatterer was used in place of the ionization chamber (see **Fig. 3**). In our
188 experimental conditions, the counting rate (counts per seconds: CPS) of the CdTe detector

189 was kept below 1 kCPS to reduce the pulse pileup effect [23,24].

190

191 2-2-2. Analysis

192 In order to analyze the exposure dose by use of the measured X-ray spectra of
193 the CdTe detector, we applied the following analysis. First, by use of the Klein-Nishina
194 formula and the response function of the CdTe detector, originally measured spectra were
195 unfolded [22]. Then, we transformed the X-ray spectra $\Phi(E)$ to air kerma by using the
196 following equation:

$$197 \quad \text{Air kerma} = \int \Phi(E) \times E \times \left(\frac{\mu_{\text{tr}}(E)}{\rho} \right) dE, \quad (2)$$

198 where E and $\mu_{\text{tr}}(E)/\rho$ are the energy [25] and the mass energy transfer coefficient,
199 respectively.

200

201 3. Results

202 3-1. Exposure dose measurements by use of ionization chambers

203 **Figure 4** shows X-ray images of the phosphor plate which we used to check the
204 irradiation areas of the 3 cc chamber in setup A. **Figures 4 (a)** and **(b)** indicate the
205 results based on the collimator plates of 25 mm ϕ (smallest) and 90 mm ϕ (largest),
206 respectively. It is clearly seen that the detection area of the ionization chamber is

207 included sufficiently in the irradiation area. From a geometrically based consideration,
208 irradiation areas of 178 mm^ϕ and 642 mm^ϕ can be formed by use of the collimator-plates
209 of 25 mm^ϕ and 90 mm^ϕ, respectively, in setup B (without a shield box) at the position
210 where the chamber was set. On the other hand, in setup A (with a shield box), both
211 irradiation areas were limited to be 114 mm^ϕ, as shown in **Fig. 4**. This was caused by
212 the shield-box-collimator of 100 mm^ϕ placed in front of the shield box. In the irradiation
213 parts in the figure, DVs measured with the phosphor plate not including the ionization
214 chamber are also shown; namely, DV of 3537.5±0.9 for the 25 mm^ϕ collimator plate, and
215 that of 3542.2±0.9 for the 90 mm^ϕ collimator plate. From equation (1), the relative doses
216 corresponding to the collimator plates of 25 mm^ϕ and 90 mm^ϕ were estimated to be
217 1.000±0.002 and 1.010±0.002, respectively. The difference in values was consistent
218 with the result, which is presented in the next paragraph (**Fig. 5 (b)**).

219 **Figures 5 (a)-(d)** shows a comparison of exposure doses measured with
220 ionization chambers between setup A (with a shield box, solid circles) and setup B
221 (without a shield box, open circles) in **Fig. 3** for four tube voltages. The results of 3 cc
222 chamber are presented. The X-axis shows the diameter of the collimator plate. A
223 linear function was applied for fitting to the experimental data, and an extrapolated data
224 corresponding to 0 mm^ϕ was obtained. Then the exposure doses were normalized by the

225 extrapolated value, and the normalized values are plotted on the Y-axis. It is clearly seen
226 that the data measured without the shield box are systematically larger than those with
227 the shield box. The differences in data with or without the shield box at 40 kV, 70 kV,
228 100 kV, and 130 kV were 0.9%, 1.3%, 1.1%, and 1.0%, respectively. The error bars in
229 the figure are standard deviations of the measured values for five measurements, and in
230 the extrapolated value, the contribution of these uncertainties is considered. As a result,
231 the statistical uncertainties of the extrapolated data for 40 kV, 70 kV, 100 kV, and 130 kV
232 were approximately 0.5%, 0.2%, 0.1%, and 0.3%, respectively.

233 **Figure 6 (a)** shows a comparison of the results for the two ionization chambers.
234 The solid and open circles indicate the results for the 3 cc and 0.6 cc chambers,
235 respectively. All of the air-kerma values measured with the 0.6 cc chamber are
236 consistent with those of the 3 cc chamber. This result indicates that our experiments did
237 not depend on the volume of the ionization chambers.

238

239 3-2. Exposure dose measurements with the CdTe detector

240 **Figure 6 (b)** shows X-ray spectra measured with a CdTe detector with or without
241 the shield box. The X axis shows the energy [keV], and the Y axis shows the counts.
242 Using equation (2), we derived corresponding dose with the spectra; the relative values

243 of derived air kerma of the conditions with (setup C in **Fig. 3**) and without the shield box
244 (setup D in **Fig. 3**) were 1.000 ± 0.002 and 1.018 ± 0.002 , respectively. As described
245 above, the results measured with the ionization chamber shown in **Fig. 5 (b)** indicate a
246 1.6% difference between measured values with and without the shield box with use of the
247 90 mm ϕ collimator plate; the result with the CdTe detector was consistent with that of the
248 ionization chamber.

249

250 4. Discussion

251 In the present study, we proposed an accurate measurement method for air kerma
252 by use of diagnostic X-ray equipment. In general, diagnostic X-ray equipment has a
253 movable diaphragm, and this becomes a generator of scattered X-rays. To construct an
254 accurate X-ray field, we proposed to use a shield box to reduce the scattered X-rays, and
255 we estimated the contamination rate by the scattered X-rays.

256 It was considered that a free-air condition is suitable for calibration of ionization
257 chambers. We consider that our method is applicable only to the diagnostic X-ray region,
258 and that it is useful for reducing scattered X-rays from the movable diaphragm of clinical
259 X-ray equipment. As described above, the experiments were validated because the
260 contamination rate by the scattered X-rays measured with one ionization chamber was

261 consistent with that measured with another ionization chamber, the phosphor plate, and
262 the CdTe detector. This finding strongly support the verification of our method. Next,
263 we describe the evaluation of the accuracy of our method.

264 As shown in **Fig. 5**, the extrapolation method works well because experimental
265 data deviated evenly from a linear fitted line. The effect of the shield box was clearly
266 presented by the data; the open circles (setup B, without a shield box) were systematically
267 larger than the closed circles (setup A, with a shield box). Here, we estimate the
268 differences between these data corresponding to the X (diameter of shield-box-
269 collimator) = 100 mm $^\phi$. As represented in **Fig. 5**, they were 1.5-1.7% for tube voltages
270 of 40-130 kV. The differences are considered to be due to contamination by scattered
271 X-rays from air, which is indicated by (A) in **Fig. 1 (b)**. Reducing these scattered X-
272 rays is important for deriving an accurate exposure dose, because the extrapolated data
273 (related to the 0 mm $^\phi$ of the collimator plate) became systematically 0.9-1.3% larger than
274 the ideal values when extrapolation was applied to setup B (without a shield box). From
275 these findings, we concluded that a more accurate value of exposure dose can be obtained
276 with use of our shield box.

277 Here, we also discuss the contamination rate of scattered X-rays from a movable
278 diaphragm, which is indicated by (B) in **Fig. 1 (b)**. In **Fig. 5**, the amount of these X-

279 rays was observed by the differences between the extrapolated value of the exposure dose
280 and other data in setup A (with a shield box). In the present case, scattered X-rays from
281 the movable diaphragm were estimated to be at most 1.8-2.3%. Although these
282 estimated values are not common, they become a good example to explain our method
283 when experiments are performed with diagnostic X-ray equipment installed in clinical
284 examination rooms.

285 **Figure 7** shows a relationship of the extrapolated values of the exposure dose in
286 terms of the measured values [C] and air kerma [J/kg] at 70 kV. In the dimension of the
287 measured value [C], statistical uncertainty is considered only to these data. In the
288 present case, the statistical uncertainty of the extrapolated value was 0.7%, as represented
289 by the right-hand graph in **Fig. 7**. On the other hand, as shown in the left graph in **Fig.**
290 **7**, the uncertainty of the air kerma (extrapolated value) was determined by consideration
291 of both the statistical uncertainty (0.7%) and the uncertainty of the calibration factor (5%).
292 Therefore, the final uncertainty of the measured value becomes approximately 5%.

293 When we want to calibrate another ionization chamber by using our secondary
294 X-ray field, the ionization chamber can be calibrated with 5% uncertainty. At this time,
295 the calibration factor has a larger uncertainty compared with the contribution of scattered
296 X-rays. However, if we can use an accurately calibrated ionization chamber, our

297 method of using a shield box may be more valuable. Our secondary X-ray field will
298 also play an important role in the calibration not only of ionization chambers, but also of
299 other radiation detectors such as solid detectors. We plan to calibrate an optically
300 stimulated luminescence (OSL) dosimeter by using our secondary X-ray field. The
301 detection efficiency of the OSL dosimeter is completely different from that of the
302 ionization chambers; for example, the relative efficiency of 20 keV X-rays is 20% larger
303 than that of 60 keV [26]. In other words, when an experimenter calibrates the OSL
304 dosimeter, the contribution of low energy X-rays (scattered X-rays) should be considered.
305 With the proposed calibration method, it is hoped that the contribution of the scattered X-
306 rays is properly estimated; firstly, the ionization chamber for standard is measured and
307 analyzed by the proposed method (as represented in **Fig. 5**), secondary, a radiation
308 detector which experimenter wants to calibrate is measured with the same condition and
309 also analyzed with the proposed method, and then, the extrapolated values are compared.
310 In this procedure, the effect of the low energy X-ray contamination on each detector was
311 properly corrected.

312

313 5. Conclusion

314 In conclusion, we proposed a practical calibration method for which we used an

315 original shield box and collimator plates to prevent scattered X-rays, and we evaluated
316 the contamination rates by them for construction of a secondary X-ray field by means of
317 general diagnostic X-ray equipment. Our equipment is portable; we considered that our
318 equipment was useful for calibration of ionization chambers with X-ray equipment used
319 in clinical examination rooms. We applied the method to a general experimental room
320 in Japan, and we found that the contamination rates of scattered X-rays from the air and
321 the movable diaphragm were less than 1.7% and 2.3%, respectively. The precision and
322 accuracy of the extrapolation method were approximately 0.7% in the measured value
323 [C], and 5% in the air kerma [J/kg]. We found that our method was more accurate than
324 the uncertainty of the calibration factor used. Our method will become valuable when
325 a more accurately calibrated ionization chamber is available.

326

327

328 Conflict of interest

329 We have no conflict of interest.

330 6. References

- 331 [1] Amy Berrington de Gonzalez, Sarah Darby. Risk of cancer from diagnostic X-ray:
332 estimates for the UK and 14 other countries. *The Lancet*. 2004;363:345-351.
- 333 [2] Uffmann M, Prokop CS. Digital radiography: The balance between image quality and
334 required radiation dose. *European Journal of Radiology*. 2009;72:202-208.
- 335 [3] Takegami K, Hayashi H, Okino H, Kimoto N, Maehata I, Kanazawa Y, Okazaki T,
336 Kobayashi I. Practical calibration curve of small-type optically stimulated
337 luminescence (OSL) dosimeter for evaluation of entrance skin dose in the diagnostic
338 X-ray region. *Radiological Physics and Technology*. 2015;8:286-294.
- 339 [4] Grosswendt B. Backscatter factors for x-rays generated at voltages between 10 and
340 100 keV. *Physics in Medical and Biology*. 1954;29(5):579-591.
- 341 [5] Klevenhagen SC. Experimentally determined backscatter factors for x-rays generated
342 at voltages between 16 and 140 kV. *Physics in Medical and Biology*.
343 1989;34(12):1871-1882.
- 344 [6] Kato H, Method of calculating the Backscatter Factor for Diagnostic X-rays Using the
345 Differential Backscatter Factor. *Japanese Journal of Radiological Technology*.
346 2001;57(12):1503-1510.
- 347 [7] Hayashi H, Taniuchi S, Kamiya N, et al. Development of a Pin-hole Camera Using a

348 Phosphor Plate, and Visualization of the Scattered X-ray Distribution and Optical
349 Image. Japanese Journal of Radiological Technology. 2012;68(3):307-311.

350 [8] Takegami K, Hayashi H, Konishi Y, et al. Development of multistage collimator for
351 narrow beam production using filter guides of diagnostic X-ray equipment and
352 improvement of apparatuses for practical training. Medical Imaging and Information
353 Sciences. 2013;30(4):101-107.

354 [9] Hayashi H, Takegami K, Okino H, et al. Procedure to measure angular dependences
355 of personal dosimeters using diagnostic X-ray equipment. Medical Imaging and
356 Information Sciences. 2015;32(1):8-14.

357 [10] Hayashi H, Takegami K, Konishi Y et al. Indirect Method of Measuring the Scatter
358 X-ray Fraction Using Collimators in the Diagnostic Domain. Japanese Journal of
359 Radiological Technology. 2014;70(3):213-222.

360 [11] Ariga E, Ito S, Deji S et al. Determination of half value layers of X-ray equipment
361 using computed radiography imaging plates. Physica Medica 2012;28:71-75.

362 [12] Trout ED, Kelley JP, and Lucas AC. Determination of Half-Value Layer. Radiology.
363 1959;73:107-108.

364 [13] Trout ED, Kelley JP, and Lucas AC. Determination of Half-Value Layer. The
365 American Journal of Roentgenology, Radium Therapy, and Nuclear Medicine.

366 1960;84(4):729-740.

367 [14] Knoll GF. Radiation Detection and Measurement third edition. John Wiley & Sons,
368 Inc. New York. 1999;321. ISBN 0-471-07338-5.

369 [15] Lucien P. Energy loss, range, and bremsstrahlung yield for 10-keV to 100-MeV
370 electrons in various elements and chemical compounds. Atomic Data and Nuclear
371 Data Tables. 1972;4:1-127.

372 [16] Khan FM. The Physics of Radiation Therapy fourth edition. Lippincott Williams &
373 Wilkins. 2010.

374 [17] Rasband WS, ImageJ, U. S. National Institutes of Health, Bethesda, Maryland, USA,
375 <http://imagej.nih.gov/ij/>, 1997-2015.

376 [18] Kimoto N, Hayashi H, Maehata I, et al. Development of All-in-one Multi-slit
377 Equipment for Measurements of the Input-output Characteristic of a Phosphor Plate,
378 Japanese Journal of Radiological Technology, 2013;69(10):1165-1171.

379 [19] Maehata I, Hayashi H, Takegami K, et al. Fabrication of Improved Multi-slit
380 Equipment to Obtain the Input-output Characteristics of Computed Radiography
381 Systems: Correction of the Heel Effect, and Application to High Tube-voltage
382 Experiments, Japanese Journal of Radiological Technology, 2014;70(9):867-876.

383 [20] Okino H, Hayashi H, Nakagawa K *et al.* Measurement of response function of CdTe

384 detector using diagnostic X-ray equipment and evaluation of Monte-Carlo simulation
385 code, Japanese Journal of Radiation Radiological Technology, 2014;70(12):381-1391.

386 [21] Fukuda I, Hayashi H, Takegami K *et al.* Development of an experimental apparatus
387 for energy calibration of a CdTe detector by means of diagnostic X-ray equipment,
388 Japanese Journal of Radiation Radiological Technology, 2013;69 (9):952-959.

389 [22] Maeda K, Matsumoto M and Taniguchi A. Compton-scattering measurement of
390 diagnostic x-ray spectrum using high-resolution Schottky CdTe detector. Med. Phys.
391 2005;32:1542-1547.

392 [23] Then SS, Geurink FDP, Bode P, et al. A pulse generator simulating Ge-detector
393 signals for dead-time and pile-up correction in gamma-ray spectrometry in INAA
394 without distortion of the detector spectrum. J. Radioanal. Nucl. Chem.
395 1997;215(2):249-252.

396 [24] Cano-Ott D, Tain JL, Gadea A. Pulse pileup correction of large NaI(Tl) total
397 absorption spectra using the true pulse shape. Nucl. Instrum. Methods. 1999;430:488-
398 497.

399 [25] Hubbell JH. Photon mass attenuation and energy-absorption coefficients, The
400 International Journal of Applied Radiation and Isotopes, 1982;33(11):1269-1290.

401 [26] Takegami K, Hayashi H, Okino H, et al. Energy dependence measurement of small-

402 type optically stimulated luminescence (OSL) dosimeter by means of characteristic
403 X-rays induced with general diagnostic X-ray equipment, Radiological Physics and
404 Technology, in press (DOI: 10.1007/s12194-015-0339-9).

405

406 Figure captions:

407 Fig.1 Comparison between ideal, real conditions in measurements with ionization
408 chamber, and our proposed method. **(a)** Ideal condition of chamber; it measures only
409 direct X-rays. **(b)** Actual condition; it also measures scattered X-rays. **(c)** Our
410 proposed method for measuring only direct X-rays with an ionization chamber. There
411 are a newly developed box and collimator plates to shield from scattered X-rays. These
412 collimator plates have different diameters. (A) and (B) show scattered X-rays caused
413 by the air and the movable diaphragm, respectively.

414

415 Fig.2 Schematic drawings of our newly developed shield box which is 284 mm high,
416 334 mm wide, and 300 mm thick. The front surface is made of 2 mm lead, 2 mm
417 aluminum, and, in addition, 2 mm copper to absorb the characteristic X-rays of lead.
418 The ionization chamber is held by a clamp which is fixed to the upper side of the shield
419 box. For checking the irradiation area and a position of the ionization chamber, a
420 phosphor plate can be inserted at the back.

421

422 Fig.3 Experimental conditions for the X-ray equipment and the detectors. We
423 performed the experiment by using four conditions (setups A-D); different combinations

424 of two kinds of detectors, and with or without shield box.

425

426 Fig.4 X-ray images of a phosphor plate which was placed at the rear of the shield box.

427 **(a)** and **(b)**: Results for 25 mm ϕ and 90 mm ϕ collimator plates. The detector was placed

428 at the center of the irradiation field. The digital value (DV) of the image measured with

429 the phosphor plate and the converted dose from the DV are given.

430

431 Fig.5 Experimental results measured with ionization chamber for 40 kV to 130 kV as

432 a function of diameter of collimator plate. The Y-axis shows dose, which was

433 normalized by the extrapolated value. The solid and open circles refer to the conditions

434 of setup A (with a shield box) and setup B (without a shield box), respectively.

435

436 Fig.6 Verification of our method. **(a)** Comparison of the results for 70 kV between

437 two different-size ionization chambers. The solid-circle data (3 cc chamber) and open

438 circle data (0.6 cc chamber) are consistent with each other. **(b)** X-ray spectrum

439 measured with the CdTe detector with and without the shield box. We plotted the

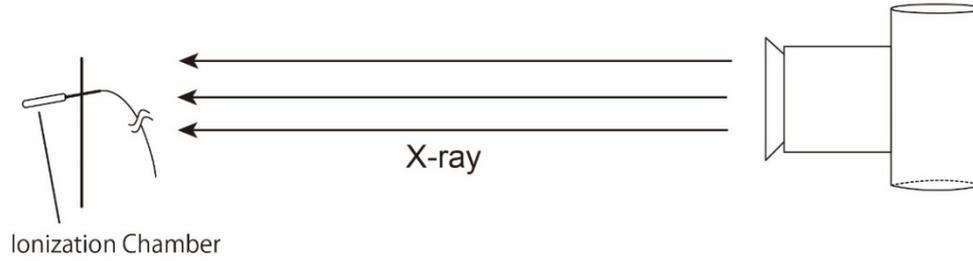
440 original and unfolded spectra, in which the lines with solid and closed circles represent

441 measured data with shield box and without it, respectively.

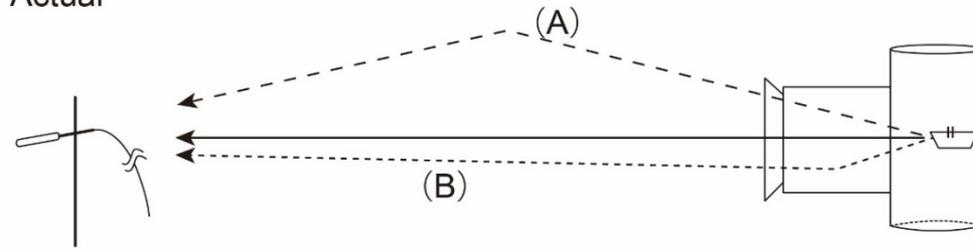
442

443 Fig.7 Uncertainty estimation of our method. The data in the right figure have
444 statistical uncertainty. In this case, the extrapolated data have an uncertainty of 0.7%.
445 The data in the left figure show the total uncertainty in which both the statistical
446 uncertainty (0.7%) and that of the calibration factor (5%) are considered.

(a) Ideal



(b) Actual



(c) Our method

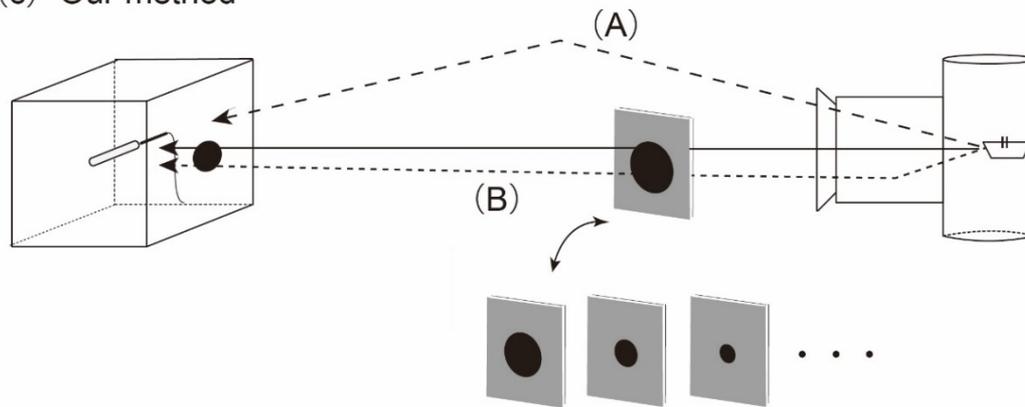


Fig.1

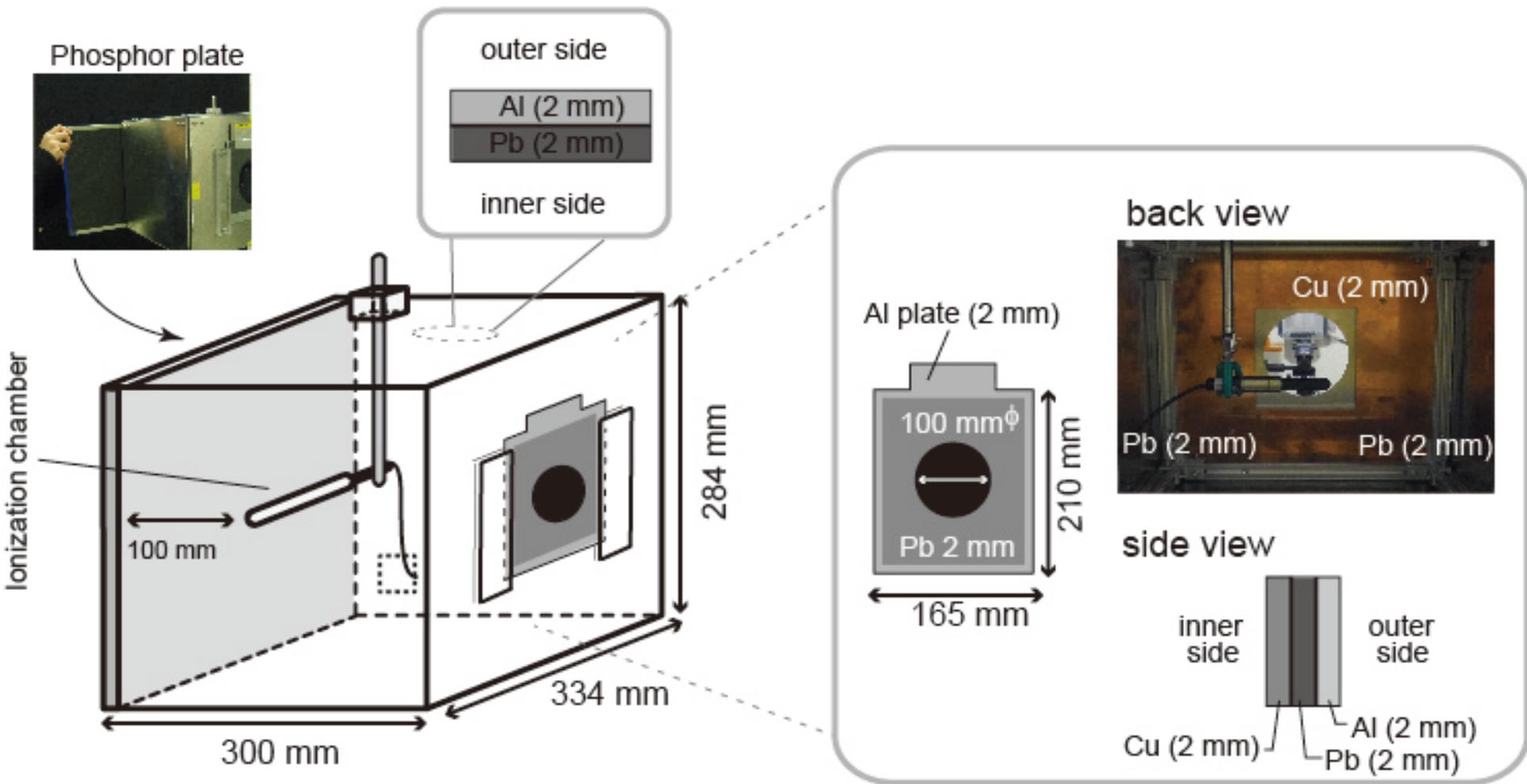


Fig.2

Detector

ionization chamber

setup A



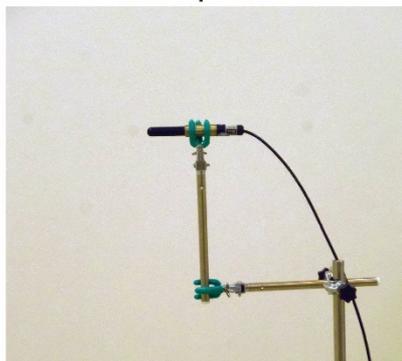
CdTe

setup C



with shield box

setup B



without shield box

setup D

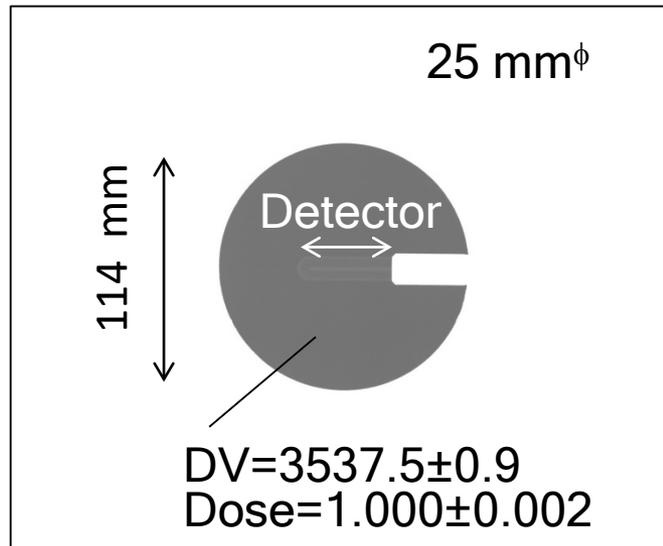


X-ray equipment



Fig.3

(a)



(b)

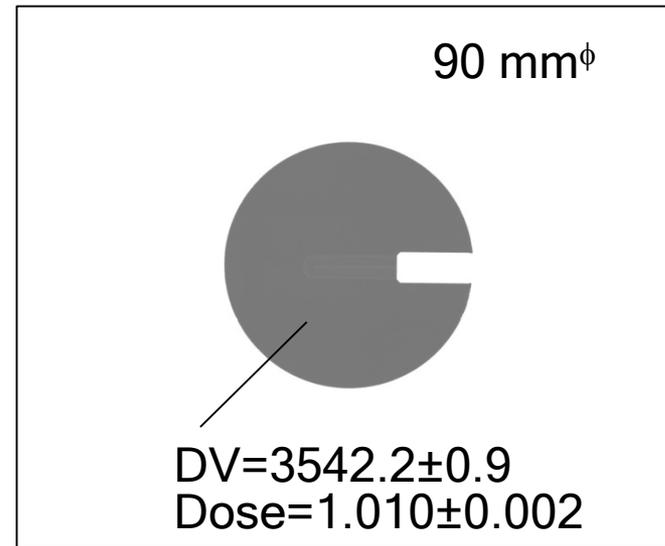
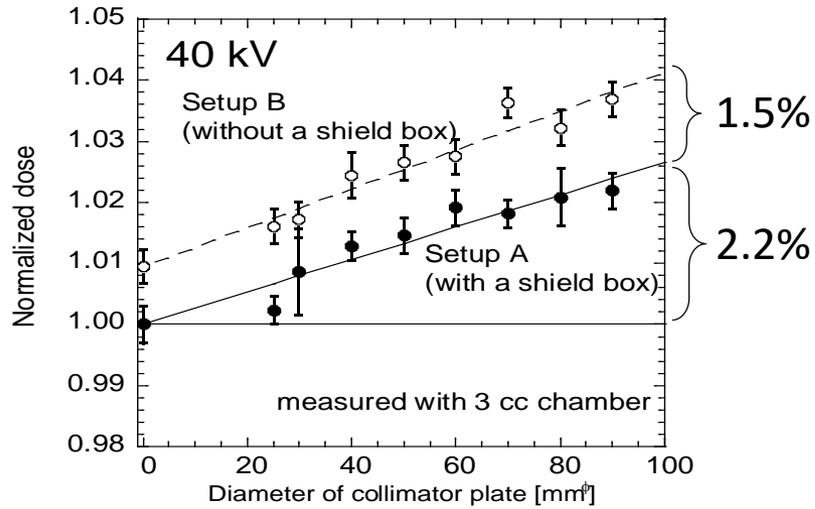
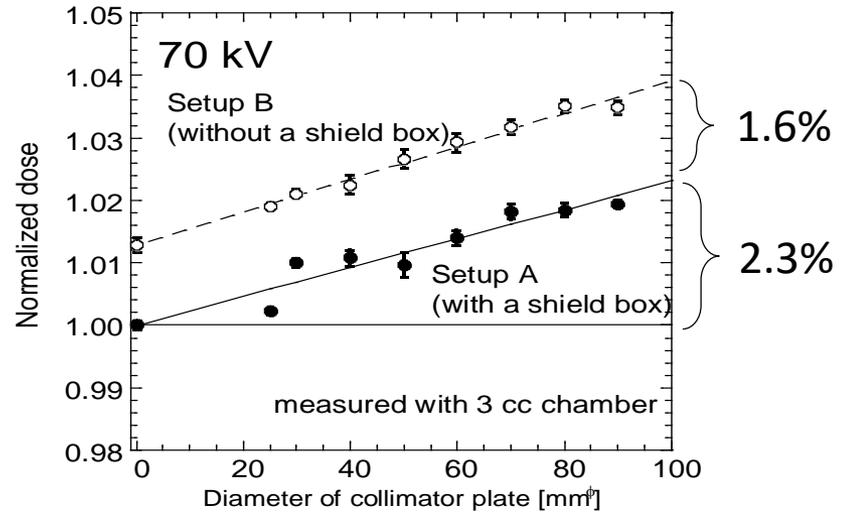


Fig. 4

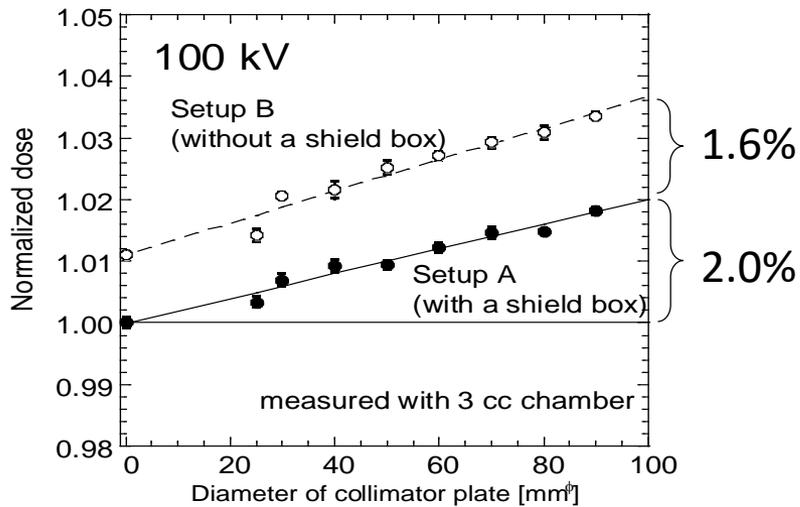
(a)



(b)



(c)



(d)

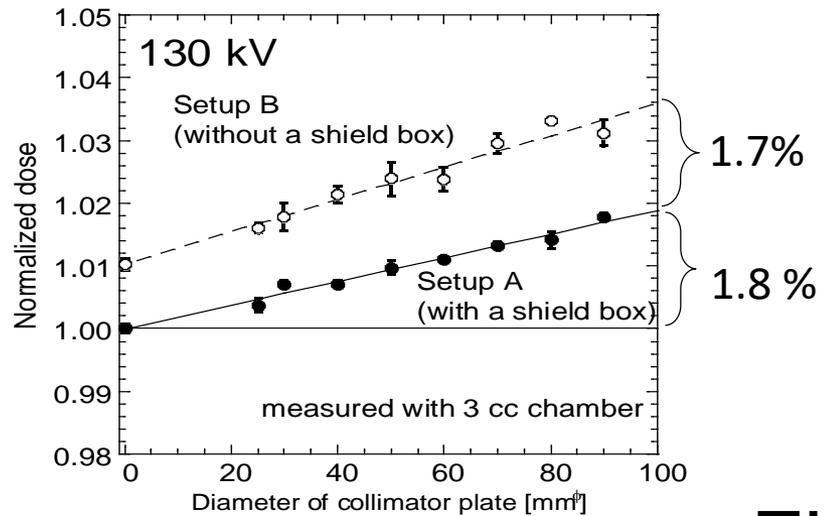
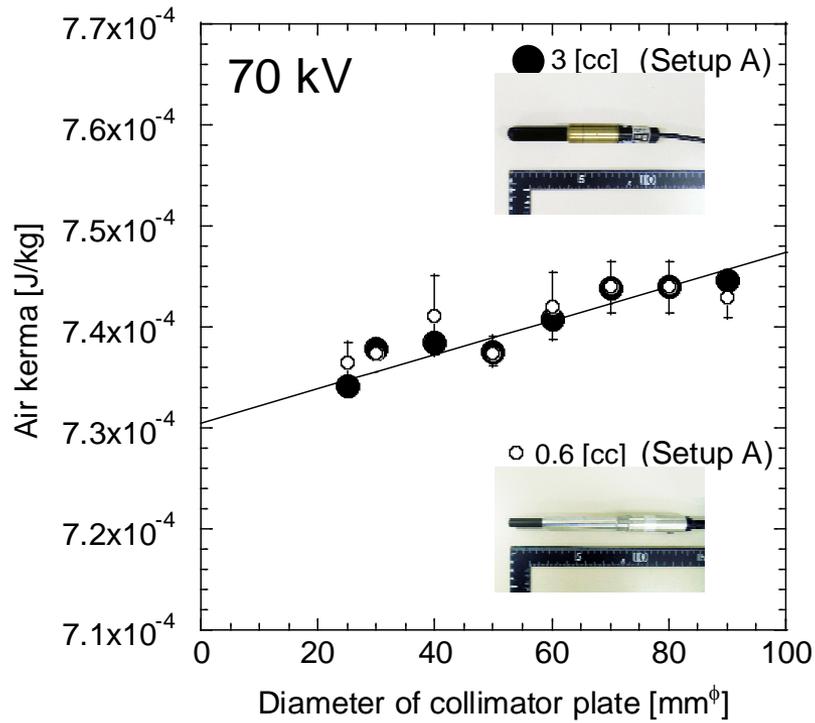


Fig.5

(a)



(b)

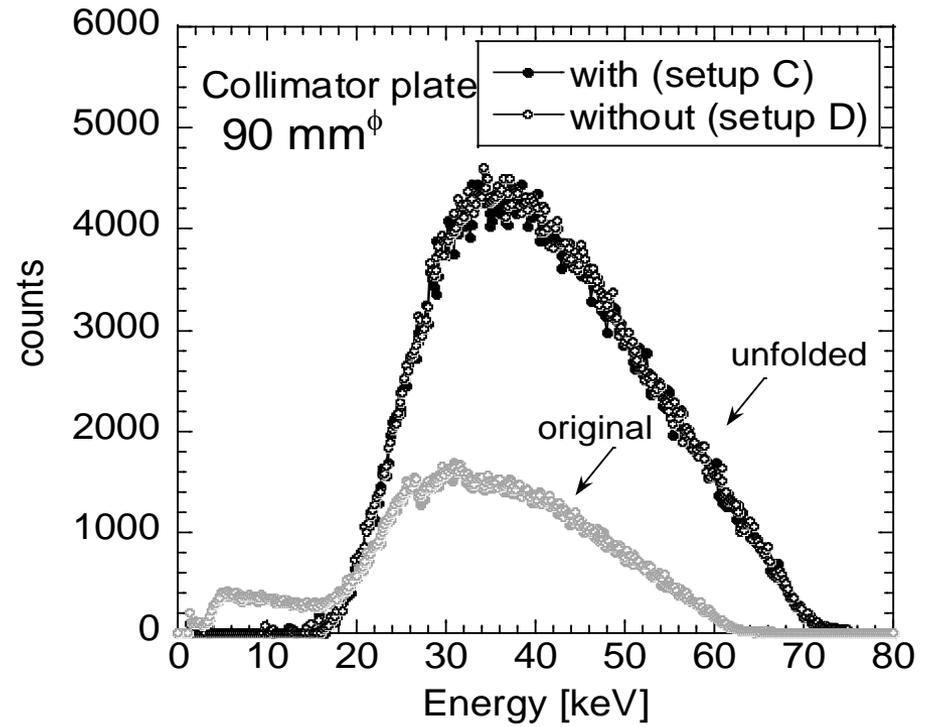
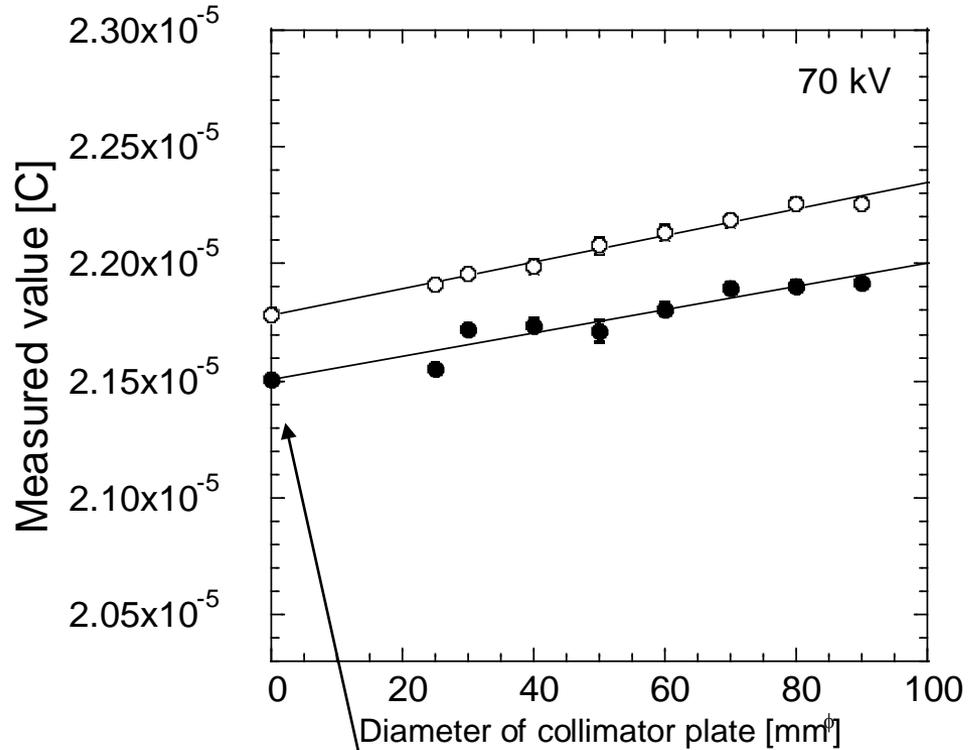
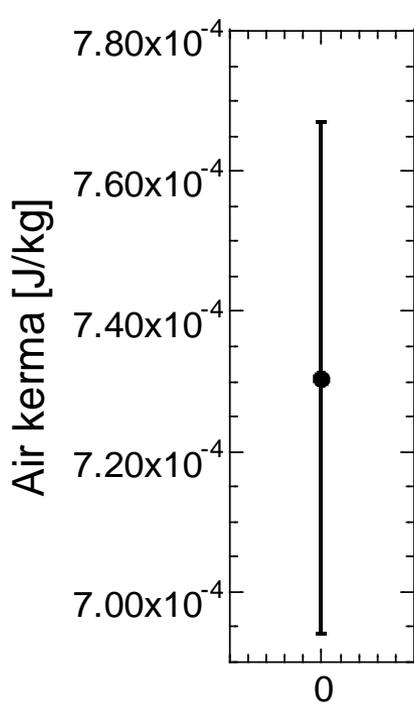


Fig. 6



Uncertainty of 5% ← × Calibration factor → Uncertainty of 0.7%

Fig.7