

The effect of source to image-receptor distance on effective dose for some common X-ray projections

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Abstract. A number of studies on the effects of source to image distance (SID) on dose to the patient have been published. These generally assume that the X-ray beam is collimated to a rectangular region of fixed size at the entrance surface of the patient at each SID. This is poor radiographic practice. In this work, effective doses have been determined using the commercially available Monte Carlo simulation package PCXMC. Three collimation cases were considered, namely to regions of clinical interest at the entrance and at the centre of the patient, and a fixed beam size at the image receptor. For cases of collimation to a central region of interest or to the image receptor, the saving in effective dose at increased SIDs is modest.

Introduction

The International Commission on Radiological Protection (ICRP) stated in 1970 that “short focus–skin distances result in a low transmission ratio due to a high divergence of the beam and consequently, the incident exposure of the patient for a given exposure at the film or other recording device is high” [1]. For extended fluoroscopic procedures, inadequate source to image distance (SID) is a contributor to excessive skin doses and leads to increased risk of deterministic effects [2]. However, the SID has been claimed by a number of authors to also have a significant influence on the stochastic radiation risk to the patient [3, 4]. This is also implied by at least one textbook [5]. However, Welander and Wickman [6] predicted that for the case of collimation to a constant beam size at the image receptor at different SIDs, the mean energy imparted to the patient is independent of the SID. This is logical, because it is now recognized that the effective dose is proportional to the dose–area product (DAP), which is independent of SID when the collimation is to the image receptor at each SID [7]. This finding has been supported by Poletti [8], who showed that for the case of collimation to the image receptor the effective dose is essentially independent of SID. In view of the continuing claims [9] about the value of long SIDs as a method of reducing the stochastic risk, the earlier work [8] has been extended to consider the effect of SID on effective dose when collimation is to a fixed field size at varying positions within the patient, such as at the entrance surface, mid plane or surface of the image receptor.

Theory

Following Welander and Wickman [6], an estimate of the maximum value of the mean energy imparted to a uniform phantom, $\bar{\epsilon}$, is given by:

$$\bar{\epsilon} = \psi_0 a \left(\frac{I_\psi}{I_a} \right)^2 (1 - e^{-\mu x_1}) \quad (1)$$

where ψ_0 is the energy fluence at the entrance surface of the phantom, a is the beam area at distance I_a from the source, I_ψ is the distance to the point at which the energy fluence is held constant (generally the image receptor), μ is the linear attenuation coefficient for the phantom material and x_1 is the object thickness. When $I_\psi = \text{SID}$ and $I_a = I_\psi$, then the maximum mean energy imparted is independent of distance. As the area to which the beam is collimated is shifted closer to the source ($I_\psi = \text{SID}$, $I_a < I_\psi$), the contribution of the squared term increases. Welander and Wickman considered three cases, these being collimation to a fixed field at the image receptor, to a region of interest (ROI) at the centre of the phantom and to the entrance surface, as illustrated in Figure 1. It is assumed that for a given tube potential setting and energy spectrum, the attenuation coefficient for the average photon energy may be used. The results for these three cases are illustrated in Figure 2, which shows the relative mean energy imparted to a 20 cm thick phantom, at a range of SIDs, for each case.

Equation (1) is an overestimate of energy imparted that assumes all secondary photons are absorbed by the phantom [6] and it applies only to a uniform slab. Both of these are significant approximations. Furthermore, energy imparted is only an approximate indicator of risk. By comparison effective dose makes allowance for the radiosensitivities of the organs and tissues of the body, allowing a good estimate of stochastic risk. For example, Poletti [8] showed that for some projections, the effective dose actually increases as the SID is increased, whereas energy imparted is independent of SID, for collimation to a fixed field size at the image receptor.

Method

The commercial Monte Carlo package PCXMC (STUK. Radiation and Nuclear Safety Authority, P.O. Box 14, FIN-00881 Helsinki, Finland) was used to

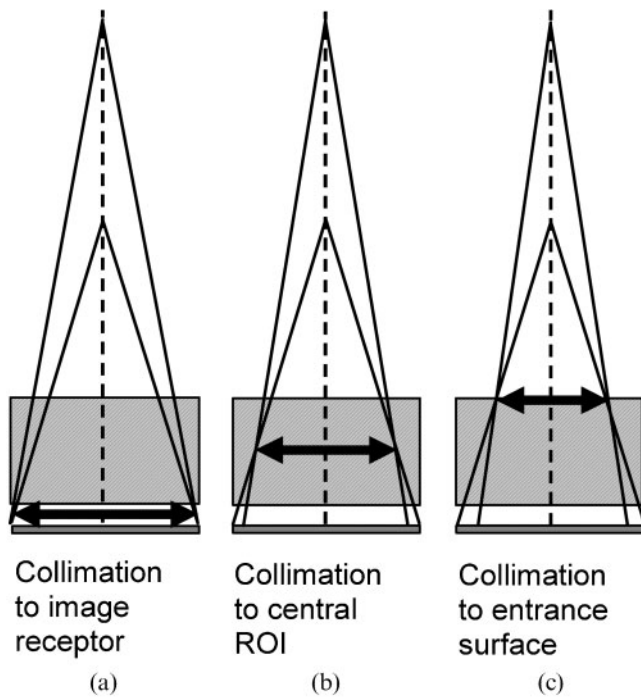


Figure 1. The three cases considered by Welander and Wickman, collimation to (a) image receptor, (b) central region of interest (ROI) and (c) entrance surface. The arrows indicate the size and position of the beam area.

determine the effective dose for each projection and SID. This package uses a phantom based on that of Cristy [10], as modified by the National Radiological Protection Board (NRPB) [7]. Some further changes have been made to the phantom as detailed by Tapiovaara et al [11].

In calculating the effective dose over a range of SIDs, the following conditions were applied.

- (1) The tube voltage (kVp) was kept constant regardless of SID. This was justified because the tube voltage is the contrast control, and should not be used to compensate for SID changes.

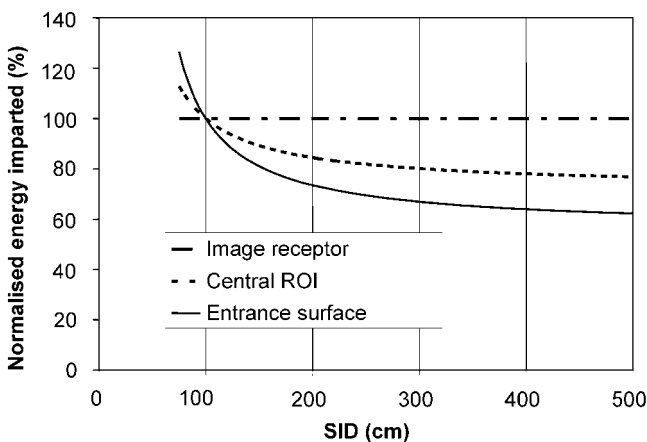


Figure 2. Theoretical normalized energy imparted as source-to-image distance (SID) is changed, for collimation to image receptor, central region of interest (ROI) and entrance surface.

- (2) At each SID, it was assumed that an antiscatter grid of identical performance was used, focused for that SID.
- (3) At each SID the mAs was chosen to result in the same dose at the position of the image receptor, using:

$$mAs_{\text{new}} = mAs_{\text{old}} \frac{SID_{\text{new}}^2}{SID_{\text{old}}^2} \quad (2)$$

- (4) The entrance surface air kerma (ESAK) at each SID was determined using:

$$ESAK_{\text{new}} = ESAK_{\text{old}} \times \frac{SSD_{\text{old}}^2}{SID_{\text{old}}^2} \times \frac{SID_{\text{new}}^2}{SSD_{\text{new}}^2} \quad (3)$$

- (5) It was assumed that the proportion of the energy absorbed by the image receptor due to scattered radiation is constant at each SID. This has recently been shown to be a reasonable assumption [12].

The standard patient size only was considered, in part because the patient scaling provided by PCXMC is not particularly realistic. Differences in patient body type and composition were not considered, as PCXMC does not allow this.

No attempt was made to cover the entire range of radiographic projections used. Instead, the projections for which reference effective doses are listed in the New Zealand code of safe practice [13] were used, adding abdomen anteroposterior (AP) as it is included in NRPB-R262 [7]. These are listed in Table 1. For these projections the image receptor sizes, centring points and typical tube voltage settings were taken from NRPB-R262 [7]. Effective doses were calculated, as noted above, using PCXMC. Three collimation cases were considered, as per Figure 1. First, the X-ray field size was kept constant at the image receptor, with the beam sizes given in NRPB R262. Second, the X-ray beam size was kept constant at an ROI at the central depth within the patient. The size of this ROI was calculated from the image receptor beam sizes in NRPB R262, based on a standard SID of 100 cm, using similar triangles. Third, the X-ray field size was kept constant at the entrance surface, again calculated from the R262 field sizes using similar triangles. The ROI sizes and locations are given in Table A1 in the Appendix. Space precludes the inclusion of all of the calculated light field dimensions. However, some example data are given in Table A2 in the Appendix.

In order to compare the influence of SID on effective doses for different projections, all results were normalized to an ESAK of 1.0 mGy at an SID of 100 cm. For chest projections, the range of SIDs was increased as chest radiographs are usually performed at greater distances, typically 180 cm. In order to investigate the effect of very long SIDs, the range of SIDs was considered out to an SID of 10 m, for one projection only, the lateral lumbar-sacral joint (LSJ), because for this projection the effective dose was found to be particularly sensitive to changes in SID. It is presumed that the effect of extended SIDs on effective dose will therefore be less for the other projections.

The projections used, with the tube voltage and field sizes from NRPB-R262, are given in Table 1. The filtration was assumed to be 3 mm Al equivalent and the X-ray tube target angle was assumed to be 16°. This was

Table 1. Technique data used for radiographic projections as input data for PCXMC Monte Carlo calculation. The point 0,0,0 is on the central axis at the base of the trunk. The *x* direction is from right to left, *y* is from front to back, and *z* is upwards and downwards from the base of the trunk

Body region	Projection ^a	Image receptor (cm)		Centring point (cm)			tube voltage (kVp)
		width	height	<i>x</i>	<i>y</i>	<i>z</i>	
Abdomen	AP	35	47	0	0	20	80
Chest	PA	35	44	0	0	52	120
	AP (ward)	35	44	0	0	52	80
Skull	Lat	35	46	0	0	52	120
	PA	24	30	0	0	85.5	75
Lumbar spine	Lat	28	33	0	0	85.5	70
	AP	30	43	0	0	26	75
	PA	30	43	0	0	26	75
	Lat	20	45	0	5	26	85
	LSJ, Lat	18	24	0	5.5	22	95
Pelvis	AP	42	41	0	0	11	75
Urinary tract	AP	24	21	0	0	7	70

^aAll lateral projections are left lateral.

AP, anteroposterior; PA, posteroanterior; LSJ, lumbar sacral joint; Lat, lateral.

required in order for PCXMC to calculate an appropriate X-ray spectrum from this input data using the method of Birch and Marshall [14].

Results

In Table 2, the results of the Monte Carlo calculations are summarized. For each projection, the change in effective dose between the minimum and maximum SID is given for the three collimation cases considered. Since this is an unrealistically wide range of SIDs, the case of a modest increase from 100 cm to 125 cm is also considered, similar to that recommended in a recent publication by Brennan et al [9].

To illustrate the influence of SID on effective dose, data for the lateral chest projection are given in Figure 3a. The curves were fitted using weighted least squares. Second order inverse polynomials were used, because

Table 2. Reduction in effective dose as the source-to-image distance (SID) is increased from 75 cm to 200 cm for collimation to fixed beam sizes at the entrance surface, centre of the patient and image receptor

Body region	Projection	Reduction in effective dose (%)		
		Entrance	Centre	Image receptor
Abdomen	AP	37.9	26.5	9.25
Chest	PA	44.1	32.9	17.3
	AP (ward)	39.6	26.7	2.5
Skull	Lat	68.4	46.2	18.1
	PA	35.4	27.2	14.5
Lumbar spine	Lat	34.3	15.9	0.50
	AP	35.1	19.8	0.60
	PA	35.3	24.8	7.0
	Lat	64.0	29.9	-7.4
	LSJ, Lat	64.0	33.7	-15.2
Pelvis	AP	38.2	33.6	17.9
Urinary tract	AP	45.8	13.7	-22.9

AP, anteroposterior; PA, posteroanterior; Lat, lateral; LSJ, lumbar sacral joint.

Welander and Wickman [6] have shown the integral dose depends on the inverse square factor $\left(\frac{I_y}{I_a}\right)^2$, for collimation to an area other than the image receptor (from Equation 1).

For extreme SIDs, the effect of SID up to 10 m is given in Figure 3b for the lateral LSJ projection, with collimation to a central ROI. The slope of the line of best fit is for data greater than 2.0 m and is $2.0 \times 10^{-4} \text{ mSv.m}^{-1}$ ($1.7\% \text{ m}^{-1}$) with a 95% confidence interval of $\pm 0.8 \times 10^{-4} \text{ mSv.m}^{-1}$.

Discussion

In the UK, NRPB guidance [15] requires that “the field be restricted to the essential area and should always be less than the detector size”. In New Zealand, the Code of Safe Practice [13] requires that “the X-ray beam shall be collimated strictly to the region of clinical interest and in any case shall not exceed the effective cross-section of the cassette or image receptor”. However, the case modelled by some authors, collimation to fixed beam size at the entrance surface, is invalid in many instances because it results in excessive image receptor field sizes at short SIDs and cut-off of clinically significant regions at longer SIDs. The region of clinical interest is not often at or near the entrance surface.

Furthermore, it is good radiographic practice to place the region of clinical interest as close to the image receptor as possible because this minimizes the degree of magnification and also minimizes the focal spot blur [16, 17]. Whilst in the first instance the radiographer will collimate to the image receptor with a small margin so that the collimation marks will show at the edges of the images, it is good practice to reduce the field size further whenever possible as this reduces the volume of tissue irradiated and reduces the scattered radiation to the image receptor, improving the quality of the image [16, 17]. This collimation is generally determined by the location of anatomical markers such as the sternal notch, greater trochanters, xiphisternum, lower costal margin, iliac crest or symphysis pubis. At an accepted standard SID such as 100 cm, collimation to these markers will generally result in the

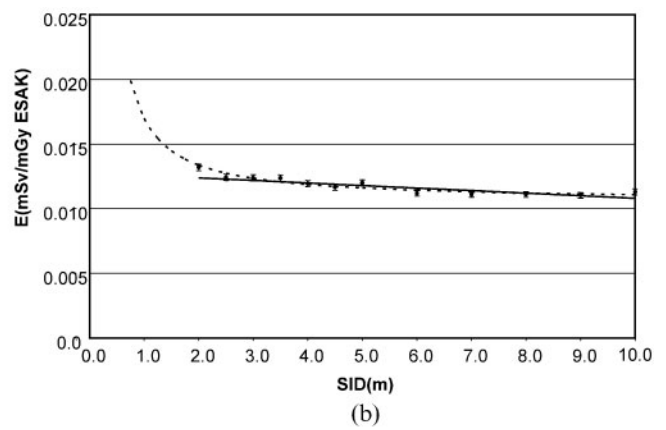
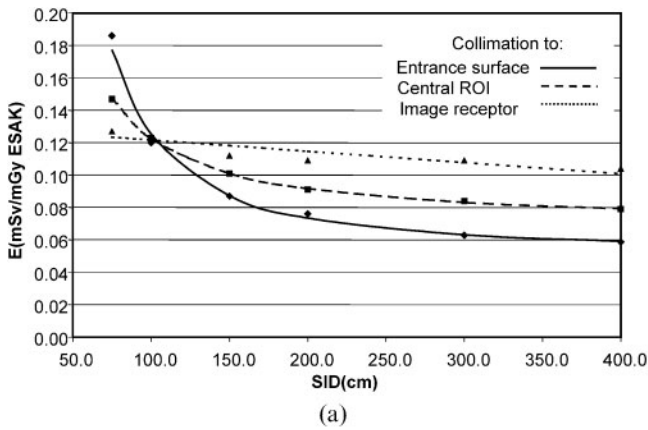


Figure 3. (a) Dependence of effective dose on source to image distance (SID) for the lateral chest projection for three collimation cases. (b) Dependence of effective dose on SID for the lateral lumbar-spinal joint (LSJ) projection for extended SID up to 10.0 m. Collimation is to a central ROI. The line of best fit is for SID of 2.0–10.0 m.

region of clinical interest being demonstrated in the images. This process corresponds approximately to the case in Tables 2 and 3 of collimation to a central ROI. However, this method will be affected if the SID is changed.

If the SID is increased, then in many cases these anatomical markers will no longer be correct and their use may lead to cut-off of part of the region of clinical interest. The case of change in SID for collimation to the entrance surface is shown in Figure 4. Therefore, when changing SID, the markers need to be adapted for the new distance. Generally “rules of thumb” are used such as a distance in centimetres inside or outside the marker. The dose saving at greater SIDs should therefore be estimated from the data in Tables 2 and 3 for the case of collimation to a central ROI, and the anatomical markers must be adapted for each SID.

This present work has been performed using computer simulation. However, Brennan et al [9] have placed much emphasis on the value of experiments compared with

theoretical studies when estimates of patient dose are to be made. While direct experimental verification of theory is always a principal aim in the analysis of scientific problems, difficulties do arise, particularly in aspects of diagnostic X-ray system experimentation. Particularly difficult are the correct maintenance of exact X-ray beam sizes as the SID is changed, the appropriate adjustment of mAs to ensure identical dose to the image receptor at each SID and the exact matching of the optical density of the film for each image. Furthermore, for studies on patients, the normal variation of the patient size and composition causes considerable difficulty, as does the choice of grids with identical performance, focused at each distance. As an example of experimental difficulties, in a recent article involving use of automatic exposure control (AEC) an increase of 60% in the mAs at an increased SID was reported [9]. However the theoretical mAs increase is 69%, suggesting that the dose to the image receptor was poorly controlled by

Table 3. Reduction in effective dose as the source-to-image distance (SID) is increased from 100 cm to 125 cm, for collimation to fixed beam sizes at the entrance surface, centre of the patient and image receptor

Body region	Projection	Reduction in effective dose (%)		
		Entrance	Centre	Image receptor
Abdomen	AP	12.7	8.6	2.9
Chest	PA	16.0	12.0	7.0
	AP (ward)	10.6	6.8	-1.6
	Lat	27.5	17.9	9.7
Skull	PA	8.6	10.8	0.9
	Lat	22.1	6.0	3.4
Lumbar spine	AP	10.9	8.2	5.7
	PA	6.7	1.4	7.8
	Lat	18.9	10.3	-3.0
	LSJ, Lat	24.1	9.9	0.6
Pelvis	AP	8.8	3.6	9.3
Urinary tract	AP	14.5	2.1	-9.6

AP, anteroposterior; PA, posteroanterior; Lat, lateral; LSJ, lumbar sacral joints.

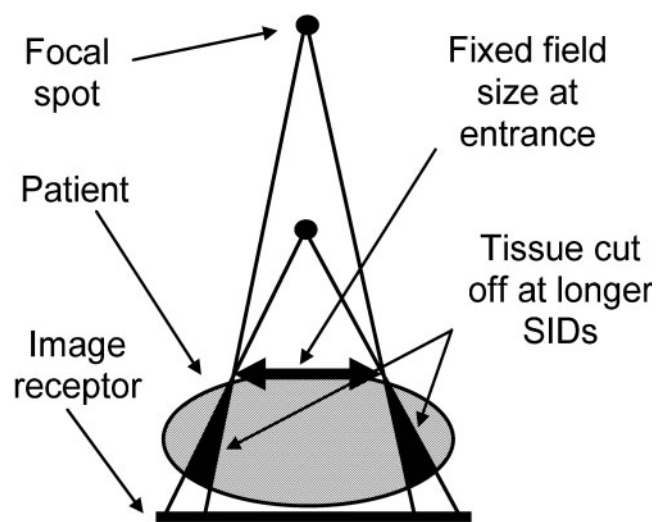


Figure 4. Collimation to a fixed beam size at the entrance surface, showing the cut-off of tissue at greater source-to-image distances (SIDs). In practice, if the SID is increased, the light-field dimensions have to be increased also so as to avoid this cut-off.

the AEC. Consequently, the claimed effective dose reduction appears excessive.

While theoretical studies provide better control of all variables, studies that do not use effective dose as the criterion are not valid because of the variation in radiosensitivity of the organs and tissues, and because the partial irradiation of organs at the edges of the X-ray field may have significant effects. Thus the results of Welander and Wickman [6] using total energy absorbed as the criterion were reasonable in general, but inaccurate in some specific instances.

This work has argued that the optimum technique is to collimate to the region of clinical interest [13, 14]. The data in Tables 1 and 2 show that for the case of collimation to the image receptor there is little or no change in effective dose as the SID is increased. Therefore, if the SID were to be increased under these circumstances as a misguided dose reduction measure, there would actually be little or no reduction in effective dose for most projections and an increase in effective dose for some. This case corresponds to a disturbing trend with computed radiography (CR) systems, where the collimation is to the image receptor for many projections, with subsequent use of the CR system's software collimation to confine the viewed image to the region of clinical interest. Increases in SID will result in no dose reduction in this case. This practice also leads to excessive doses and poor quality images and must be actively discouraged.

The results of the present study demonstrate that most of the effect of SID increase is obtained in the first 200 cm or so. However, to investigate the effect of greater SIDs, PCXMC simulations were performed out to 10 m SID for the lateral LSJ projection. This projection was chosen as it has the greatest dependence of effective dose on SID. For collimation to a central ROI, there was only 1.7% reduction in effective dose per metre increase in SID beyond 200 cm. However, longer SIDs have been suggested, including one article by Hirose et al [18] who studied the use of an SID of 20 m for chest radiography. However, as has been shown in the present study, the dose reduction at such a considerable distance will be minimal. The increase in X-ray tube loading, the distance between the patient and radiographer and the available room sizes would all militate against such an extreme technique. Furthermore, the increased exposure time required would increase the movement blurring in the image, which is an especially important consideration for chest imaging.

From the above discussion it can be concluded that, with optimum radiographic technique requiring collimation to the region of clinical interest, changes in SID require adjustment of the entrance field size for the new technique. This leads to a modest reduction in effective dose, depending on the position of clinical interest within the patient. Extremely long SIDs result in little further dose reduction.

Conclusion

The best radiographic practice is to collimate to a region of clinical interest within the patient, which is generally centrally located and preferably placed as close to the image receptor as possible. If the X-ray beam is collimated to this region as the SID is increased, then there is a modest reduction in effective dose to the patient. However,

the reduction is not nearly as significant as has been claimed by a number of authors, who have incorrectly assumed that the X-ray field size should be of a fixed size at the entrance surface at all SIDs.

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Appendix 1

Additional data

Table A1. The calculated sizes of the region of interest (ROI) for each projection from the data given in NRPB-R262 [7]. Each of the three collimation cases is considered, namely collimation to a fixed X-ray field size at the image receptor, at an ROI centrally located within the patient and at the patient entrance surface. An object to image distance of 5 cm was used

Body region	Projection	Field size at image receptor (cm)		Central ROI distance ^a	Size of central ROI (cm)		Size of surface ROI (cm)	
		Width	height		width	height	width	height
Abdomen	AP	35	47	15.0	29.8	40.0	26.3	35.3
Chest	PA	35	44	15.0	29.8	37.4	26.3	33.0
	AP	35	44	15.0	29.8	37.4	26.3	33.0
Skull	Lat	35	46	22.2	27.2	35.8	21.2	27.9
	PA	24	30	15.0	20.4	25.5	18.0	22.5
L-spine	Lat	28	33	13.0	24.4	28.7	22.1	26.1
	AP	30	43	15.0	25.5	36.6	22.5	32.3
	PA	30	43	15.0	25.5	36.6	22.5	32.3
	Lat	20	45	22.2	15.6	35.0	12.1	27.3
Pelvis	LSJ, Lat	18	24	22.2	14.0	18.7	10.9	14.5
	AP	42	41	15.0	35.7	34.9	31.5	30.8
Urinary tract	AP, C-	24	21	15.0	20.4	17.9	18.0	15.8
	AP, C+	24	21	15.0	20.4	17.9	18.0	15.8

^aFrom image receptor.

AP, anteroposterior; PA, posteroanterior; Lat, lateral; LSJ, lumbar sacral joint.

Table A2. An example of the actual X-ray field size at the entrance surface for each source-to-image distance (SID), for the three collimation cases considered in this paper; AP abdomen projection only. The region of interest (ROI) dimensions for each case were determined by simple geometry from the data given in Table A1, for which the X-ray field size at the image receptor is given as 35 cm × 47 cm for this projection

SID (cm)	Light field dimensions in cm					
	Collimation to image receptor at each SID		Collimation to central ROI at each SID		Collimation to entrance surface at each SID	
	width	height	width	height	width	height
75	23.3	31.3	24.8	33.3	26.3	35.3
100	26.3	35.3	26.3	35.3	26.3	35.3
125	28.0	37.6	27.1	36.3	26.3	35.3
150	29.2	39.2	27.6	37.0	26.3	35.3
175	30.0	40.3	27.9	37.5	26.3	35.3
200	30.6	41.1	28.1	37.8	26.3	35.3