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# Modification of Transmission Dose Algorithm for Irregularly Shaped Radiation Field and Tissue Deficit

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Our transmission dose estimation algorithm for in vivo dosimetry was modified for use in partially blocked radiation fields and in cases with tissue deficit by using the beam data measured with flat solid phantom in various conditions of beam block and tissue deficit. The developed correction algorithm for irregularly shaped field could accurately reflect the effect of beam block, with error within ±1.0%. The correction algorithm for tissue deficit could accurately reflect the effect of tissue deficit with errors within ±1.0% in most situations and within ±3.0% in experimental settings with irregular contours mimicking breast cancer treatment set-up. Thus, developed algorithms could accurately estimate the transmission dose in most radiation treatment settings including irregularly shaped field and irregularly shaped body contour with tissue deficit.
***KEY WORDS: in vivo dosimetry, algorithm, transmission dose, irregulary shaped field, block, tissue deficit, irregular contour***

## I. Introduction

In radiation therapy, error in irradiated dose is not rare[1-4]. Measurement of transmission dose is useful for in vivo dosimetry of QA purpose. Authors already developed an algorithm for estimation of tumor dose using measured transmission dose for open rectangular radiation field without phantom deficit[5]. In this study, the algorithm for estimation of transmission dose was modified for use in partially blocked radiation fields and in cases with tissue deficit

## II. Materials and Methods

### 1. Standard conditions of measurements

Transmission dose was measured with various field size (FS), phantom thickness (Tp), and phantom chamber distance (PCD) with a acrylic phantom for 6 MV and 10 MV X-ray. Source to chamber distance (SCD) was set to 150 cm. Size of single acrylic phantom slice was 40 cm x 57.5 cm with 1 cm thickness and the density of acrylic phantom was 1.17 g/cm$^3$. Various phantom thickness was made by stacking phantom slices. Measurement was conducted with a 0.6 cm$^3$ Farmer type ion chamber.

To exclude the influence of temperature, pressure, and output variation of linear accelerator on measurement results, more than 3 measurements were made under reference condition (i.e., FS 10 cm x 10 cm, Tp=0). Average value of measurements under reference condition was defined as reference reading ($D_0$), and each measured values divided by reference reading and multiplied by 10000 were defined as corrected readings, which were used for analysis.

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Through analysis using measured data, correction algorithms were developed for estimation of expected reading of transmission dose.

### 2. Irregularly shaped radiation field

Basic measurements of transmission dose were made by using various size of collimator opening (defined as size at SSD 100 cm), Tp, and PCD. Sixteen steps of collimator openings were used (i.e. from 2 cm x 2 cm to 32 cm x 32 cm with 2 cm increments for each dimensions) for 6 MV and 10 MV X-ray energies, respectively. Used Tp were 0, 10, 20, and 30 cm. PCD were 10, 30, and 50 cm (**Fig. 1**). For all of these measurement conditions, a portion of radiation field by collimator opening was shielded by beam block (**Fig. 1**).

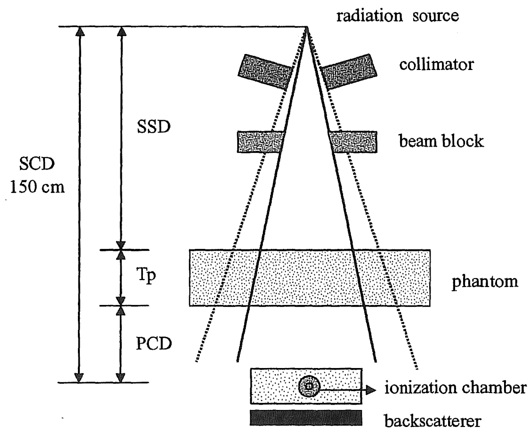


**Fig. 1** Geometric relationship between radiation source, collimator, shielding block, phantom, and ion chamber. (SCD: source chamber distance, SSD: source surface distance, Tp: phantom thickness, PCD: phantom chamber distance.)

Beam block was made with cerrobend and its thickness was 7.5 cm. Used effective FS (a portion of radiation field which was not shielded by block, defined at SSD 100 cm) were 5 x 5, 10 x 10, 15 x15, and 20 cm x 20 cm, which were embodied by beam blocks with same size of rectangular block openings, respectively.

Thus, for correction algorithm for irregularly shaped radiation fields, basic measurements were made for 400 kinds of conditions for each X-ray energy.

After the formulation of correction algorithm, the accuracy of the algorithm was tested for conditions of basic measurements of this study.

### 3. Tissue deficit

Measurements of transmission dose were made by using various size of collimator opening (defined as size at SSD 100 cm), and PCD (**Fig. 2**). Tp was fixed to 20 cm. Used collimator openings were 10 x 10, 20 x 20, and 30 cm x 30 cm. PCD were 10, 30, and 50 cm. In this basic experiment for correction algorithm for phantom deficit, phantom deficit was made to range from 0 to 100% of radiation field (**Fig. 2**).

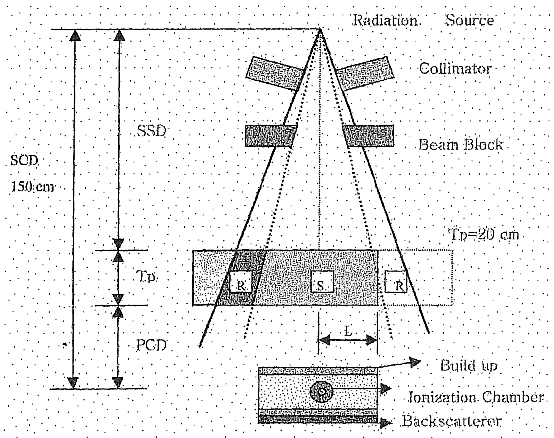


**Fig. 2** Geometric relationship between radiation source, collimator, shielding block, phantom, and ion chamber. (SCD: source chamber distance, SSD: source surface distance, Tp: phantom thickness, PCD: phantom chamber distance, L is the distance between phantom edge and central axis (CA) of radiation field)

After the formulation of correction algorithm, the accuracy of the algorithm was tested for conditions of basic measurements of this study.

Then, to test the correction algorithm for real cases with phantom deficit and irregular contours, we used stacked slices of acrylic phatoms simulating breasts of five real patients, in which two undergone mastectomy and three undergone quadrantectomy (**Fig. 3**). In these five cases simulating real treatment conditions, the accuracy of correction algorithm was tested

## III. Results and Discussion

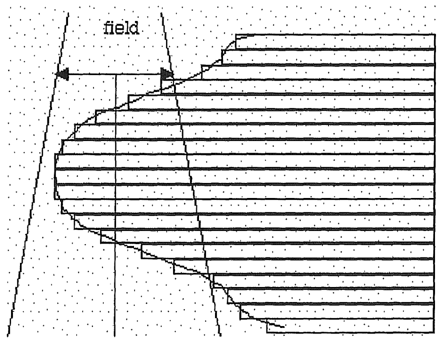


**Fig. 3** Diagrams for stacked slices of acrylic phatoms simulating breasts

### 1. Irregularly shaped radiation field

When collimator opening was smaller than block opening, there was no shielding effect so algorithm for open radiation field was applied without any correction. When collimator opening was larger than block opening (i.e., effective field size in these occasions), correction algorithm was required to match the results of the measurements. In these occasions, when we used fixed block opening, transmission dose was increased as the collimator opening was increased, but the rate of increment was far smaller than the rate of increment without shielding block (**Fig. 4**).

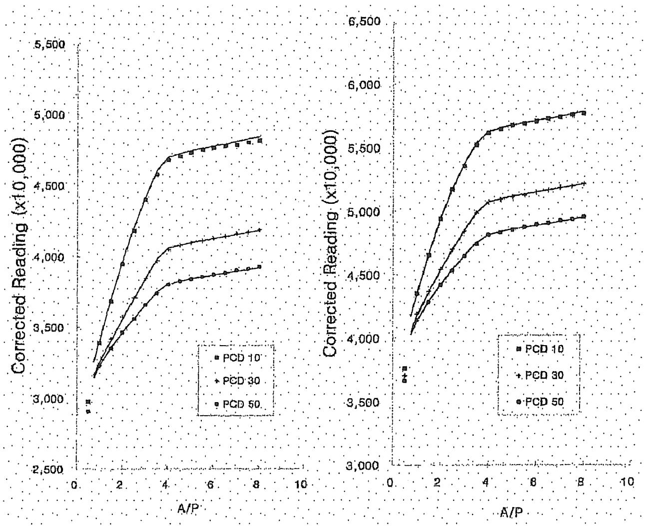


**Fig. 4** Measured and calculated transmission dose after correction of partial field block. (Tp=20 cm, effective FS = 15 x 15 cm, Rt; 6 MV X-ray, Lt; 10 MV X-ray). The marks indicate measured data and the lines indicate calculated dose.

So, we could find the fact, in these occasions, only a portion of increment of output of linear accelerator (i.e., Sc; collimator scatter factor) resulted in the increment of transmission dose. By using these results, authors formulated correction algorithm for blocked (irregular shaped) radiation field.

$$Db = De\ \{1 + (Sc_o - Sc_e)/Sc_e \times f\} \quad (1)$$

where, Db: transmission dose when collimator opening was larger than block opening.

De: transmission dose when collimator opening equals to block opening.

Sco: collimator scatter factor (Sc) when collimator opening was larger than block opening.

Sce: collimator scatter factor (Sc) when collimator opening equals to block opening.

f : correction factor

We calculated f values minimizing average values of absolute errors for all measured conditions (including Tp=0, 10, 20, and 30 cm) for each x-ray energy and block opening. Values of correction factor (f) were about 0.6 for 5 cm x 5 cm sized block opening, about 0.8 for 10 cm x 10 cm sized block opening, and about 1 for 15 x 15 and 20 cm x 20 cm sized block opening, and the f values were same for 6 MV and 10 MV X-ray.

Comparison between calculated reading by our correction algorithm and measured (corrected) reading exemplified the case of Tp=20 cm, effective FS 15 cm x 15 cm for 6 MV and 10 MV X-ray, respectively **(Fig. 4)**. The calculated readings (lines) and measured (corrected) readings (dots) agreed each other accurately. The result was same for all other conditions of 6 MV X-ray and all conditions of 10 MV X-ray.

As a whole, errors between calculated reading by using correction algorithm and measured (corrected) reading were under ±1% for 98.9% of all conditions of basic measurements, and errors were under ±0.5% for 77.5% of all conditions. Of 9 conditions with error over 1%, 8 conditions were conditions with 5 cm x 5 cm sized block opening and over 20 x 20 cm sized collimator opening, which were clinically unfeasible **(Table 1)**.

**Table 1** Distribution of error between measured and estimated transmission dose

| Energy | Block opening | No. of measured conditions | Distribution of absolute error (%) < 0.5% | 0.5% ~ 1.0% | 1.0% ~1.5% |
|---|---|---|---|---|---|
| 6 MV | 5x5 cm | 140 | 84 (60.0%) | 51 (36.4%) | 5 (3.6%) |
| | 10x10 cm | 110 | 91 (82.7%) | 18 (16.4%) | 1 (0.9%) |
| | 15x15 cm | 90 | 74 (82.2%) | 16 (17.8%) | 0 (0.0%) |
| | 20x20 cm | 60 | 49 (81.7%) | 11 (18.3%) | 0 (0.0%) |
| 10 MV | 5x5 cm | 140 | 94 (67.1%) | 43 (30.7%) | 3 (2.2%) |
| | 10x10 cm | 110 | 86 (78.2%) | 24 (21.8%) | 0 (0.0%) |
| | 15x15 cm | 90 | 88 (97.8%) | 2 (2.2%) | 0 (0.0%) |
| | 20x20 cm | 60 | 54 (90.0%) | 6 (10.0%) | 0 (0.0%) |
| Total | | 800 | 620 (77.5%) | 171 (21.4%) | 9 (1.1%) |

By applying interpolated f values between 5 x 5 and 15 cm x 15 cm sized block opening and by using f=1 for over 20 cm x 20 cm sized block opening, we could estimate measured values accurately.

In the accuracy test with the real irregularly shaped blocks in suitable radiation conditions, errors were under ±1% for all 6 cases for 6 MV X-ray and all 7 cases for 10 MV X-ray **(Table 2)**.

**Table 2** Accuracy of algorithm in actual treatment conditions with partially blocked radiation field. (10 MV X-ray, phantom study).

| Case | FS (cm x cm) | EFS (cm x cm) | Tp (cm) | PCD (cm) | Transmission dose* estimated | measured | Error (%) |
|---|---|---|---|---|---|---|---|
| lung cancer 1 | 15x23 | 14x22 | 20 | 38.8 | 5055 | 5030 | −0.49 |
| lung cancer 2 | 20x23 | 17x22 | 20 | 40.4 | 5135 | 5155 | 0.39 |
| uterine cervix cancer 1 | 15x15 | 13.5x15 | 16 | 40.5 | 5606 | 5650 | 0.78 |
| uterine cervix cancer 2 | 16x16 | 14x16 | 16 | 40.1 | 5656 | 5697 | 0.74 |
| mantle block | 32x35 | 27x26 | 20 | 30.4 | 5660 | 5618 | −0.73 |
| 40% block | 24x22 | 17x19 | 18 | 40.0 | 5442 | 5446 | 0.07 |
| 50% block | 25x20 | 14x18 | 18 | 40.0 | 5320 | 5291 | −0.55 |

T p : phantom thickness, PCD : phantom-chamber distance.

* = (Measured transmission dose)/(Reference reading)×10,000

## 2. Tissue deficit

Measurement conditions of tissue deficit experiment, measured transmission dose was increased for L > 2 cm (where L is the distance between central axis of radiation field and nearest edge of the phantom) until phantom deficit disappeared **(Fig. 5)**. Such an increment in measured dose was related with the increase of scattered ray by the phantom, which was related to the volume of phantom included in radiation field (i.e., size of effective field size). By using of these physical theory and considering the fact that, in the phantom deficit experiment, center of radiation field was not accord with center of radiation field (on the other hand, two centers were coincided in the experiments of open rectangular radiation field and irregularly shaped radiation field without phantom deficit), we developed correction algorithm after some algebra

In **Fig. 2**, a portion of radiation field (shaded R and S portion) is filled with phantom, and the rest of the radiation field is lack of the phantom. For derivation of the correction algorithm for tissue deficit, we defined $D_O$, $D_S$, $Sc_O$ , $Sc_S$ values as follows. Scs : collimator scatter factor where collimator is opened as dotted line in **Fig. 2**.

Sco : collimator scatter factor where collimator is opened as solid line in **Fig. 2**.

Ds : transmission dose where collimator scatter factor is Scs in **Fig. 2**.

Do : transmission dose where collimator scatter factor is Sco and entire radiation field is filled with phantom

And, also defined S, R as the scattering contribution by S, R portions of phantom, respectively **(Fig. 2)**.

Then, $D_O$= (primary beam portion) + (phantom scattering portion)= $\alpha \times Sc_O + \beta \times (S+2R) \times Sc_O$ (2)

$D_S = \alpha \times Sc_S + \beta \times S \times Sc_S$ (3)

(where $\alpha$ , $\beta$ ; coefficients)

$D_O$, $D_S$, $Sc_O$ and $Sc_S$ values could be calculated by using basic algorithm and basic data for open radiation fields.

If $D_{SR}$ was defined as transmission dose of given geometrical setting of phantom deficit,

$$D_{SR} = \alpha \times Sc_O + \beta \times (S+R) \times Sc_O \quad (4)$$

By using Eq. (2), (3), (4), We could remove S, R.

Then, we could get

$$D_{SR} = (Sc_S \times D_O + Sc_O \times D_S)/(2 \times Sc_S) \quad (5)$$

Comparison between calculated readings by our correction algorithm and measured (corrected) readings exemplified the case of 6 MV X-ray, PCD=30 cm (**Fig. 6**), and errors between the calculated (lines) and measured readings (dots) were under ±1% for 16 of all 17 conditions where L was over 3 cm.

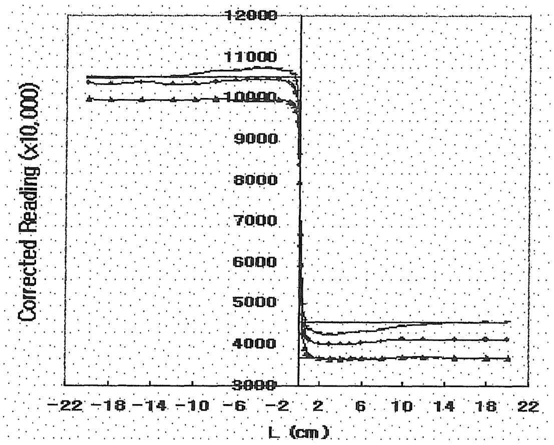


**Fig. 5** Transmission dose in case of tissue deficit (6 MV X-ray, Tp=20 cm and PCD=30 cm). L is the distance between phantom edge and central axis (CA) of radiation field. The marks indicate measured data. Horizontal lines indicate estimated dose for Tp=0 (left half) and Tp=20 cm (right half), respectively, for each field size. (plus value of L means CA traverse the phantom, while minus value of L means CA doesn't traverse the phantom).

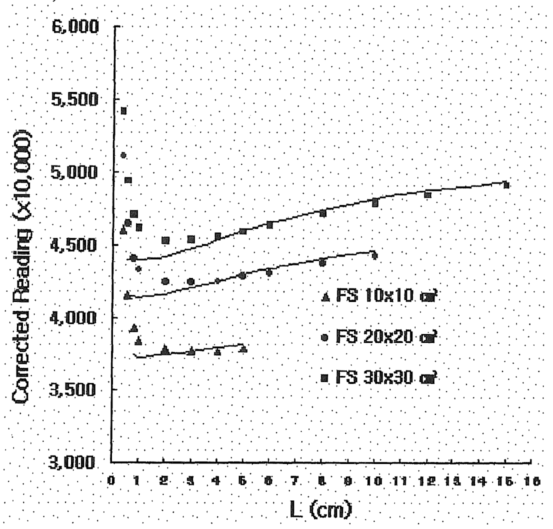


**Fig. 6** Measured and calculated transmission dose using correction algorithm for tissue deficit (6 MV X-ray, Tp=20 cm, PCD=30 cm). The marks indicate measured data and the lines indicate estimated dose.

For 6 MV X-ray, errors between the calculated and measured readings were under ±1% for 49 (98.0 %) of all 50 conditions where L was over 3 cm (for all PCD=10, 30, and 50 cm).

For 10 MV X-ray, errors between the calculated and measured readings were under ±1% for 48 (96.0 %) of all 50 conditions where L was over 3 cm.

In the experiment using phantom simulating breasts, measured dose was calculated by the correction algorithm with errors under ±3% in all cases for 6 MV X-ray and 10 MV X-ray (**Table 3**).

**Table 3.** Accuracy of algorithm in actual treatment conditions with tissue deficit(6 MV X-ray, phantom study).

| Case | Type of surgery | FS (cm x cm) | PCD (cm) | Tp (cm) | L (cm) | Transmission dose estimated | Transmission dose measured | Error (%) |
|---|---|---|---|---|---|---|---|---|
| 1 | CS | 10x20 | 39.8 | 12 | 3.3 | 5646 | 5522 | -2.19 |
| 2 | CS | 10x19 | 38.8 | 14 | 4.2 | 5147 | 4993 | -2.99 |
| 3 | CS | 10x18 | 33.3 | 14 | 2.8 | 5149 | 5086 | -1.23 |
| 4 | M | 7x19 | 37.0 | 16 | 1.95 | 4470 | 4557 | +1.95 |
| 5 | M | 6x18 | 35.9 | 13 | 1.9 | 5153 | 5131 | -0.43 |

M : mastectomy, CS : conservative surgery,
FS : field size, PCD : phantom-chamber distance,
L : distance between breast edge and central axis of radiation field.

## IV. Conclusion

The algorithm for correction of beam block could accurately reflect the effect of beam block, with error within ±1.0%, both with square fields and irregularly shaped fields.

And, the correction algorithm for tissue deficit could accurately reflect the effect of tissue deficit with errors within ±1.0% in most situations and within ±3.0% in experimental settings with irregular contours mimicking breast cancer treatment set-up.

Conclusively, developed algorithms could accurately estimate the transmission dose in most radiation treatment settings including irregularly shaped field and irregularly shaped body contour with tissue deficit in transmission dosimetry.

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