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VARIATION IN THE LUNG INHOMOGENEITY CORRECTION FACTOR WITH BEAM ENERGY

Clinical implications

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Abstract

In order to determine the magnitude of the dosimetry error introduced by failing to correct for increased transmission through lung tissue in treating thoracic malignancies, measurements in a phantom were taken using different field sizes, inhomogeneity thicknesses and photon qualities. The results indicate that the error introduced by neglecting the inhomogeneity correction is greatest at lower photon energies, smaller field sizes and greater thickness of inhomogeneity. Correction factors to account for the lung inhomogeneity were obtained from phantom measurements and were compared with those calculated using the tissue-air ratio and Batho-Young algorithms; correlation coefficients describing the relationship between measured and calculated values exceeded 0.995. The calculated values tended to overestimate the correction factor and differed most from the measured correction factors at lower energies, smaller field sizes, and greater inhomogeneity thicknesses. The importance of these results in clinical radiation therapy is discussed.

Single and multiple institution studies of the radiation treatment of lung carcinoma include patients treated with photon beams of different qualities, from ^{60}Co and 4 MeV linear accelerators to 45 MeV betatrons (1, 4, 5), without correcting the prescribed target absorbed dose for increased transmission through lung tissue. This can lead to errors in actual target absorbed dose, the magnitude varying with photon energy, field size, treatment depth and lung tissue density. Because of this, unless dosimetry includes a correction for increased lung transmission, correlating tumor control and complication rates with target absorbed doses delivered using different photon energies can be misleading, and adoption of a dose giving an acceptable rate of tumor control and complications using photons of one energy can lead to overdosage or underdosage if used with photons of a different energy.

We have measured and calculated correction factors for increased lung transmission at a variety of depths and field sizes for three photon energies. In addition, using a commercial treatment planning computer, we have constructed treatment plans for photons of different energies using actual patient contours obtained from CT slices, with and without correction for increased lung transmission, to demonstrate the clinical significance of failure to correct for increased lung transmission when patients treated with different photon energies are compared.

Methods

Measurements were made using the following photon beam qualities: ^{60}Co gamma rays (AECL Theratron 780), 10 MV roentgen rays (Varian Clinac 18) and 25 MV roentgen rays (Varian Clinac 35).

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The phantom and ion chamber arrangement included a top layer of polystyrene (density 1.04 g/cm^3) 4 cm thick to simulate the chest wall, an intervening variable thickness of either polystyrene to simulate soft tissue, or cork (density 0.24 g/cm^3) to simulate lung tissue, and a 0.6 cm^3 Farmer type chamber placed at isoscenter, 2.5 cm from the surface of the underlying polystyrene pancake type SHM phantom (EMI Ltd.). There were 12.5 cm of polystyrene beneath the ionization chamber to provide back scatter. The measurement geometry is shown in Fig. 1. Measurements were made at field sizes of 7 cm × 7 cm, 10 cm × 10 cm, 8 cm × 20 cm and 20 cm × 20 cm, defined at isocenter (100 cm for the linear accelerators; 80 cm for the ^{60}Co unit); these field sizes were chosen to encompass those used clinically in our department to treat lung carcinoma. At the largest field size, the phantom extended at least 5 cm beyond the field margins. Ion chamber readings were expressed as a percentage of the readings obtained with the polystyrene 'chest wall' in place (i.e. at 6.5 cm depth) relative to those with various thickness of intervening polystyrene or cork. Tissue-phantom ratio curves as a function of the thickness of cork or polystyrene were plotted on semilogarithmic paper and curves were fitted by hand. The measured correction factors were the ratios of the dose delivered with cork between 'chest wall' and phantom divided by the dose delivered with an equal thickness of polystyrene between 'chest wall' and phantom. Correction factors were also calculated using the tissue-air ratio algorithm and the power law tissue-air ratio algorithm of Batho and Young (2) and were compared with the measured correction factors (see Appendix).

Treatment planning was performed with a treatment planning computer (AECL TP-11) using beam data measured in our department in a water phantom. The program employed uses the Batho-Young algorithm to correct for inhomogeneities in the treatment volume. A two-field AP-PA parallel opposed portal arrangement and a three-field AP-RPO-LPO portal arrangement with central axes 120° from another, with 10 cm × 10 cm portals, were chosen for evaluation at all three photon energies. For the three-field portal arrangement, the correction factor is quoted for the dose at isocenter. For the two-field, AP-PA plan, the central axis did not traverse lung tissue, so an off-axis point, corresponding to the position of the pulmonary hilum, was chosen. The portal arrangements are shown in Fig. 2, as are the locations of the points chosen for calculating dose. For the 5 patient contours used, the mean AP separation through the isocenter was 26.9 cm, and the mean combined lung thickness traversed by the anterior and posterior beams in irradiating the hilar point was 11.6 cm. The mean distance from skin surface to isocenter for the 120° posterior oblique fields was 20.9 cm, traversing a mean lung thickness of 9.8 cm. Because the computer algorithm was not

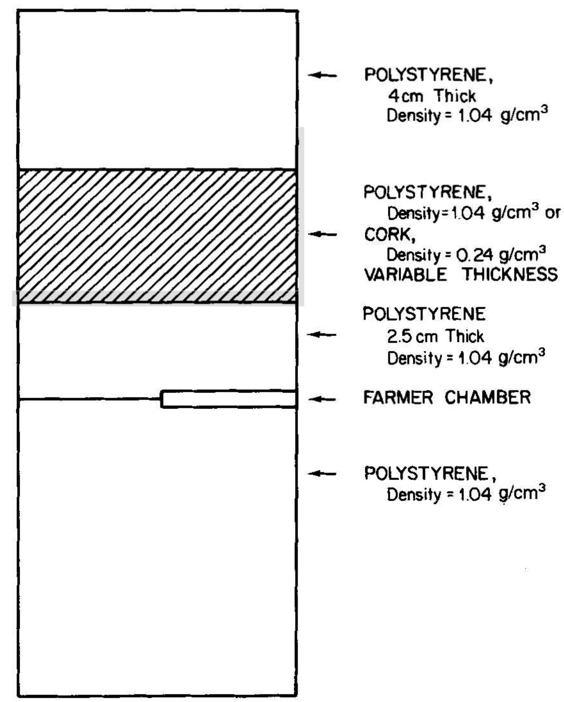

Fig. 1. Geometry for phantom irradiation.

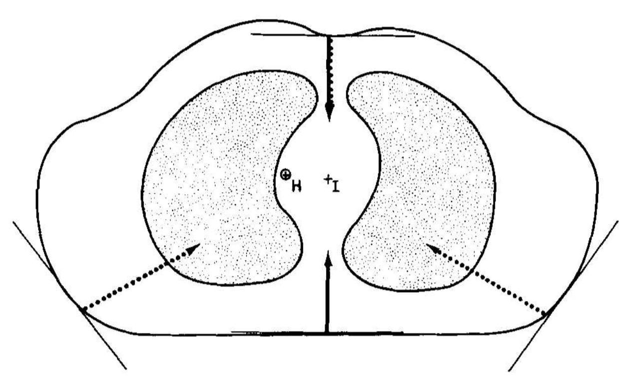

Fig. 2. Geometry for patient irradiation. AP-PA portals measure 10 cm × 10 cm at the isocenter, set at midplane. AP-PA central axes are indicated by solid arrows. Posterior oblique portals measure 10 cm × 10 cm at isocenter, and their axes are each 120° from the axis of the AP portal; AP, RPO, LPO central axes are indicated by dotted arrows. Point I denotes the dose calculation point at isocenter. Point H represents the hilar dose calculation point.

Table 1

Measured and calculated correction factors for ^{60}Co, 10 MV and 25 MV roentgen rays at 10 and 20 cm depth

Total depth (cm)	Field size (cm^2)	Photon quality	Correction factor		
			Measured	Batho-Young method	Tissue-air ratio method
10	7×7	^{60}Co	1.13	1.16	1.16
	10×10		1.12	1.14	1.14
	8×20		1.12	1.14	1.14
	20×20		1.11	1.11	1.12
20	7×7		1.68	1.80	1.82
	10×10		1.60	1.73	1.76
	8×20		1.59	1.71	1.73
	20×20		1.48	1.57	1.60
10	7×7	10 MV	1.08	1.08	1.09
	10×10		1.07	1.08	1.08
	8×20		1.07	1.08	1.08
	20×20		1.06	1.06	1.06
20	7×7		1.33	1.41	1.42
	10×10		1.30	1.38	1.39
	8×20		1.29	1.37	1.38
	20×20		1.26	1.31	1.32
10	7×7	25 MV	1.04	1.04	1.04
	10×10		1.04	1.04	1.04
	8×20		1.04	1.04	1.04
	20×20		1.04	1.04	1.04
20	7×7		1.19	1.24	1.26
	10×10		1.18	1.23	1.23
	8×20		1.18	1.22	1.23
	20×20		1.18	1.21	1.22

Table 2

Correction factors for target absorbed dose with two and three field plans

	Two fields*	Three fields
^{60}Co	1.22±0.04	1.15±0.02
10 MV roentgen rays	1.11±0.02	1.08±0.01
25 MV roentgen rays	**	1.03±0.01

* An off-axis point within the mediastinum was chosen, since the central axis traversed no lung tissue.

** Doses not available within buildup region.

capable of calculating doses within the buildup region, correction factors are not available for the AP-PA plan using 25 MV roentgen rays. In order to determine the variability in the correction factor among different patients, computed tomographic slices from 5 patients were used to generate contours and lung volumes. A representative lung density of 0.27 g/cm^3 was chosen (7). Since this lung

Table 3

Correction factor dependence upon photon energy and lung density

Lung density (g/cm^3)	Photon energy		
	^{60}Co	10 MV	25 MV
AP-PA plan			
0.15	1.20	1.11	*
0.18	1.19	1.11	*
0.27	1.17	1.09	*
0.35	1.15	1.08	*
0.47	1.12	1.07	*
AP-RPO-LPO plan			
0.15	1.16	1.08	1.03
0.18	1.15	1.08	1.03
0.27	1.12	1.06	1.03
0.35	1.11	1.06	1.02
0.47	1.09	1.05	1.02

* Doses not available within buildup region.

density represented only a mean value, we calculated correction factors on a single patient contour, using hypothetical lung densities from 0.15 g/cm^3 to 0.47 g/cm^3 (7), to determine the effect of different lung densities on the correction factor in this clinical situation.

Results

Ion chamber readings were reproducible with a mean standard error of 0.1 per cent. Measured and calculated correction factors at representative depths for ^{60}Co, 10 MV and 25 MV roentgen rays are shown in Table 1. Correlation coefficients describing the relationship between the dose measured and the dose calculated with either algorithm exceeded 0.995. As seen in Table 1, the magnitude of the correction factor, representing the ratio of target absorbed dose with and without lung correction, increased as the field size decreased, as inhomogeneity thickness increased, and as photon energy decreased.

Using the tissue air ratio and Batho-Young algorithms, the correction factor was calculated for lung densities varying from 0.135 g/cm^3 to 0.47 g/cm^3. Changes in lung density yielded the greatest charges in calculated correction factors at lower photon energy, greater thickness of inhomogeneity, and larger field size.

Correction factors using patient contours are shown in Table 2. With the AP-PA portal arrangement and a lung density of 0.27 g/cm^3, failure to correct for increased transmission through lung leads to an increase in target absorbed dose of 20 per cent with ^{60}Co and 11 per cent with 10 MV roentgen rays. For the AP-RPO-LPO portal arrangement, the increases in target absorbed dose were 12 per cent with ^{60}Co, 8 per cent with 10 MV and 3 per cent with 25 MV roentgen rays. Because the lung density chosen, 0.27 g/cm^3, is somewhat arbitrary since it represents only a mean lung density, correction factors using a single patient contour were computed while varying the hypothetical lung densities between 0.15 g/cm^3 and 0.47 g/cm^3. These results are shown in Table 3.

Several points bear emphasis in conclusion. First, the increased target absorbed dose due to increased transmission through lung tissue is greatest at low photon energies; it also is greatest with small field sizes and greater thickness of inhomogeneity. Secondly, in clinical situations, unless doses are corrected for increased transmission through lung tissue, the actual target absorbed dose may exceed the expected dose by up to 15 per cent. Thirdly, the correction factor varies with lung density and inhomogeneity thickness as expected; the variation in correction factor with lung density is greatest at low photon energies and large field sizes. Finally, the error introduced by the uncertainty in lung density is minor compared with the error introduced by failing to correct for increased transmission through lung tissue in any way.

Because target absorbed dose varies with photon energy unless corrected for increased transmission through lung tissue, nominally equivalent uncorrected doses given with ^{60}Co gamma rays and 25 MV roentgen rays can differ by more than 10 per cent, rendering any comparison of results valueless. Similarly, unless corrected for increased transmission through lung, a dose shown to be safe and clinically effective at one photon energy could result in increased complications or treatment failures if given with photons of a different energy (3, 6).

Correcting for increased transmission through lung allows comparison of data for tumor control and complications obtained in treating patients with different photon energies, and allows doses used with one photon energy to be extrapolated to different photon energies.

Appendix

Correction factors are calculated using the two following algorithms.

1) TAR ratio algorithm

$$CF = \frac{TAR(d-((1-\varrho)^* A))}{TAR(d)}$$

2) Batho-Young algorithm

$$CF = \left[\frac{TAR(d)}{TAR(d-t)}\right]^{1-\varrho}$$

d = depth of point under consideration
t = thickness of inhomogeneity
ϱ = density of inhomogeneity

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