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TISSUE AIR RATIO IN TOTAL BODY IRRADIATION

An in vivo evaluation

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Abstract

On the basis of dose readings in 102 patients treated with total body irradiation (TBI), a 'tissue air ratio (TAR) curve' has been produced. It could be useful to precalculate treatment time in TBI, for dose prescription to a specific point, provided the same source (^{60}Co) and treatment setting (lateral irradiation; 3 m source–axis distance; reference point at thighs bifurcation, near the perineum) is used. The TAR curve produced, and the formula relating tissue depth to TAR value, are presented, and compared to preexisting data for 'magna fields' treatments. This curve is exponential, and in semilog representation becomes straight, as every classic TAR curve; it is lower than others, reflecting non full-scatter situation in patient irradiation.

Key words: Total body irradiation, dosimetry, tissue air ratio.

Total body irradiation (TBI) and hemibody irradiation (HBI) are increasingly used to treat advanced diseases. In addition, TBI plays an important role in treatment regimens involving bone marrow transplantation.

The implementation of TBI and HBI presents a dosimetry challenge. In conventional radiotherapy treatment target dose is calculated on well established depth dose distribution data and only occasionally is the actual dose delivered confirmed 'in vivo'; but common dosimetry concepts can in no way be applied to field sizes and treatment distances involved in TBI and HBI. Moreover, since patients undergoing TBI occupy part of the radiation field instead of presenting a uniform flat surface to the beam, an entirely new approach to treatment is required. The delivery of a uniform radiation dose to all parts of the patient's body is difficult (1, 2) and there is no consensus on the methods of depth dose determination for TBI or HBI. It is therefore customary to measure the dose actually delivered to each patient at one or various reference points; on this basis depth dose is sometimes calculated (3).

In 'magna field' treatments, a general formula relating beam energy, source-to-patient distance and patient dimensions to the absorbed dose is lacking. It is therefore apparent that a method for HBI and TBI treatment planning similar to those already applied in other situations would be extremely useful.

On the basis of experimental dose measurements for a single specific and easily reproducible TBI treatment geometry, we developed a tissue-air ratio (TAR) curve for ^{60}Co radiation, that can be used to calculate treatment time prior to the actual irradiation setting, provided the same physical and geometric conditions are met.

Material and Methods

At the Institute of Radiology of the University of Genoa from August 1983 to December 1985 102 patients, 60 males and 42 females, with a median age of 24 years (range 1 to 48 years), underwent TBI prior to bone marrow transplantation. The irradiation was carried out with a ^{60}Co unit (Theratron 780, AECL), with horizontal beam, at 3 m source–axis distance, with a field size of 131×131 cm^2. The patient was irradiated bilaterally, half dose per side, lying on a couch, in a semisupine position, with the lower limbs adduced and flexed, so that the entire body was comprised in the field.

Lungs were partially shielded by a 5 mm lead block and the eyes and mouth were shielded by an 11 mm lead block. In pediatric patients, the thickness of the blocks was 3 and 6 mm respectively.

The couch was 80 cm high and 180 cm from the wall, so virtually no scatter from floor and walls interfered with the 'in air' dose reading. This was performed with the

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same dosimeters and in the above described geometric conditions. Readings with and without couch and lead shields showed no relevant differences (less than 2%); 'in air' dose rate without couch and shields was employed for subsequent calculations.

Dose was measured with an ionization chamber (PTW DI-DL4 with Condenser R meter model 570, or 25 R Victoreen Condenser Chamber). Energy response of these chambers was considered negligible (~1% and ~3% calibration factors difference, below 100 keV and over 1200 keV respectively); the thimble was placed at the reference point, that is the bifurcation of the thighs, near the perineum, with the stem fixed along the right thigh.

Doses and fractionations employed were: 9.9 Gy/3 fractions/3 days, 12 Gy/6 fr/3 days, 10 Gy/1 fr/1 day, and 12 Gy/3 fr/3 days.

Four to 12 readings per patient were obtained with the chamber at the reference point.

Body surface dose was also measured with pairs of thermoluminescent dosimeters (TLDs), at the forehead, cheek, axilla, anterior thorax, abdomen, hip and ankle. Non-uniformity of dose was as expected in such conditions (2), with (mean) +15% dose at the ankle and +17% at the forehead. Mouth and eye dose (under lead shield) was −2%. Lung dose, under lead shield, as measured at the thoracic surface, was +8%, and −7% at the underarm.

Dose rates (Gy/min) read with the chamber were corrected, as usual, for temperature and pressure; all values obtained were used to calculate a mean dose rate per patient. This was divided by the dose rate in air at the same place and time. 102 TAR values were thus obtained, and correlated to tissue depth (half 'bitrochanteric diameter', i.e. lateral patient diameter at the level of the reference point; in our group of patients, this ranged from 15 to 39 cm). All the data were plotted and the equation relating TAR value to depth was calculated. Statistical analysis was carried out using the standard linear regression model.

Results

In Fig. 1, TAR values were plotted in natural scale, against depth. Points tended to outline an exponential curve and in a semilog representation, with TAR values substituted by log (TAR×1000), the line became straight.

The relationship between paired values was calculated using least square method; it is expressed by the formula:

$$Y = 3.001 - 0.015X$$

where X = half bitrochanteric diameter (in cm) and Y = log (TAR×1000)

The r value of regression coefficient is = 0.79, which is highly significant ($p<10^{-6}$).

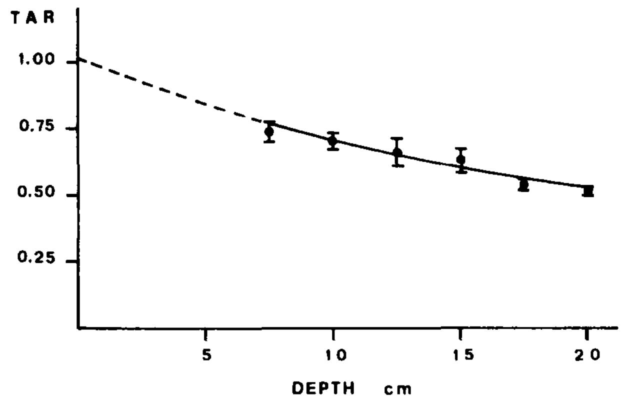

Fig. 1. TAR curve produced, in natural scale. Experimental values between 7.5 and 20 cm tissue depth. Only 2.5 cm intervals, with 95% confidence intervals, are depicted.

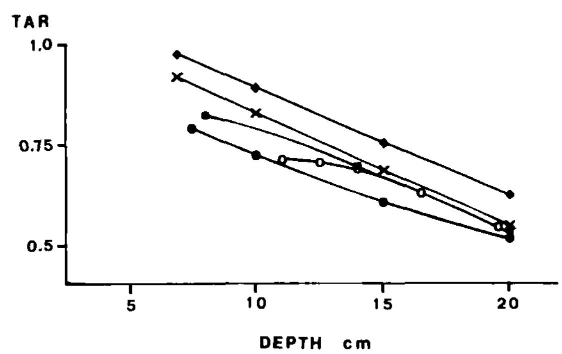

Fig. 2. Comparison between present curve and preexisting TAR data for TBI geometric conditions. ◆: Van Dyk et al., field size 75×75 cm; ×: Van Dyk et al., field size 30×30 cm; ■: Hochhäuser; ○: Johnson; ●: present curve.

Discussion

In Fig. 2, the new TAR curve is compared to preexisting data for similar geometric conditions, i.e.: Van Dyk et al. (4) TAR data for a water phantom, with an equivalent field of 75 cm^2 and of 30 cm^2, 90 cm source-to-surface-distance: the Hochhäuser & Balk (5) data for an Alderson phantom with a 165×30 cm field, 400 cm source–axis distance; and the Johnson (6) curve. The latter is the only curve obtained from 57 'in vivo' readings with TLDs placed in the patients' rectum.

The following points should be considered:

1) TAR values are a measure of absorbed depth dose, relative to 'in air' dose in the same conditions; they are measured, by definition, in conditions of full scatter, in an 'infinite' phantom. Such conditions are never fully realized in clinical practice, but in TBI they are even more difficult to obtain, since in air readings suffer from much more air traversed, and scatter from walls is very different from usual. Furthermore the phantom (or patient) is far from 'infinite', but instead occupies part of the beam, thus full scatter conditions can in no way be obtained.

2) Van Dyk et al. (4) reported the highest TAR values, as they were obtained with the best scatter conditions. As usual, TAR values at a given depth are higher for larger fields. They are better comparable to 'classic' TAR curves (7) than to the TBI setting, where fields are frequently larger, but the absorber is smaller.

3) Hochhäuser & Balk (5) produced TAR data obtained in an anthropomorphic phantom; the curve is lower and flattens as depth of dose readings is reduced; it better fits clinical conditions.

4) Johnson's (6) curve is lower than those previously described. This is probably the result of a less complete scatter contribution within a real human pelvis, than in a homogeneous phantom; moreover, for shorter diameters the curve becomes flatter, perhaps indicating that a shielding effect of pelvic bones and an increasing scatter defect prevail over the width reduction, as was already discussed and experimentally illustrated by Hochhäuser (8).

5) The above considerations tend to show TAR values in TBI as a complex function of unusual, never well evaluated, and not easily computable physical determinants: non-uniformity of traversed materials, irregular absorber's shape, altered beam composition, scattering materials (walls, patient's body, couch, shieldings, . . .), site and means of dose readings. The flattening of Johnson's (6) curve is probably an artifact, and it could lead to clinical errors, since on this basis small patients would receive relatively higher doses compared to larger individuals.

6) The TAR curve presented here has the shape of Van Dyk's curves, but a different slope; they converge at larger depths, where scatter conditions are more similar. Our curve is lower than Johnson's curve, due to lesser scatter contribution at the bifurcation of the thighs, where less tissue surrounds the dosimeter. The availability of pediatric patients gave us values of bitrochanteric diameter less than 22 cm, which is the least value presented in Johnson's curve. We did not observe a curve deflection for lower values; on the contrary the shape is almost exponential (as every 'classic' TAR curve (9)), with an effect, in the portion of the curve considered, contrary to that demonstrated by Johnson. Our data produced a curve which can be applied to the majority of treatment settings, as it is improbable that a patient will have a bitrochanteric width of less than 15 cm or more than 40 cm. In any case, the linear shape of the curve in semilog plot, and the formula representing it, permit us to calculate values below and above the measured values.

We are at present using our formula to precalculate TBI treatment time, and the measured data closely reproduce the calculated values.

Our TAR values can be directly applied by everyone working in the same clinical and physical setting, i.e., same source and treatment geometry.

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