AAPM REPORT NO. 17

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A REPORT OF TASK GROUP 29
RADIATION THERAPY COMMITTEE
AMERICAN ASSOCIATION OF
PHYSICISTS IN MEDICINE

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The Physical Aspects of Total and Half Body Photon Irradiation

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I. Introduction

1.1 General Considerations

In recent years, there has been a revived interest in the use of very large radiation fields for the treatment of a variety of malignant diseases. The ability to provide these very large fields and the corresponding delivery of a specified dose of radiation have been challenging for the medical radiation physicist[33,78,100]. Because of the constraints of radiotherapy apparatus, the techniques associated with total and half body radiotherapy have been as varied as the number of radiation oncologists using them. The treatments are complicated by uncertainties in absolute dosimetry as well as large dose variations across the target volume, making it very difficult to assess clinical efficacy when comparing results from various treatment centers. Furthermore, the actual dose delivered to the patient is often limited by normal tissue tolerance. Ideally, accurate dose response data should be available so the clinician can optimize the therapeutic effect while minimizing normal tissue complications. In principle, this information can be derived by assessing how patients respond to therapy. In practice such information is difficult to obtain. Herring[36] noted that institutions with a large number of patients generally treat with a small dose range while the collecting of patient data from different institutions is confounded by a lack of uniformity of dose prescription, dose delivery and clinical evaluation. Multi-institutional clinical trials are designed to minimize these variations; however, patient selection and adherence to therapy protocols remain ongoing problems. One thing is sure: the techniques and dosimetry of radiation treatments are more controllable than many of the other variables associated with clinical trials. It is, therefore, imperative that the medical physicist should provide an accurate control of the radiation dose delivery such that uncertainties in target and normal tissue doses do not become the limiting factors in evaluating clinical trials nor, for that matter, in individual patient treatments.

This raises the question: with what accuracy must the dose be delivered? The International Commission of Radiation Units and measurements (ICRU)[40] has recommended an overall accuracy in dose delivery of ±5% based on an analysis of dose response data and an evaluation of errors in dose delivery. Recent half body and total body irradiation data[48,97] indicate that a 5% change in lung dose could result in 20% change in the incidence of radiation pneumonitis, a complication which is usually fatal for whole lung irradiation. With such sharp dose response effects, a ±5% accuracy may be insufficient. However, if the prescribed dose is well below the onset of normal tissue toxicity or if the normal tissue dose is limited locally then perhaps the ±5% suggested accuracy can be relaxed. The APARA principle[99] (As Precise As Readily Achievable, technical and biological factors being taken into account) should be applied especially for very large field irradiation. In any case a statement of target or normal tissue dose should always include a statement about the associated uncertainties in dose delivery. Both pieces of information are needed in any clinical evaluation.

1.2 Clinical Indications

Total and half body radiotherapy have been administered for a large variety of clinical situations each with different prescribed dose-fractionation schemes. The following summarizes some of the clinical needs for large field radiotherapy.

a) High dose total body irradiation. High dose total body irradiation (TBI) with megavoltage photon beams is frequently used to destroy the bone marrow and leukemic cells, to immunosuppress the patient prior to receiving a bone marrow transplant[92] (BMT), or both. Aplastic anemia[3,93], and a number of leukemias[92] and lymphomas[57,70], respond to this treatment. Ewing's sarcoma[43], advanced non-Hodgkin's lymphoma[57,70], oat cell carcinoma of the bronchus[57] and lymphosarcoma[7] have also been treated by TBI as an adjuvant. Usually this treatment regimen is combined with a comprehensive chemotherapy program prior to the TBI and bone marrow transplant. Total doses ranging from 300 to 1000 cGy have been given in a single fraction. In recent years, increased usage has been made of doses ranging from 1000 to 1400 cGy given in several fractions per day for several days[20,68]. This wide disparity in dose and fractionation, the technical complexity of TBI techniques, and the variation in chemotherapy programs complicate the interpretation of radiation response to TBI. High degrees of pulmonary toxicity are associated with a number of these treatment regimens and appear to be related to lung dose[48]. There is some evidence that this pulmonary toxicity[3] as well as bone marrow cell killing[17,20] is dose rate dependent. Presently, some institutions use fractionation schemes[68,81] to reduce pulmonary complications while maintaining a high immunosuppressive effect, whereas other institutions are using a lower dose, single treatment[77] for the same purpose.

b) Low dose total body irradiation. Low dose TBI with megavoltage photons giving about 10 to 15 cGy per day for 10 to 15 days is used for treatment of lymphocytic leukemias[47], lymphomas[39] or neuroblastoma[14]. The lower doses reduce the risk of serious complications. However, precise dose response data are not available; hence, a detailed understanding of the associated dosimetry is a prerequisite.

c) Half body irradiation. The last 10 years have demonstrated a dramatic increase in the use of high dose half body irradiation (HBI) for the palliation of widely disseminated metastatic disease[24,25]. This technique has become so successful that it is being used in a number of clinical trials as an adjuvant form of primary therapy for Ewing's sarcoma[44] and bronchogenic carcinoma[67,94]. Lower HBI has also been used for ovarian ablation in metastatic breast cancer[26]. The aim of the half body technique is to give a sufficiently high dose to alleviate the effects of symptomatic disease while at the same time maintaining a sufficiently low dose to minimize complications. The narrow range of therapeutic ratios dictate accurate dosimetry and precise dose delivery.

d) Total lymphoid irradiation. Total lymphoid irradiation (TLI) has been shown to be a powerful immunosuppressive agent and, therefore, has been suggested as an adjunctive therapy for organ transplantation and a number of autoimmune diseases[35] such as rheumatoid arthritis, aplastic anemia, multiple sclerosis and systemic lupus erythematosus.

Total lymphoid irradiation using 3600 cGy in 16 fractions has produced favorable results for patients with rheumatoid arthritis[85]. At this time the precise role of TLI is not clearly understood. Similar to TBI and HBI, accurate dose-response data can only be derived from a precise knowledge of the delivered dose. However, TLI has the additional problem of complicated shielding and very irregularly shaped fields.

While all the above clinical uses of total and half body irradiation are presently in vogue, the dosimetric concerns outlined in this report are relevant for any clinical situation requiring large field radiotherapy.

1.3 Intent of this report

The desire for uniformity of dose delivery and dose prescription for very large field radiotherapy has resulted in a number of conferences and workshops[9,61,63,88,90], attended by physicists, radiobiologists and clinicians. These conferences have consistently indicated a wide disparity in treatment techniques and dose prescription. Standardization of reporting delivered doses was considered essential if biological and clinical parameters are to be understood. However, none of the conferences came to a consensus as to dosimetric procedures.

The intent of this report is:

a) to review methods of producing the very large fields needed for TBI, HBI and other large field procedures,
b) to make recommendations regarding dosimetric measurements that are required prior to initiating such procedures,
c) to consider the practical problems of specifying and delivering a controlled radiation dose for such large fields.

Unfortunately, we must leave unanswered many interesting radiobiological questions about dose, dose rate, fractionation and the possible equivalence between different treatment regimens. Improvement in dosimetry and a standardization of technique will, however, certainly help the medical physicist along with his colleagues in radiation oncology and radiobiology answer some of these questions.

2. Irradiation Methods

2.1 Introduction

Before instituting large field radiotherapy procedures, medical decisions must be made about the total dose to be given to the patient, the dose rate, the desired degree of dose uniformity, the duration of treatment per fraction, the total number of fractions and the total treatment time. At times, some of these parameters may have to be compromised because of the limitations in the available therapy equipment.

Much of the early clinical experience with TBI and HBI procedures was gained at centers with facilities specifically designed for large field irradiations. With the increased use of large field radiotherapy to treat a variety of diseases, many patients are now being treated with conventional units. Current methods of large field irradiation can therefore be broadly divided into three categories:

a) dedicated facilities specifically designed for treatment with large fields.
b) facilities designed for conventional radiotherapy treatments but modified to produce very large fields.
c) facilities designed for conventional treatments but using unconventional geometries to provide the desired field sizes.

One of the first dedicated TBI facilities in North America was described by Heublein[37] in 1932. In a lead lined ward, with four beds at one end and a Coolidge "deep therapy" tube at the other end, four patients could be simultaneously irradiated. The beds were about 5 m and 7 m from the tube, which was operated continuously for 20 hours at 185 kVp, 3 mA, with a 2 mm Cu filter. Exposure rates ranged from 0.68 'R'/h to 1.26 'R'/h and doses were prescribed as a percentage of an erythema dose[37], which was about 750 'R' ('R' represents the unit of exposure as determined at that time).

2.2 Dedicated facilities with multiple sources

Numerous facilities have been designed since 1932 for total body irradiation. Webster[103] reviewed the physical considerations for the design of irradiators that would produce a relatively uniform (±10%) total body dose. He concluded that the following characteristics are desirable: 1) small room size to minimize cost, shielding, and building space; 2) sources distributed above and below the patient; 3) a minimum number of sources to reduce cost and maintenance and 4) simple procedure for control of exposure. After reviewing the design and dosimetry of single, dual, four, six and eight source irradiation facilities, he concluded that a minimum of four radiation sources would be necessary to uniformly irradiate the entire body (Figure la). Jacobs and Pape[42] describe a total body irradiation chamber similar to a "four poster bed", used at the City of Hope Medical Center in the early 1960's. Four rods were housed at each end of the treatment bed with each rod containing two 300 Ci caesium-137 sources separated 2 m from each other. The source rods retracted into the floor to turn the sources "off" and lead filters were used to obtain variable dose rates.

Several multiple source irradiators were constructed at U.S. Government sponsored research laboratories in the 1950's and 1960's mostly for animal studies. The Naval Medical Research Institute16

unit consisted of 60 cobalt-60 slugs that could be positioned around an animal or a patient. Brucer[5] described the original eight source facility at Oak Ridge Institute of Nuclear Studies (ORINS). Subsequently, three separate multiple source irradiators with-exposure rates from 1.5 R/min to 40 R/min containing 5 to 8 cobalt-60 or caesium-137 sources were used at the ORINS in the mid 1960's and early 1970's to study radiation effects in mammals and to assess the radiation risks to astronauts during extended space exploration[2].

2.3 Dedicated facility with dual sources

Dedicated multiple source irradiation rooms obviously are too expensive for most medical facilities. Hence, Sahler[79] developed a dual source cobalt irradiator in 1959. This facility consists of a conventional rotating cobalt unit plus an industrial large field

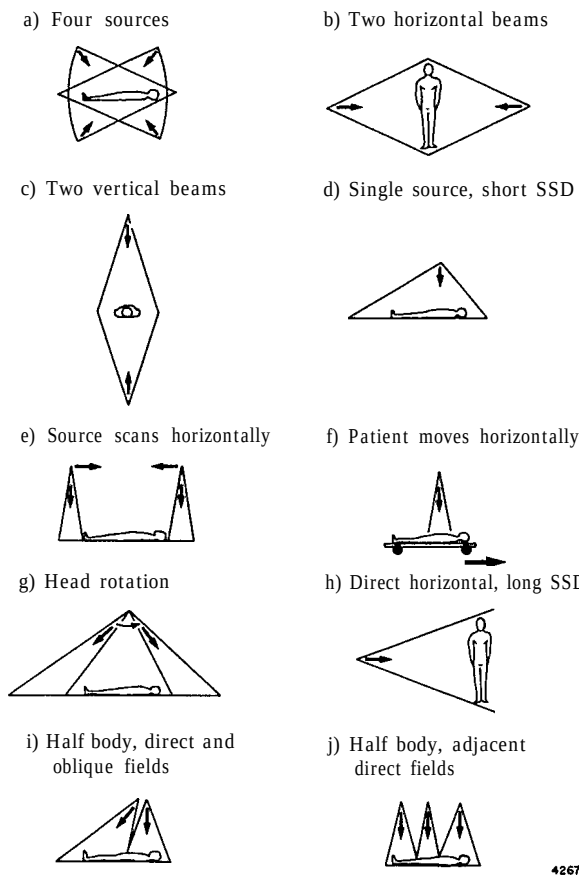

Figure 1. Different methods of total and half body irradiation.

cobalt irradiator. These two units could be arranged to produce a parallel opposed field at about 3 m from the patient for low dose rate TBI (Figure lb). However, the collimators of the conventional cobalt unit had to be removed for TBI in order. to-.--obtain a large field. Surmont et a1.[86] performed TBI in the 1960's at the Institut Gustave Roussy using twin opposed cobalt-60 sources separated by a movable concrete wall. Two track mounted, mobile, parallel opposed cobalt-60 sources, with specially designed collimators, are used by Thomas et a1.[91] for TBI in their bone marrow transplant program at the Fred Hutchinson Cancer Research Center in Seattle, Washington and by the Munich Cooperative Group[53] for BMT (Figure lb). The dosimetry of the dual sources in Seattle has been reviewed by Lam et a1.[55]. The dual, parallel opposed cobalt source method of TBI is appealing because of the reproducibility of the set up, the constant, but low dose rate achievable with the sources, and the relative dose uniformity provided by the technique.

Lutz et al.[58,59] have described a unique dual source large field and TBI facility using ceiling and floor mounted parallel opposed 4 MeV linear accelerators (Figure 1c). Each machine can travel vertically and the source-to-source distances can be varied from 240 cm to 410 cm. The field size is variable with a maximum of 75 cm x 210 cm. A flattening filter was designed to improve beam uniformity and dose rates are variable from 4 to 225 cGy/min[58].

2.4 Dedicated facilities with single sources

Investigators at the Princess Margaret Hospital in Toronto designed and constructed a special single source, large field, cobalt-60 irradiator[56] in 1977 (Figure 1d). Field sizes of 50 x 160 cm^2 can be obtained at 90 cm from the source (i.e. 83 x 265 cm^2 at 150 cm). Special flattening filters were designed to account for the dose variation across the beam due to inverse square effects. In the long axis, the two collimators move independently so that patients receiving half body treatments can be treated with one collimator closed to the central ray. When both the upper and lower half body are to be treated with a six week time interval, then setting the central ray to the umbilicus for both treatments yields a uniform dose distribution in the junction region. With a high activity cobalt-60 source of nearly 10,000 Ci, high dose treatments can be given with short treatment times (approximately 50 cGy/min). A special lead attenuator has been designed and installed near the source to allow for low dose rate treatments if desired.

2.5 Conventional units modified for large field treatments

Large field treatment techniques based on conventional equipment can be categorized as either moving or stationary. Cunningham and Wright11 describe a ceiling track mounted Picker C-3000 cobalt-60 teletherapy unit with a specially designed collimator; the source could scan the length of a patient positioned at about 120 cm SSD (Figure 1e). More recently, Quast[75] described a similar method in which the patient is moved on a mobile couch beneath a fixed cobalt-60 source (Figure 1f). Engler et al.[19] developed an arc TBI technique using a 42 MV betatron, and Pla et a1.[71] modified a column mounted 4 MV linac to sweep over a patient (Figure 1g). A similar technique using an isocentric cobalt unit fitted with automatic arcing facilities and a specially designed curved couch has been described by Mulvey et a1.[66].

Recently, Peters and Herer[69] have described a very simple procedure for removing the collimating system from a widely used cobalt-60 therapy unit. In less than fifteen minutes this unit could be modified to handle large field stationary treatment procedures. The dosimetric data are similar to those obtained on the dedicated single source unit of the Princess Margaret Hospital and the treatment procedures are identical.

2.6 Conventional treatment units

There are numerous reports on the dosimetry of unmodified x ray or cobalt stationary sources at both relatively short treatment distances [1,6,7,23,28,31,32,34,36,49,50.54,63,64,65,68,81,87]. Depending on the maximum field size attainable, patients have been treated laterally, usually in a seated or reclining position, or anteriorly/posteriorly, usually while lying on their side on a stretcher (Figure 1h). Most of these treatments have been at a "low" dose rate, usually less than 10 cGy/min, dictated by the geometry of the treatment set-up, although in some instances dose rates have also been purposely reduced to these low levels for radiobiological reasons. Varying the dose rate has been made easier by the advent of linacs, where outputs can be changed electronically. Most recently, the trend has been to use the higher dose rates available with the linacs combined with multiple fraction treatments as opposed to the single fraction, low dose rate treatment given in the past[1,68,81].

2.7 Multiple fields

Although the application of multiple adjacent fields is another possibility for TB1[8], its use is rare since adjacent fields provide the additional dosimetric problems associated with field junctions as well as the concern about cells circulating through the body and therefore potentially receiving a reduced dose. For HBI, however, when both halves of the body are to be irradiated adjacent fields are used since the treatment of the upper and lower halves are generally separated by a time interval of 4 to 6 weeks. A single oblique field (Figure 1i) or two adjacent fields with the beams pointing vertically down (Figure 1j) can be used to treat the lower half body. A reduced dose variation can be achieved when fields are abutted on the skin surface by relocating the junction for the PA fields compared to the junction for the AP fields for the adjacent lower half body fields.

2.8 Selecting a large field technique

Those wishing to implement large field radiotherapy in a clinic may choose a) to design a "dedicated" unit, b) to develop a special treatment method such as sweeping beam, a moving couch, or modified collimator or, c) simply to use an existing therapy unit within its geometric constraints. Practically, the choice will depend on the equipment available; its workload with conventional treatments and its subsequent availability for large field radiotherapy; the number of patients to be treated and the frequency of their treatments; and, perhaps most important, the resources available for development of a good technique. Because these parameters are unique to each radiotherapy department, it is beyond the scope of this report to state which is the best method of treatment.

However, there are a number of physical parameters that should be considered and optimized for each individual institution. The most

common options relate to a) the energy of radiation, b) treatment distance, c) choice of antero-posterior (AP) treatments, lateral treatments or a combination of these, and d) dose rate. Figure 2 shows the ratio of the peak dose to the midline dose on the central ray as a function of patient thickness for parallel-opposed radiation fields. Data are graphed for 3 energies, cobalt-60, 6 MV x rays and 25 MV x rays, and a field size of 50 x 50 cm^2 at a number of different SSD's. The horizontal shaded region indicates a dose uniformity within 15%. For AP treatments, adult patient diameters usually range between 18 and 26 cm. The shaded region A represents these patient diameters and the 15% dose uniformity. The effects of tissue inhomogeneities and the effects of dose build-up near the surface are not considered. Only cobalt-60 at an SSD of 80 cm falls outside of this dose uniformity for patient diameters greater than 25 cm. Hence, for most large field techniques, AP treatments will provide better than 15% uniformity even for cobalt-60 radiation. The lateral opposed beam procedure can be represented by shaded region B, where lateral patient diameters are assumed to range between 38 and 50 cm. Only 25 MV x rays at a distance of 300 cm will yield a dose uniformity within 15% for a 50 cm diameter patient. Lower energies and shorter treatment distances will result in a greater dose variation. It is clear that the use of cobalt-60 with lateral opposed beams will tend to produce quite large dose variations. Several conclusions can be

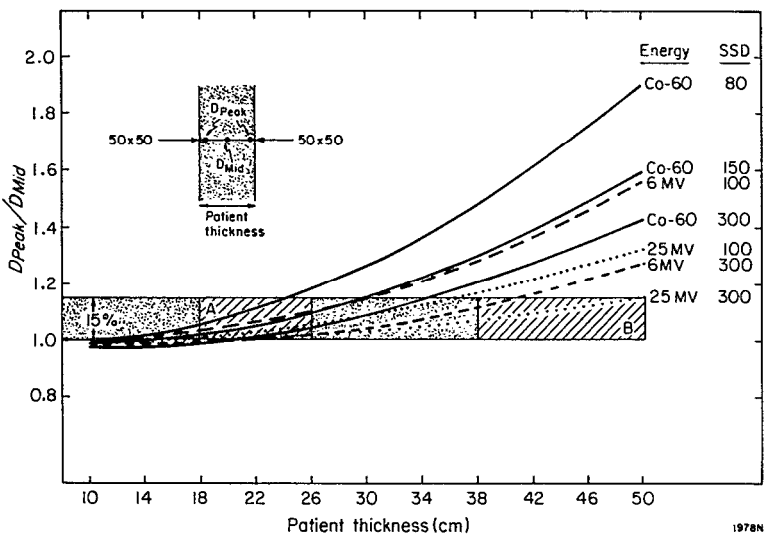

Figure 2: Ratio of peak dose to midplane dose on the central ray versus patient thickness. The horizontal shaded region represents a 15% spread in this ratio. Cross hatched region A represents the typical range of adult patient diameters in the anterior-posterior direction while cross hatched region B represents the range of adult patient diameters in the lateral direction.

drawn from these data:

1) the higher the energy, the lower the dose variation (excluding the the effects of the build-up region and tissue inhomogeneities).
2) the larger the treatment distance, the lower the. dose variation.
3) the larger the patient diameter, the larger the dose variation.
4) AP/PA treatments will yield a variation not larger than 15% for most megavoltage energies and distances.
5) Lateral opposed beams will usually give a greater dose variation compared to AP/PA treatments especially for adult patients. For pediatric cases or higher energy x-ray beams, a ±15% uniformity might be achievable with bilateral fields.

Reducing the allowable dose variation to 10% will place even greater constraints on the choice of energy and the use of lateral opposed fields for adult patients may not yield the desired 10% uniformity. These conclusions have only considered the effect of maximum patient thickness. The variation in patient thickness at different levels in the body will add to this dose variation and will be considered in Section 4.2.1.

If high energy x rays from a linear accelerator are used, some consideration should be given to the effects of the low dose in the build-up region. There are presently no data indicating clinical problems because of this effect, but the dose in the build-up region can be increased by the addition of a beam spoiler[15] such as a plate of plastic near the skin surface. The choice of the material, its thickness and location will be dependent on the dose criteria recommended by the clinicians.

Some clinical protocols require low dose rate treatment at the rate of 5 to 10 cGy/min. For cobalt-60 machines, reduced dose rates can be achieved by placing absorbing materials across the beam and performing careful measurements under the absorber to account for the effects of both attenuation and scatter from the absorber. For linear accelerators the beam operating conditions may have to be adjusted to reduce the beam current or pulse repetition frequency. If conventional dose rates are desired then the dose rate on a linear accelerator may have to be adjusted upward to account for the effects of inverse-square fall-off as a result of the long distance treatments. In either case, dosimetry measurements should be performed under the operating conditions to insure accurate dose delivery.

Radiotherapy rooms are not usually designed to provide the long distances that may be required for large field treatments. Therefore, it is worth emphasizing to designers of radiotherapy departments that at least one treatment room (probably the one with the highest energy therapy machine) should be designed to accommodate large field radiotherapy.

2.9 Rack-up technique

Some clinical procedures, such as BMT, require large field irradiation at a specified time within a comprehensive drug and radiation treatment protocol. Because the large field irradiation cannot be delayed, a back-up method of treatment should be considered either within the same institution or at another near-by radiation therapy department. This is especially true if linacs are used since these tend to have a greater down time compared to cobalt-60 irradiators.

3. Basic phantom dosimetry

3.1 Introduction

Once a decision is made on the method- of the large field treatment procedure for a particular institution, a number of basic dosimetric parameters should be measured using appropriate phantoms. The diversity in the production of large fields used for TBI and HBI techniques means that the dosimetric data required will vary from one institution to another. The following sections discuss the types of measurements that should be performed and the corresponding special concerns. This report attempts to deal with the most general situation where both the field size and distance may vary depending on the specific needs of the patient. Some institutions may require only one set-up geometry. In that case, the number of measurements can be reduced although the general principles discussed below will still apply. For some institutions, the expertise is available to perform reasonable checks at the extended treatment distance to verify that conventional data might be applicable to their new treatment geometry. Once the appropriate checks have been made and a sufficient accuracy can be achieved, these institutions may wish to proceed with their data for conventional treatments but with appropriate corrections for the new geometry.

The general approach recommended here is to use a three step process as outlined in Figure 3.

Step 1. Determine an absolute calibration of the radiation beam using the large field geometry and the largest phantom available. For many institutions, this will correspond to using the same phantom that is used for conventional calibration procedures.

Step 2. Correct this dose such that it represents both: (a) the dose that would be obtained for a phantom that covers the entire beam, and (b) the dose that would be obtained for a deep phantom i.e. full scattering conditions.

Step 3: For patient treatments, corrections should be made for patient dimensions both in terms of the area of the patient intersecting the radiation beam as well as patient thickness. This will allow adjustments to be made for the extremely large variations that are possible when comparing young pediatric cases to adult patients.

The following sections address the special dosimetric concerns in each of the three steps listed above.

3.2 Dosimetry phantoms

As indicated by the report from AAPM Task Group 21[89] (TG21), water is the recommended material for dosimetry phantoms. TG21 does indicate that polystyrene and acrylic plastics can also be used; however, the dose calibration is to be referenced to water. The transfer of dose in plastic to dose in water for X or gamma rays is accomplished by application of the ratio of the average. mass energy-absorption coefficients of water to that of plastic. Although this parameter changes slowly with beam energy, large radiation fields can produce a huge amount of multiple scatter in large phantoms resulting in noticeable changes in these ratios. An example of this

is shown in Figure 4, where photon energy spectra for cobalt-60 as calculated by Bruce and Johns[4] are compared for 10 x 10 cm^2 field and an infinite field size both at a depth of 10 cm in water. Johns and Cunningham[45] have indicated that the average mass energy absorption coefficient ratios of water to medium for cobalt-60 change by about 0.5% when comparing these ratios for primary radiation alone versus primary plus scatter for a 10 x 10 cm^2 field at a depth of 10 cm. In changing from 10 x 10 cm^2 to an infinite field size the ratios change by a further 1.1% and increasing the depth to 20 cm can result in an additional change of 1% (see Table 1)[10]. With uncertainties in the determination of very large field spectra, it is clear that the least error will be produced if water is used as phantom material for large field dosimetry. However, the data of Table 1 can be used if measurements are made in other phantom materials.

For x ray beams from high energy accelerators, the relative changes in these mass energy absorption coefficient ratios are smaller primarily due to: a) slower variation in mass energy absorption coefficients as a function of energy at these higher energies, and b)

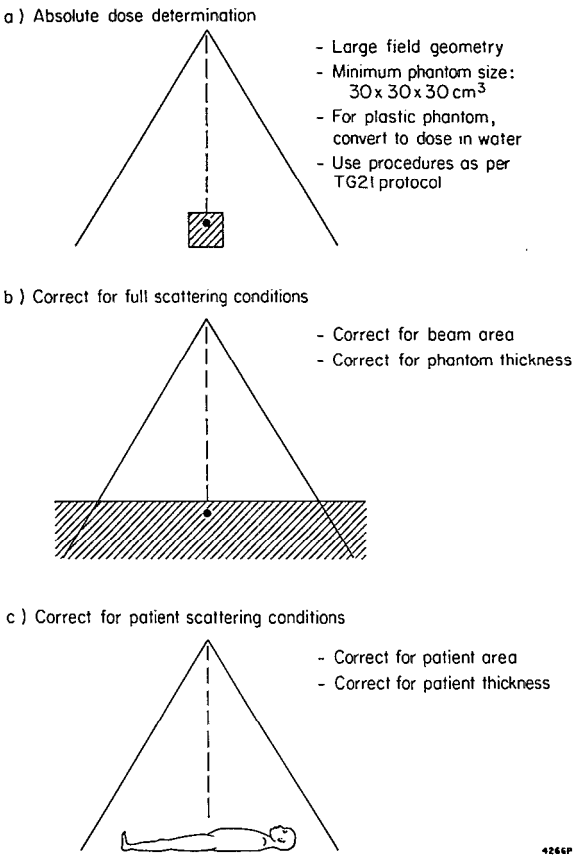

Figure 3: Procedures involved in dose delivery for total and half body irradiation.

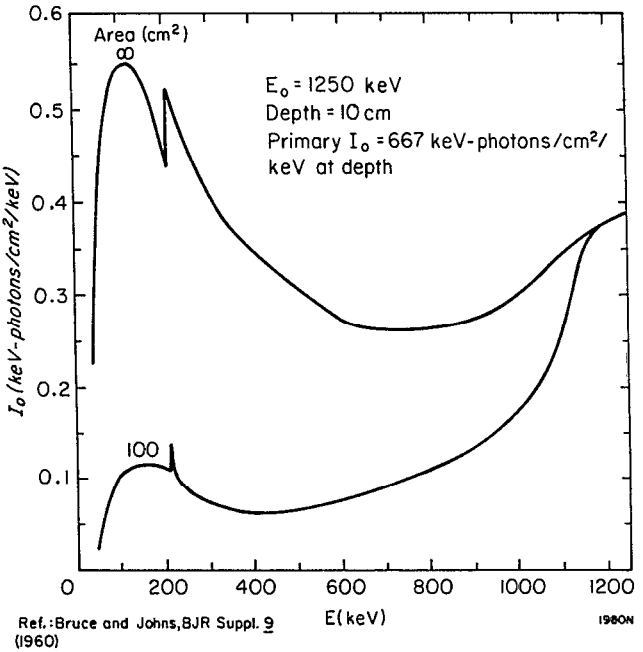

Figure 4: Calculated spectral distributions for cobalt-60 gamma rays in a water phantom at a depth of 10 cm for a 10 x 10 cm^2 field and an infinite field size. Curves are adapted from reference 4. Note that the primary cobalt-60 photons are not shown.

smaller variation in energy spectra at different depths compared with the primary spectrum since the attenuation of primary photons is compensated by the production of scattered photons. Typical examples of mass energy absorption coefficient ratios for the high energy beams are also given in Table 1.

TG21[89] further recommends that the dimensions of a dosimetry phantom should provide a 5 cm margin on all four sides of the largest field size to be employed and a depth sufficient to provide maximum backscatter at the point at which the dose determination is made. For conventional field sizes, 10 cm of phantom material beyond the dosimeter depth is considered to be adequate although it has been shown that this could underestimate the dose by 1-2% for cobalt-60[27,72,95], 6 and 10 Mv x ray[72] fields, with equivalent squares from 30 x 30 cm^2 to 70 x 70 cm^2. For 25 Mv x rays, 10 cm of water behind the measurement point provides full scatter to within 0.5%[27]. For TBI, phantoms which provide a 5 cm margin on all sides of the field and 10 cm beyond the maximum dosimeter depth could be as large as 200 x 50 x 40 cm^3. A water phantom of these dimensions weighs 400 kg. Obviously, a phantom of this size and weight is not practical for routine use.

It is recommended that the minimum phantom size for calibration purposes be 30 x 30 x 30 cm^3 and that whenever possible additional

Table 1

Patios of average mass energy absorption coefficients for various energies, depths and field sizes. The numbers in brackets are for the primary beam spectra. Data are from reference 10.

			$\left(\frac{\bar{\mu}_{en}}{\rho}\right)^{water}_{med}$		
Beam	Depth (cm)	Field Radius (cm)	Carbon	Poly-styrene	Lucite
cobalt-60	0	2.8	1.111	1.032	1.029
	10	5.6	1.113	1.035	1.030
	5	50	1.122	1.044	1.036
	20	50	1.135	1.059	1.045
			(1.111)	(1.032)	(1.029)
6 MV	0	2.8	1.115	1.035	1.031
	10	5.6	1.116	1.036	1.031
	5	50	1.123	1.043	1.036
	20	50	1.132	1.053	1.041
			(1.112)	(1.035)	(1.030)
12 MV	0	2.8	1.122	1.049	1.038
	10	5.6	1.123	1.049	1.038
	20	50	1.128	1.055	1.042
			(1.120)	(1.049)	(1.039)
	0	2.8	1.128	1.059	1.044
18 MV	10	5.6	1.128	1.059	1.044
	20	50	1.132	1.062	1.047
			(1.125)	(1.059)	(1.044)
	0	2.8	1.126	1.058	1.043
26 MV	10	5.6	1.128	1.059	1.044
(thick target)	20	50	1.133	1.064	1.047
			(1.124)	(1.058)	(1.044)
	0	2.8	1.129	1.065	1.047
26 MV	10	5.6	1.131	1.066	1.048
(thin target)	20	50	1.140	1.074	1.053
			(1.129)	(1.067)	(1.049)
	0	2.8	1.141	1.085	1.058
45 MV	10	5.6	1.140	1.085	1.058
	20	50	1.142	1.086	1.059
			(1.137)	(1.085)	(1.059)

Table 2: Multiplicative correction factors to adjust data measured in limited phantom sizes to data representing infinite phantom conditions (based on references 72 and 95).

Equivalent Field size (cm^2)		40 x 40			50 x 50			75 x 75		
Energy		Co-60	6 MV	10 MV	Co-60	6 MV	10 MV	Co-60	6 MV	10 MV
Equivalent Phantom size (cm^2)	Depth (cm)									
30 x 30	0.5	1.008			1.014			1.022		
	1.5	1.009	1.004		1.016	1.012		1.026	1.012	
	2.5	1.010	1.005	1.006	1.019	1.013	1.013	1.030	1.014	1.015
	5.0	1.013	1.006	1.007	1.025	1.017	1.015	1.040	1.020	1.019
	10.0	1.024	1.014	1.011	1.043	1.029	1.021	1.065	1.037	1.024
	20.0	1.054	1.035	1.017	1.089	1.059	1.039	1.127	1.063	1.044
	30.0	1.087			1.145			1.210		
40 x 40	0.5	1.000			1.007			1.014		
	1.5	1.000	1.000		1.008	1.008		1.017	1.008	
	2.5	1.000	1.000	1.000	1.009	1.009	1.007	1.020	1.010	1.010
	5.0	1.000	1.000	1.000	1.012	1.011	1.008	1.027	1.014	1.011
	10.0	1.000	1.000	1.000	1.019	1.014	1.009	1.040	1.022	1.012
	20.0	1.000	1.000	1.000	1.032	1.023	1.022	1.069	1.028	1.026
	30.0	1.000	1.000	1.000	1.054			1.113		
50 x 50	0.5	1.000			1.000			1.008		
	1.5	1.000	1.000		1.000	1.000		1.010	1.000	
	2.5	1.000	1.000	1.000	1.000	1.000	1.000	1.011	1.001	1.003
	5.0	1.000	1.000	1.000	1.000	1.000	1.000	1.015	1.003	1.003
	10.0	1.000	1.000	1.000	1.000	1.000	1.000	1.021	1.009	1.003
	20.0	1.000	1.000	1.000	1.000	1.000	1.000	1.035	1.005	1.005
	30.0	1.000	1.000	1.000	1.000	1.000	1.000	1.056		

Table 3: Multiplicative correction factors to adjust data measured in limited phantom thicknesses to data representing infinitely thick phantom conditions.

Equivalent field size or phantom size* (cm^2)	30 x 30			40 x 40			50 x 50			75 x 75		
Energy	Co-60	6 MV	10 MV	Co-60	6 MV	10 MV	Co-60	6 MV	10 MV	Co-60	6 MV	10 MV
Distance from exit surface (cm)												
0.5	1.070	1.045	1.024	1.077	1.053	1.029	1.080	1.059	1.034	1.088	1.078	1.046
1.0	1.058	1.031	1.020	1.067	1.040	1.026	1.074	1.048	1.030	1.082	1.070	1.042
1.5	1.037	1.019	1.013	1.047	1.028	1.016	1.055	1.036	1.020	1.065	1.057	1.031
5	1.020	1.008	1.005	1.028	1.012	1.007	1.036	1.018	1.011	1.048	1.025	1.014
10	1.006	1.001	1.000	1.010	1.002	1.001	1.013	1.006	1.004	1.017	1.010	1.007
20	1.000	1.000	1.000	1.000	1.000	1.000	1.001	1.000	1.000	1.002	1.000	1.000

* Equivalent field size or phantom size whichever is smaller.

phantom material be placed around this minimum phantom size to achieve full scattering conditions. The determination of dose in phantoms of limited dimensions will have to be corrected to obtain data for full scattering conditions. This can be done by using the multiplicative correction factors given in Tables 2 and 3 to adjust data measured in limited phantom sizes to data representing full phantom conditions. For example, if measurements are performed in a 30 x 30 cm^2 phantom (effectively infinitely thick) but the field size is 50 x 50 cm^2 then the measurements should be multiplied by the factors listed in Table 2 under a field size of 50 x 50 cm^2 to yield data that are representative of a full phantom covered by the total beam. For cobalt-60 at a depth of 10 cm in water, this represents a factor of 1.043. These data were derived from measurements by Van Dyk et al.[95] for cobalt-60 and by Podgorsak et al.[72] for cobalt-60 and higher energies. For cobalt-60 the Van Dyk and Podgorsak data agreed within 1% except for the larger depths and larger field sizes (i.e. depth = 20 cm and field size = 40 x 40 to 75 x 75 cm^2) where the results agreed within 3%.

The data by Podgorsak et al.[72] also show that using the area over perimeter rule for equivalent square determination provides a good approximation for converting long rectangular fields (or long rectangular phantoms or patients) to their equivalent square counterparts, i.e.

$$\text{Side of equivalent square} = \frac{2\,ab}{a + b} \qquad (1)$$

where a and b are the lengths of the long and short sides of the rectangular field or phantom, whichever is smaller in area.

Table 3 shows correction factors to make adjustments for measurements in phantoms that are not infinitely deep. For example, if for cobalt-60 radiation, measurements are made in a phantom that is 25 cm thick and the field size is 30 x 30 cm^2 at a depth of measurement of 20 cm, then the measurement should be multiplied by 1.020 to yield a result equivalent to an infinitely thick phantom. The data in this table are derived from Van Dyk et al.[95] and Podgorsak et al.[72] for cobalt-60 and from Podgorsak et al.[72] for 6 and 10 MV x rays.

3.3 Dosimeters

As suggested by TG21[89], the primary method of dosimetry considered will be the use of an ionization chamber with a calibration factor directly traceable to a national standards laboratory. Because of the large fields under consideration, long lengths of cables may be exposed to ionizing radiation. Hence, special attention should be given to ensure that the stem and cable effects are low. Radiation induced cable currents are known to be directly proportional to the length of cable irradiated[84]. Furthermore, Rawlinson[76] has shown that cables irradiated in high energy x ray beams (e.g. 25 MV) demonstrate a net removal of charge due to dose build-up within the cable. This radiation induced current was found to be approximately 1 to 2% of the ionization current for a 1 cc ionization chamber when 1 meter of cable was irradiated. Ionization chambers with smaller volumes (e.g. 0.1 cm^3) will demonstrate proportionately larger cable effects. This effect was found to increase with lower surface dose. The radiation induced cable current could be reduced by a factor of 20 by placing full build-up material over the cable. The irradiation of cables in

cobalt-60 beams demonstrated much smaller effects. To minimize these cable effects, the following precautions should be taken[76]:

1) The cable lengths irradiated should be kept minimal.
2) The cable should be covered with build-up material to ensure full electronic equilibrium within the cable.
3) Very small volume ionization chambers should be avoided.

Large radiation fields in a phantom tend to produce a large, low energy component in the photon spectrum for cobalt-60 as shown in Figure 4. Therefore, it is important that the chamber response is quite energy independent. A sample calculation has shown that an ionization chamber with calibration factors varying between 0.980 R/scale division for cobalt-60 gamma rays to 0.900 R/scale division for a beam with an HVL of 0.5 mm of copper (i.e., equlvalent monoenergetic energy of 60 keV) should, in fact, have a calibration factor of 0.970 R/scale division for the infinite field spectrum of Figure 4. This was determined from the following weighted averaging procedure:

$$F_{eff} = \frac{\sum_i F_i N_i E_i}{\sum N_i E_i} \qquad (2)$$

where F_{eff} = effective calibration factor for large field spectrum.
F_i = calibration factor for energy E_i.
N_i = number of photons for energy E_i.
E_i = energy of energy interval i.

Clearly, ionization chambers with a larger variation in energy response will show a greater than 1% deviation when comparing “in air” cobalt-60 measurements to large field phantom calibrations whereas chambers with a reduced variation in energy response will show a lower than 1% deviation.

For surface dose measurements and for relative measurements in the build-up region, a thin parallel plate ionization chamber is preferred since the volume averaging over rapidly varying dose gradients will be reduced. Thermoluminescent dosimeters (TLD) can also be used for such measurements.

3.4 Dose calibration

The dose calibration should be performed using the principles and methodology of AAPM TG21[89]. The following discussion outlines some of the special concerns for the large field calibration procedures.

The calibration is best made under geometric and phantom conditions that most nearly represent the actual treatment geometry. The depth of calibration should be as specified by Table X1 of the TG21 protocol[89]. The field size should correspond either to the field size that will be used for all treatments or to some reference field which is representative of large field treatments. Further changes in field size must then be related to this reference field size.

Recent data by Cunningham et al.[10] have shown that the average mass energy absorption coefficient ratios are dependent on the photon spectrum at a depth in a medium. This variation becomes more pronounced with large field sizes and lower energies such as cobalt-60 gamma rays. Table 1 shows data taken directly from Cunningham et al.[10] and should be used in conjunction with large field absorbed dose calibrations as recommended by the AAPM TG21 protocol.

Large field treatment techniques are often given at long distances from the source compared to conventional procedures. As a result, the treatments are usually performed close to treatment room walls or near the floor resulting in a “dose rate to a small mass Of tissue” which exceeds the values calculated from the inverse square law because of scatter contribution from the walls or the floor to the measuring chamber[95]. This is especially true for cobalt-60 units and lower energy linacs. It is therefore recommended that "in air" calibrations not be performed.

When using linear accelerators and low dose rates, it is important to check the constancy of the monitor ionization chamber since the machine operating conditions can change over long irradiation times.

3.5 Central ray data

As indicated in the previous section, in-air measurements tend to be confounded by scatter from the walls and floor of the treatment room. For this reason it is recommended that central ray data requiring in-air measurements such as tissue-air ratios (TAR) should be avoided. Hence, only quantities, such as percentage depth doses, tissue-phantom ratios or tissue-maximum ratios, based on phantom measurements in the selected geometry should be used. Some dose ratio parameters, such as tissue-air ratios, tissue-maximum ratios and tissue-phantom ratios, are normally considered to be distance independent. However, there is evidence[1] that extreme deviations from conventional treatment distances results in distance-dependent changes in these quantities. Furthermore, the conversion of percentage depth dose data from one distance to another using the Mayneord factor may also be in error by 2 to 6%[54]. Therefore, central ray measurements should be performed for the large field treatment geometry.

A cylindrical chamber is normally used in a water phantom beyond the depth of maximum dose for measurements along the central ray. In the build-up region, parallel plate chambers should be used. However, some parallel plate chambers are not waterproof; hence, measurements may have to be made in plastic phantoms. The depth of measurement in the plastic phantom will have to be scaled to derive an equivalent depth in water using the procedure recommended by TG21. The dose in the build-up region is strongly dependent on the treatment geometry (field size, SSD) and any intervening attenuating materials; hence, measurements should be made under these conditions.

3.6 Inverse square test

Inverse square law measurements are generally performed in air and are, therefore, affected by scatter from the collimators, the floor and the walls[95]. The application of the variation in output as a function of distance measured in air, to phantom geometries, unless appropriately checked, should be avoided. Phantom geometries will tend to remove some of the radiation scattered from the floor and walls. In section 3.4, it was recommended that the dose calibration be performed at a distance representative of the treatment geometry. The inverse square law need only be tested for the variations from this calibration geometry as might be encountered in the clinical situation. If there are sizeable differences (i.e. >2%) from the inverse square calculation, then dose calibrations should be performed

at a number of appropriate treatment distances (i.e. such that inverse square deviations will always be less than 2%).

3.7 Output factors

Output factors measured in air may be confounded by the problems discussed in the previous sections. If only one treatment geometry is used then the dose calibration is needed only for this geometry and relative output factors are not needed. If a range of field sizes are to be used, then instead of measuring output factors in air, it is recommended that dose calibrations be determined in phantom for a number of field sizes. The output for intermediate geometries can be determined by interpolation.

It should be noted that if beam attenuation devices, such as shielding trays, are to be used routinely then the output as a function of field size should be made with these devices in place. As indicated in section 3.9, beam attenuation materials produce scattered radiation which is dependent on the geometrical arrangement including the field dimensions. Output factors have been shown to change by 13% depending on intervening filter materia1[78,100].

3.8 Beam profiles

In addition to measuring percentage depth doses, tissue-phantom ratios or tissue-maximum ratios along the central ray, the same quantity should also be measured along several rays which intercept the long and short field axes and are parallel to the central ray. The measurements should be normalized to the central ray normalization point. These data give dose profiles at any desired depth in the phantom along the two beam axes. When linear accelerators are used with the collimator rotated such that the patient lies along the field diagonal, there may be a large dose decrease towards the field corners since the beam flattening filters usually have circular symmetry and are often designed to flatten the field along the two principal planes but not along the diagonals[65,87]. For such situations, or for specially designed large field irradiators, it may be necessary to design special filters to achieve an adequate flatness[1,56]. It should be recognized that for linear accelerators, the energy spectrum of the photon beam may vary as a function of distance from the central ray. Hence, the dose profiles measured at shallow depths will not necessarily have the same shapes as those at larger depths.

Ideally, dose profiles would be measured at various depths in a full water phantom. Practically this may be very difficult and plastic phantoms of limited dimensions may have to be used. In this case, the ionization chamber should be located near the center of the plastic phantom and the total phantom should be moved across the beam. Although this technique will yield some indication of the dose variation across the radiation beam, differences as large as 5% have been noted when comparing an “in air” profile to a full phantom profile especially towards the beam edges[75]. A 30 x 30 x 30 cm^3 phantom should provide beam profiles within 3% of full phantom measurements[52].

3.9 Attenuation data

Attenuating materials may be used for a variety of reasons in large field treatments: e.g. shields, compensators, and attenuators

for fractional dose reduction to specified organs. Measurements and calculation of attenuation coefficients or half-value layers (HVL) are generally performed under narrow beam conditions. However, for broad

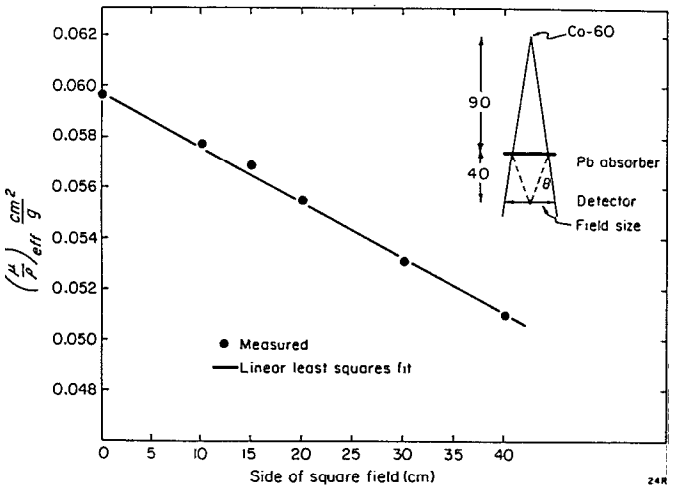

Figure 5: Attenuation coefficients versus side of square field for lead in a cobalt-60 beam. Source-detector distance = 130 cm. Source-absorber distance = 90 cm. Field size is defined at 130 cm. Data is adapted from Reference 102.

beams, measurements and calculations by Van Dyk[102] indicate that attenuation coefficients determined under typical clinical geometries can vary from narrow beam data by 16% for energies between cobalt-60 gamma rays and 25 MV x rays. Sample data are shown in Figure 5 for cobalt-60 gamma rays along with the geometric parameters. As a first approximation, these data indicate that broad beam attenuation coefficients vary linearly with the side of square field. Hence, to determine broad beam coefficients, attenuation curves for three different field sizes under treatment conditions are sufficient to produce a range of coefficients. These measurements are necessary especially if clinical situations require much attenuating material in a large portion of the field. Note that the broad beam attenuation coefficient to be used will be dependent on the maximum scattering angle, Q, as shown in the inset of Figure 5. The maximum scattering angle will be determined from the absorber size if the absorber is smaller than the radiation field, or the size of the radiation field interacting with the absorber if the absorber covers the total beam.

For linear accelerators there is an additional problem that the attenuation coefficient changes as the point of interest moves away from the central ray due to a change in the primary beam photon spectrum[60]. This may require attenuation coefficient measurements as described above to be performed at various distances from the central ray. This is particularly important if attenuating materials will be used clinically to reduce the dose to organs located away from the central ray.

4. Patient dosimetry

4.1 Prescription of dose

The problem of dose prescription when delivering TBI or HBI doses is closely related. to the problem of dose description. For most conventional radiation therapy treatment situations, the homogeneity of the resulting dose distribution is sufficient to use a single number to describe the dose[41]. This shorthand representation of the dose is acceptable so long as it is adequately defined for clear interpretation; e.g. minimum target absorbed dose, mean target absorbed dose, isocenter dose, etc. An example is the use of mantle fields to treat Hodgkin's disease where the use of a single number to prescribe or describe the dose is common practice because the treatment technique is relatively standard (the patient positioned supine and prone for extended distance AP/PA fields). Since the treatment conditions are well understood, the dose distribution from one institution to the next is similar and predictable. By comparison, there is no standard treatment technique for TBI, HBI or other large field treatments. As a result, significant differences in the dose distributions exist with different treatment methods. Two institutions can prescribe the same dose to some selected point; but, if different treatment techniques are used, the dose to other points can vary considerably. Figure 6 shows the difference in the dose for AP/PA total body fields as compared to bilateral fields for the same point dose prescription for 6 MV x rays on a humanoid phantom*. The neck region receives a 35% higher dose depending on the treatment technique. Clearly, TBI and HBI need more comprehensive dose prescription and reporting procedures.

Various techniques for prescribing the dose for TBI have been reported[29,51]. One method uses a calculation of the integral dose to determine the "average" dose[9]. Besides the obvious disadvantage of having to perform a difficult and tedious dose calculation, this approach does not identify high and low dose regions. Another method suggests the use of a "limited average" including areas of high and low dose. This method suffers from the problem that these differences can cancel and leave the average unchanged. A third method prescribes a minimum tumour dose at the midpoint of the maximum separation. Although this gives a minimum dose for all patients, it does not tell us where this minimum dose occurs. A fourth approach uses a single point prescription, but also specifies limits for the highest and lowest dose acceptable for any point in the body. In addition, dose limits are set for certain specific tissues such as the lungs. Using this technique, a typical prescription might read:

> The dose to the midpoint at the level of the umbilicus is 800 cGy. All points in the body must fall within the limits 840 cGy and 720 cGy (+5% and -10%). The dose to more than half the lung volume must not exceed 800 cGy. The dose rate at the prescribed point must not exceed 10 cGy/min.

This example of a TBI prescription uses the midpoint at the level of the umbilicus as a convenient prescription position. For patients treated with the legs slightly bent, the central ray of the beam can

* Alderson-Rando

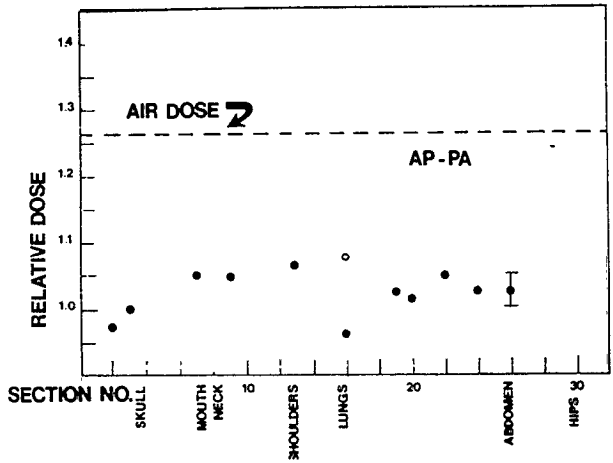

Figure 6(a): Relative dose for AP/PA total body fields using 6 MV x rays on a humanoid phantom. The closed circles represent the dose measured at the center of the phantom section. The open circles show the maximum dose for that particular section. Normalization is at the level of the umbilicus. The "air dose" is the highest dose possible as measured at the normalization point with just the thickness of material needed for electron equilibrium.

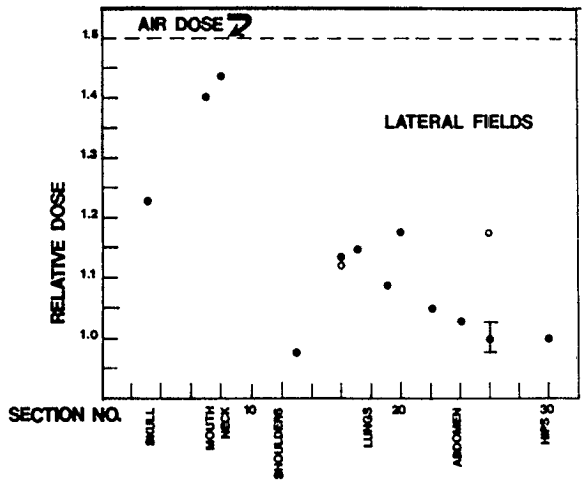

Figure 6(b): Same as Figure 6(a) except that bilateral 6 MV fields are used. Notice that compensation must be used for the head and neck region in order to bring the inhomogeneity within approximately 15%.

be positioned at this location to provide symmetric coverage of the head and feet. Another advantage for using this level for dose prescription is that this region of the body does not exhibit large contour variations and does not contain low density tissues. Thus, the dose calculation is straightforward. The dose rate should be specifically quoted at the prescription point since for some techniques the maximum dose rate could be higher by as much as a factor of 2. If moving beams are used to cover the target volume, then both an instantaneous and average dose rate should be quoted.

This prescription allows flexibility in the adopting of a treatment geometry. Institutions performing TBI or HBI may use different field arrangements and set up geometries as long as dose limits included in the prescription are followed. Bose distribution modifiers, such as bolus or compensators, may be required to meet the prescription. Figure 6b shows that the bilateral field arrangement would exceed the limits of the prescription. The doses in the region of the head, neck and mouth are excessive. On the other hand, the AP/PA technique seems satisfactory. It is also possible to apply compensators (see section 4.3.2) to reduce the dose to the head, neck, and mouth when using the lateral field technique. Either a change to AP/PA fields or addition of compensators would sufficiently modify the dose so that the prescription can be met.

The above discussion illustrates the importance of developing and adhering to a comprehensive dose prescription for TBI and other large field treatments. This is particularly important when total dose and dose fractionation are being modified in an attempt to cure the disease while avoiding radiation induced complications.

4.2 Effects of contour variation and finite patient size

Basic central ray data are usually measured under full scattering conditions. If, however, measurements are made in phantoms which are smaller than the radiation beam then corrections will have to be made to provide full scattering conditions as per Section 3.2. In general, patients do not provide full scatter and adjustments may have to be made for their finite dimensions.

The effects of finite patient size can essentially be divided into 3 components. In the first instance the variation in patient contour on the incident side of the radiation beam will affect the primary and scatter dose conditions at a depth. The second effect considers the lack of lateral scatter in patients that are effectively smaller than the overall dimensions of the radiation beam. The third effect deals with finite thickness of the patient on the exit side of the beam. These effects will now be consider& in greater detail.

4.2.1 Incident contour variation

The effect of contour variation occurring on the incident side of the radiation beam has been a standard dosimetry problem since the early days of radiotherapy. Contour variations cause changes in both the primary and scatter dose at a specified point within the patient. This effect is easily handled by contour correction procedures available in most commercial treatment planning computer programs. ICRU Report 24[40] has considered a number of procedures for performing relatively simple manual calculations. The ratio of tissue-air ratio

method (or the ratio of tissue-phantom ratio method) is considered the method of choice as long as large field data are available since it does consider the effect of field size and depth and it is effectively independent of SSD. The effective SSD method is similar in that it allows for beam dimensions and depth but its actual application is somewhat more cumbersome. The isodose shift method is not recommended since the shift factors are only approximate and do not consider the effects of field size, depth nor distance.

4.2.2 Lack of lateral scatter

Few authors have considered the effect in the dose distribution of field sizes that are larger than the patient. The data of Section 3.2 (specifically Table 2) gives an indication of the effects of finite phantoms. Faw and Glenn[21], for TBI dosimetry, performed measurements in phantoms with cross-sectional areas ranging between 10 x 10 and 122 x 122 cm^2 and field sizes between 10 x 10 and 135 x 135 cm^2. Their TAR data for limited phantom sizes and cobalt-60 radiation are compared in Table 4 to the data of Van Dyk et al.[95] using field sizes equivalent to these phantom sizes. The mean difference between the two sets of data is about 1% with the maximum difference being 4%. The conclusion by Faw and Glenn[21] was that "the relative dose, i.e. percent depth dose or tissue-air ratio, at a specific point, is a function of the field size or the phantom size whichever is smaller". This conclusion is confirmed by the data of Table 4.

Table 4

Comparison of cobalt-60 tissue-air ratios for limited phantoms irradiated in a large field[a] and for limited field sizes in a large phantom[b]. (The latter are shown in brackets).

	Phantom (field) size				
Depth	10 x 10	20 x 20	30 x 30	50 x 50	122 x 122
0.6	1.05 (1.03)	1.07 (1.05)	1.07 (1.07)	1.07 (1.08)	1.07 (1.09)*
5.0	0.91 (0.90)	0.97 (0.95)	1.00 (0.98)	1.02 (1.00)	1.02 (1.02)*
10.0	0.71 (0.71)	0.80 (0.80)	0.85 (0.83)	0.89 (0.87)	0.89 (0.89)*
15.0	0.54 (0.55)	0.64 (0.64)	0.69 (0.69)	0.76 (0.73)	0.77 (0.75)*

a. Faw and Glen (Ref. 21)
b. Van Dyk et al (Ref. 95)
* extrapolated

Another approach to this problem is to use the conventional methodology of irregular field calculations. The outer dimensions of the patient, from a beams eye point of view, can be used to represent the outside borders of the radiation beam. These dimensions can then

be entered into an irregular field calculation program to provide percentage depth dose and absolute dose rates at a series of specified points. If computerized calculations are performed, extreme care should be taken to verify that the basic radiation data (such as TAR's, TPR' percentage depth doses) stored in the computer to perform these calculations are accurate for large field sizes. Linear extrapolation from small to large fields can result in large errors in dose determination. Furthermore, the method of calculating the dose near the beam edge should be scrutinized to ensure that the points of dose calculation are not affected by penumbral effects in the calculation program.

This methodology of irregular field dose calculations has been simplified for manual calculation purposes by Quast[75]. He divides the change of scatter dose as a function of field size into steps of equal relative dose increase. Each resulting field size step is called a "beam zone" and corresponds to an increase in phantom size delivering the same scatter dose increase to a central point. By placing a transparency of these beam zones over a patient topogram, all zones covering the patient can be counted and the effective scatter can be determined.

4.2.3 Lack of backscatter

The dose at a point in a patient is determined by the number of primary and scattered photons interacting in a small volume (i.e. within the electron range) about the measurement point. When measurements are made for basic central ray data, the phantom sizes are such that the maximum multiple photon scatter is achieved. By reducing the thickness of the phantom, the multiple photon scatter component will be reduced. The degree to which the dose is reduced is dependent on the distance the measurement point is from the exit surface, the field size and the energy of radiation. Measurements for cobalt-60 indicate that, for a 30 x 30 cm^2 field, full backscatter is achieved with 30 cm of tissue equivalent material behind the point of measurement[27,72,95]. For large cobalt-60 fields, and at a distance of 0.5 cm from the exit surface, the dose could be reduced by as much as 9% due to this lack of total multiple photon scatter[72,85] (see Table 3).

At the exit surface, there is the additional concern of lack of electronic equilibrium. The dose at 0.002 cm from the exit surface has been shown to be 82-84% of the full scatter dose for a 30 x 30 cm^2 cobalt-60 field and the corresponding value for 25 MV x rays is 90%[27]. For cobalt-60 the dose to the skin increases rapidly as backscatter material up to a thickness of 0.5 cm is added. This rise is mainly due to additional backscattered electrons. Further addition of backscatter material increases the contribution of scattered photons until full multiple scattering is achieved. The reduction of backscattered photons at 0.5 cm from the exit surface for cobalt-60 radiation[72,95] and 6 and 10 MV x rays corresponds approximately to the peak scatter factor. For AP treatments, the couch will result in some backscatter such that electron equilibrium will be retained. Whether full photon scatter will be retained is dependent on the couch material and its thickness.

This possible lack of full scatter is especially important for *in vivo* dosimetry (see Section 4.7). If measurements are made at

entrance and exit surfaces to establish midplane doses using full phantom data, corrections will have to be made in the exit surface measurements for the lack of full scatter at this position.

4.3 Methods of compensating for contour variation-

4.3.1 Tissue-equivalent bolus

The simplest method to compensate for tissue curvature is to use tissue-equivalent bolus material placed directly on the skin. Rice flour and sodium bicarbonate is often used in small bags and placed on the patient as needed. However, the density of this material is not quite tissue-equivalent (lower by about 13% based on CT scan data) and the control of thickness is very difficult. Alternatively, a semi-flexible material such as Superflab* provides excellent tissue-equivalence and, since it is produced in predetermined slab thicknesses allows for good control over thickness.

The use of bolus is easy and useful only if the loss of skin sparing is not of concern. Otherwise missing tissue compensators should be used.

4.3.2 Missing tissue compensators

The use of missing tissue compensators for large field radiotherapy is more difficult compared with conventional therapy fields for two reasons. First, the long treatment distance needed to obtain large field sizes require the mounting of the absorbing material on the face of the treatment head at a considerable distance from the patient, or placing it at some intermediate position distant from the treatment unit. Neither approach is suited to easy indexing of the compensator relative to reference marks on the patient's skin. Second, since immobilization of TBI or HBI patients is not a common practice, patient movement can be a problem. For these reasons compensation for contour variations can be handled. by applying a simple one-dimensional compensator constructed of lead or copper strips[28,50]. Figure 7 illustrates how this compensation technique might be used. Two simple one dimensional compensators of two steps each are shown. One is positioned for the head and neck, and the other for the legs and feet. The advantage of using this type of compensator is that the positioning is critical for only one border if patient posture is rigidly controlled. For the compensator arrangement shown in the Figure, only the edge position where the shoulder projects out from the neck is important. Since the two pieces of head compensator are rigidly attached, aligning the one edge automatically brings the other to the correct location. Careful positioning of the leg compensator is less of a problem as the thickness changes much more gradually in going from the hips to the feet.

Figure 6b demonstrates the importance of accurately placing the compensator edge at the patient's shoulder. If the absorber is shifted superiorly, an overdose to the neck of approximately 40% can occur. On the other hand, if the compensator is moved toward the feet, it will shield the shoulders which are the widest part of the body and already produce the lowest dose. The contour change from the neck to the shoulder is very rapid and positioning must be maintained

* Mick Radio-Nuclear Instruments, Inc., Bronx, N.Y.

within a tolerance of l-2 cm. In practice, this is not a simple matter.

Various investigators28150 have described techniques for calculating the thickness for missing tissue compensators for TBI. One simple method involves a comparison of the tissue-phantom ratio (TPR) for different parts of the body. For example, the TPR for the head is compared to the TPR for the trunk to find the amount of attenuation needed to bring the midline dose in the head into agreement with the midline dose in the abdomen.

The decrease in the photon intensity necessary to obtain homogeneity for the two sections is approximated by using the ratio of tissue-phantom ratios (TPR) as follows:

$$\frac{I}{I_o} = \frac{TPR_T(A_T, d_T)}{TPR_H(A_H, d_H)} \cdot C_{profile} \qquad (3)$$

where I_o and I are the photon intensities before and after adding the compensator, $TPR_T(A_T, d_T)$ and $TPR_H(A_H, d_H)$ are the tissue-phantom ratios for the trunk and head having corresponding areas AT and AR and

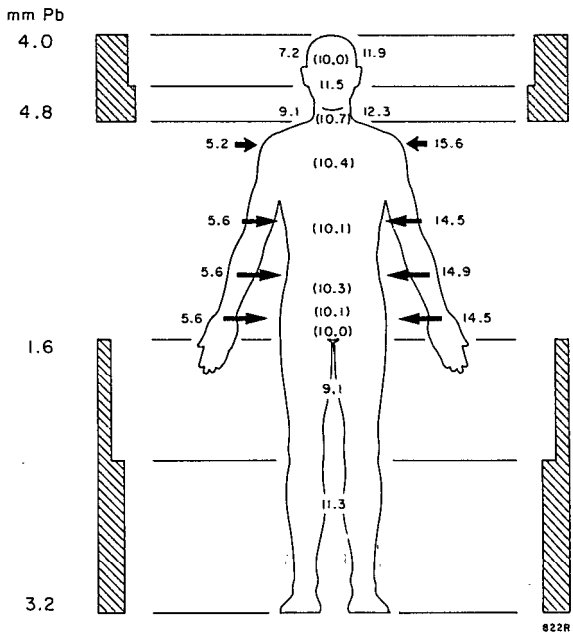

Figure 7: Simple one-dimensional compensator used for lateral field irradiation technique. The compensator corrects for tissue variations along one line only. The numbers shown in this figure are direct dose measurements. The numbers in parenthesis are calculated from the entrance and exit surface measurements.

midline depths d_T and d_H respectively. The correction factor, $C_{profile}$, accounts for the variation in the beam intensity as function of the distance from the central ray. (The central ray is positioned at the level of the- umbilicus for TBI). The compensator thickness is calculated using:

$$\frac{I}{I_o} = e^{-\mu_B t} \qquad (4)$$

where μ_B is the broad beam linear attenuation coefficient specific to this geometry (see section 3.9) and t is the thickness of the absorber needed[102].

4.4 Inhomogeneities

Basic dosimetric measurements are generally performed in water phantoms. As a first approximation, patient calculations are often performed assuming that patient tissues are water equivalent. For muscle and fatty tissues this is a good approximation; however, for bone and lung tissue this could result in very large dose errors.

Practical treatment planning dose calculation programs allow for corrections to a dose distribution by applying an inhomogeneity correction factor to the water-like calculations. ICRU[40] has described a number of such inhomogeneity correction procedures. However, most of these procedures were developed and tested for conventional field sizes. Data by Van Dyk et al.[96] indicate that many such procedures produce inaccurate results (by as much as ±12% in the middle of lung) when the field sizes are extended beyond 30 x 30 cm^2 These data are summarized in Table 5. A more recent review of tissue inhomogeneity corrections for photon beams has been produced by Cunningham[13]. The more sophisticated calculation algorithms such as the "equivalent TAR method" provide good accuracy to better than 3% if CT scan information is used as basic patient data even for very large field sizes[96].

Several factors should be considered in large field inhomogeneity corrections. First, if the correction algorithm is not field size dependent, then it is bound to be in error for large field sizes. Second, any tissue-air ratio method taken to the power of the density tends to underestimate the correction factor for large field sizes[96,104]. A detailed review by Wong[104] of the modified Batho power law method gives a descriptive analysis of the reasons why it breaks down. More recently, El-Khatib et al.[18] have shown that TPR's or TMR's taken to a power of this density provide much better results than using TAR's. Third, methods such as the ratio of TAR's or effective SSD provide reasonably good results assuming that TAR or PDD data are available for the large fields in use. Finally, the more sophisticated equivalent TAR method allows for effects of scatter changes due to density variations even in the third dimension. It should be noted that none of the above correction methods takes into account the problem of electron disequilibrium. Indeed, the problem of electron disequilibrium is not considered to be an important effect for lungs in large fields. However, if lung shields are used in high energy photon beams (>6 MV) then electrons generated in the non-shielded regions can travel long distances in the low density lung tissues. This has the net effect of enlarging the effective penumbra

Table 5

Dose correction factors for lung in the anatomical phantom using different calculation methods with 50 x 60 cm^2 cobalt-60 radiation fields. (Average lung density = 0.37, overlying tissue = 3.5 cm). Data from Reference 96.

		Dose correction factors						
Depth from surface	Lung depth	Linear attenuation method	Effective attenuation method	Generalized Batho per McDonald et al	Generalized Batho per Sontag and Cunningham	Simplified equivalent TAR	Detailed equivalent TAR	
4.5	1.0	1.035	0.964	0.948	0.948	1.000	.988	
6.5	3.0	1.105	1.053	0.967	0.970	1.035	1.018	
8.5	5.0	1.175	1.132	0.985	0.994	1.073	1.061	
10.5	7.0	1.245	1.216	1.009	1.027	1.116	1.110	
12.5	**9.0**	**1.315**	**1.307**	**1.037**	**1.067**	**1.173**	**1.169**	**- 12%**
14.5	11.0	1.385	1.404	1.066	1.111	1.219	1.217	
16.5	13.0	1.455	1.494	1.098	1.159	1.274	1.264	
18.5	15.0	1.525	1.589	1.126	1.208	1.334	1.328	
20.5	**17.0**	**1.595**	**1.690**	**1.159**	**1.261**	**1.409**	**1.386**	**- 20%**

at the edge of the shielded region resulting in a higher lung dose and a lower target dose.

4.4.1. Methods of lung dose determination

The most important parameter in providing inhomogeneity corrections is the geometric outline of the inhomogeneities actual density of the inhomogeneity is also required but its tolerances are not as stringent for the same error levels[101]. Marinello et al.[62] systematically evaluated the factors influencing lung dose calculations specifically for TBI with cobalt-60 gamma rays. They concluded the following. (1) The lung inhomogeneity correction factor is independent of the height of the patient. Once a certain amount (6 ax) of scattering material is added superior and inferior to a simulated lung phantom, no further variation in correction factor is observed. (2) The correction factor in the middle of lung is independent of lung height in the superior-inferior direction. Only for very young children and for lung heights of about 10 cm did the correction factors increase by approximately 2%. (3) Similar to the conclusion by Van Dyk et al.[96], a simple ratio of TAR method should provide inhomogeneity corrections to the middle of lung that are accurate to 3%.

The following five methods of determining lung doses are listed in order of decreasing complexity as well as decreasing accuracy.

a) CT data and pixel based dose calculations. By performing CT scans of patients who are to be irradiated, precise patient specific anatomic and density information will be obtained. If these pixel based data can be entered into the treatment planning computer and used directly for dose calculation purposes an accuracy of ±3% can be achieved[96].

b) *CT data for contours and average density data.* For those institutions which do not have a treatment planning system capable of using the CT pixel data directly, it is still possible to use CT scans to determine the outlines of the external-contour of the patient as well as major internal structures such as lungs. In this case an average uniform density will have to be assumed for the lungs. The most accurate method of density determination will be to use the "track cursor" in the CT scanner console or an automatic contouring procedure, to outline the lung so that an average lung density specific to that particular lung will be obtained[101]. Lacking this possibility, an average density related to the patient's age can be used. It has been observed[98] that the average density of lung decreases linearly with age, the average density at 5 and 80 years being 0.35 and 0.19 g/cm^3, respectively. Using this average age related density as well as the CT derived contour data as input, dose calculations will be accurate to ±5% for 67% of the patients with normal lungs although 33% of patients with normal lungs as well as most patients with diseased lungs could have calculated lung doses in error by more than 5%[101].

c) *Transmission measurements.* The use of a small cobalt-60 beam or a higher energy x ray beam with a detector behind the patient will allow the determination of an equivalent unit density thickness of the patient at specific points through-out the lung region[22]. This equivalent thickness as well as the actual physical thickness will allow a dose correction factor to be determined specific to that patient[96]. The only practical problem is the determination of a suitable location for the transmission measurement.

d) *Radiographs.* An even simpler but less accurate method is to take lateral radiographs of the thorax for AP/PA treatments or AP radiographs for lateral treatments to determine the total lung thickness. An age related average density[98] can then be used to determine an effective calculation depth.

e) *Nomograph relating dose correction factor and patient thickness.* In a paper evaluating the response of lung to radiation absorbed dose[97], twenty three patients had CT scans performed for dose calculation purposes. When the dose correction factors for the middle of lung for cobalt-60 gamma rays were plotted as a function of patient thickness, 80% of the data points fell within 1.5% of a straight line. The maximum deviation for normal lungs was 3.5%. Diseased lungs showed. much larger deviations. Table 6 summarizes these results for cobalt-60 radiation. Additional higher energy data were derived by first converting these dose correction factors to an equivalent depth and then determining the dose correction factor for higher energy x rays. Lacking any detailed patient anatomic data, other than the patient thickness, a dose correction factor can be determined by using the data of Table 6. Although this procedure is better than not making any correction, large errors could occur for diseased lungs.

Table 6

Dose correction factors adjusting for low-density lung tissues versus patient thickness.

Patient thickness (cm)	Lung dose correction factor cobalt-60	6 MV	25 MV
12	1.04	1.04	1.02
16	1.09	1.09	1.06
20	1.14	1.13	1.09
24	1.19	1.17	1.12
28	1.24	1.21	1.14

4.4.2 Methods of bone dose determination

The factors affecting the dose to bone have not been considered in as much detail as the dose to lung since it has been assumed that the overall changes in the dose distributions as a result of bone inhomogeneities are not nearly as great as the changes due to large volumes of low density lung tissues. However, the majority of TBI procedures are performed for hematological diseases and, therefore, the dose to blood forming organs, such as bone, is particularly important.

To consider the dose to bone, it should be noted that local energy absorption of radiation is a two-stage process[46]. First, electrons are set in motion by the interaction of photons. Second, electrons travel through the medium and deposit their energy by ionization along their paths. (The first is kerma, the second is absorbed dose). Assuming that electronic equilibrium exists, and that the absorbed dose to water is known, the latter can be converted to a dose to bone by

$$D_{bone} = D_{water} \left(\frac{\bar{\mu}_{en}}{\rho} \right)_{water}^{bone} \qquad (5)$$

where D_{bone} and D_{water} are the dose to bone and water, respectively, and $\left(\frac{\bar{\mu}_{en}}{\rho} \right)_{water}^{bone}$ is the ratio formed by averaging the mass energy absorption coefficients for bone and water over the photon spectrum at that point.

Examples of $\left(\frac{\bar{\mu}_{en}}{\rho} \right)_{water}^{bone}$ are shown in Table 7 for compact bone (r = 1.65 g/cm^3) for primary beam spectra as well as spectra generated by Monte Carlo calculations at a depth of 20 cm and a field radius of 50 cm. It is clear from this table that, per gram, the dose to bone is very much dependent on the spectrum of photons at the location of bone. For cobalt-60 radiation the calculated dose to bone can differ by 19% depending on whether an in air primary cobalt-60 spectrum is used or whether a phantom spectrum at a depth Of 20 cm and field radius of 50 cm is used. In reality, the dose to bone will likely be anywhere between these two extremes. For high energy x rays the differences in calculated dose to bone for “in air”

and "in phantom" spectra decreases. The dose delivered to bone approaches the dose delivered to muscle tissue as the energy is increased to 12 MV. The data for 26 MV indicate that there is a strong dependence on photon spectrum even for the same nominal energy. Clearly, an accurate statement of dose to bone will require a knowledge of the spectrum at the location of bone. The data by Cunningham et al.[10] can be used to make a first approximation.

Table 7

Ratios of mass energy absorption coefficients for various photon spectra. These ratios for tissue to water are essentially the same for the primary beam spectrum and the spectrum at depth of 20 cm and field radius of 50 cm. Data are from Reference 10.

	$\left(\frac{\bar{\mu}_{en}}{\rho}\right)_{water}^{tissue}$	$\left(\frac{\bar{\mu}_{en}}{\rho}\right)_{water}^{bone}$		
Photon spectrum		Primary spectrum "in air"	Spectrum "in phantom" (depth of 20 cm, radius of 50 cm)	% difference between primary and phantom spectrum for one.
cobalt-60	.991	0.954	1.134	19
6 MV	.991	0.959	1.089	14
12 MV	.990	0.979	1.025	5
18 MV	.989	0.993	1.029	4
26 MV (thin target)	.990	1.005	1.072	7
26 MV (thick target)	.990	0.991	1.046	6
45 MV	.989	1.027	1.048	2

For lower energy photons (orthovoltage range) the electron tracks are so short that the absorbed dose takes place where the photons interact. At higher energies where the electron tracks are longer, the energy is carried away from the site of interaction and the absorbed dose may be deposited at a different location. Because of the size of bone and the differences in atomic numbers of bone, bone marrow and soft tissues, a lack of electronic equilibrium will exist in the interface region. When the electron tracks are long compared to the thickness of bone, the dose to bone is calculated by

$$D_{bone} = D_{water} \left(\frac{\bar{\mu}_{en}}{\rho}\right)_{water}^{tissue} \bar{\bar{S}}_{tissue}^{bone} \qquad (6)$$

where $\bar{\bar{S}}_{tissue}^{bone}$ is the ratio of averaged stopping powers (averaged over both the spectrum of photon energies and the resulting spectrum of electron energies).

Johns and Cunningham[46] have calculated the dose to bone for cobalt-60 radiation as shown in Figure 8 using the primary beam spectrum. In this case the dose to bone is about 5% less than the dose to muscle tissue. For a large field and a large depth the dose

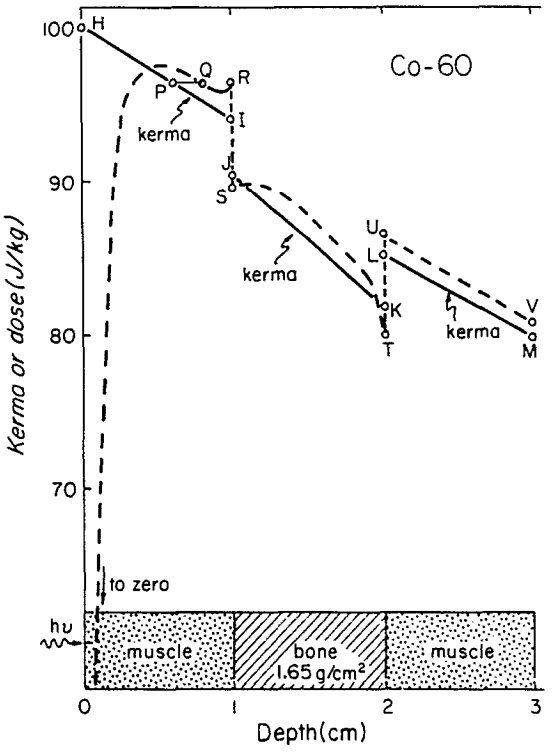

Figure 8: Relative kerma or dose versus depth for a composite muscle-bone phantom. This calculation was performed for a primary cobalt-60 beam spectrum. For a cobalt-60 spectrum for a 50 cm field radius at a depth of 20 cm the dose to bone would be 19% higher than shown while the dose to tissue would remain the same. This Figure was adapted from Reference 45, Page 260.

to bone could be 12% higher than muscle tissue as shown by the data of Table 7.

The problems of: (1) the difficulty in determination of the photon spectrum at the position of bony structures within a patient, (2) the complicated anatomical relationship between bone, bone marrow, blood forming cells and soft tissues, and (3) the difficulty in calculating the dose in interface regions, leave a great deal of uncertainty in the determination of dose to bone and the corresponding blood forming cells. Clearly, additional research into these questions is required.

4.4.3 Methods of compensating for inhomogeneities

When a prescribed tumor dose is well above the lung tolerance, the dose to lung will have to be reduced to minimize the probability of lung complications. A variety of methods can be used to reduce the dose to lung. The following summarizes several alternatives in order of decreasing complexity and decreasing accuracy.

a) *Lung compensators based on dose calculations using CT scan data.* The use of CT data for dose calculations can be applied to provide accurate corrections for tissue inhomogeneities. With the availability of accurate dose calculations, the production of compensators to correct for dose variation as a result of both surface contour effects and tissue inhomogeneities is a relatively easy although time consuming next step. By performing calculations on a number of different planes using appropriate CT scans, a three dimensional compensator can be produced such that a uniform dose is applied to lung and other tissues. This procedure could be extended one step further to produce a uniform but lower dose to lungs compared to the dose to all other tissues.

b) *Constant thickness lung attenuators.* Once the maximum dose to lung has been determined, constant thickness attenuating material can be placed in the beam to reduce the maximum dose to lung to the desired level. The shape of these attenuators can be determined from simulator or diagnostic x ray films or from therapy beam verification (port) films. The attenuators can be made to cover the total lung region only or, if the mediastinal dose is not a serious concern, they could be made rectangular in shape to cover most of the thorax. Of course, a constant thickness attenuator allows an overall reduction in lung dose but does little to decrease the dose variation throughout the lung volume. In some institutions, these attenuators are placed directly on the skin surface[71]. It is, then, essential that the surface dose be measured to avoid increased skin dose due to absorber-tissue interface effects. For some energies these effects can be minimized by placing a couple of mm of tissue equivalent plastic between the absorber and the skin.

c) *Lung blocks for part of the treatment.* The simplest procedure to reduce the dose to the lung is to use standard shielding blocks for a fraction of the total treatment time. Hence, if the lung dose is to be reduced by 20%, then a shielding block could be placed above the lung for the appropriate length of time. The transmission of radiation through the blocks and the scatter from

the high dose regions should not be neglected in determining the fraction of time the block should be in place. The positioning of such blocks should be determined radiographically. Again this procedure -allows for an overall reduction in lung dose but does nothing to reduce the variation throughout the lung volume.

4.5 Dose distributions

Previous sections have discussed the various problems in delivering a uniform dose to the patient. The effects of many of these parameters can be evaluated by looking at a series of typical dose distributions for a number of different treatment conditions calculated in the thorax region of an "average" male patient. The homogeneous water-like distributions were calculated using the scatter-air ratio methodology developed by Cunningham[12]. Inhomogeneity corrections were made using the equivalent TAR method as derived by Sontag and Cunningham[83]. This method corrects for tissue inhomogeneities as well as for the lack of backscatter and side scatter for non-infinite phantoms. The accuracy of the inhomogeneity correction methods has been tested specifically for large field treatments and found to be within ±3% for cobalt-60 to 25 MV x rays[96].

CT scans were used to derive the external patient contour as well as the internal densities. Figure 9 illustrates the resultant distributions for cobalt-60 radiation. Parts (a), (b) and (c) show distributions for AP treatments with an SSD of 150 cm. Part (a) includes an external contour correction but assumes the patient is water equivalent. This would be representative of the abdominal area where inhomogeneities have relatively little effect. Part (b) demonstrates a distribution for the situation with full bolus (or full compensation) such that the patient essentially simulates a rectangular volume of tissue. In part (c), inhomogeneity corrections are included for the full bolus situation. In each figure, an approximate percent variation throughout this dose distribution is indicated. It is clear that the use of full bolus (or compensation) reduces the dose variation in a homogeneous calculation but that lung inhomogeneities produce a large increase in dose. The same distributions were also calculated for lateral opposed fields with an extended SSD of 300 cm. In each case, the dose variation increases by about a factor of 2. Figure 10 shows an identical set of calculations for 25 MV x rays. In most instances, the dose variation is reduced by a factor of 2 compared to the cobalt-60 calculations. These data are consistent with the central axis data shown in Figure 2. Clearly, the construction of individualized lung compensators would provide improved dose uniformity throughout the lung volume.

4.6 Anthropomorphic phantom measurements

Phantom measurements are helpful for understanding the complex dose distributions characteristic of TBI or HBI techniques. When tissue inhomogeneities occupy a portion of a large treatment field, dose correction routines may be inaccurate. The use of a "standard man" or "standard woman" anthropomorphic phantom provides a check for the calculated values. Measurements with closely spaced thermoluminescent dosimeters (TLD's) are most useful for half-body or total body treatment using AP/PA fields with the patient supine and prone on (or near) the treatment room floor. The patient's anatomy does not distort for this positioning, and the phantom construction is

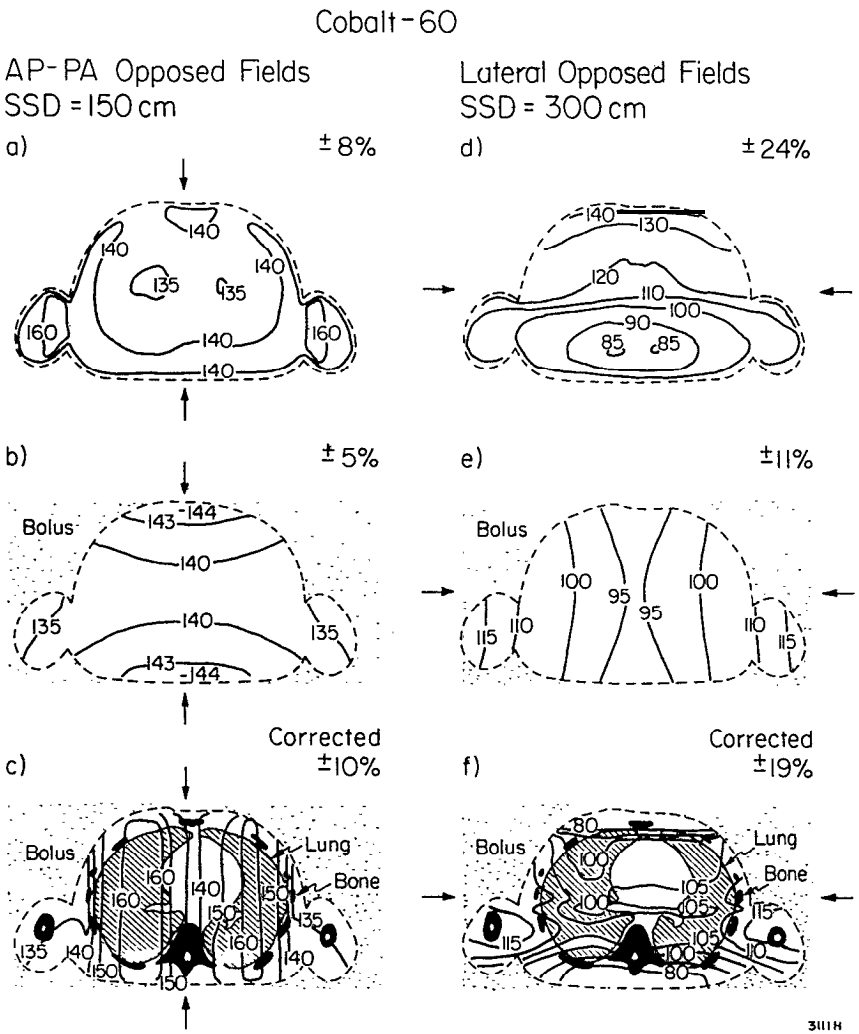

Figure 9: Dose distributions for large parallel opposed fields of cobalt-60 radiation.

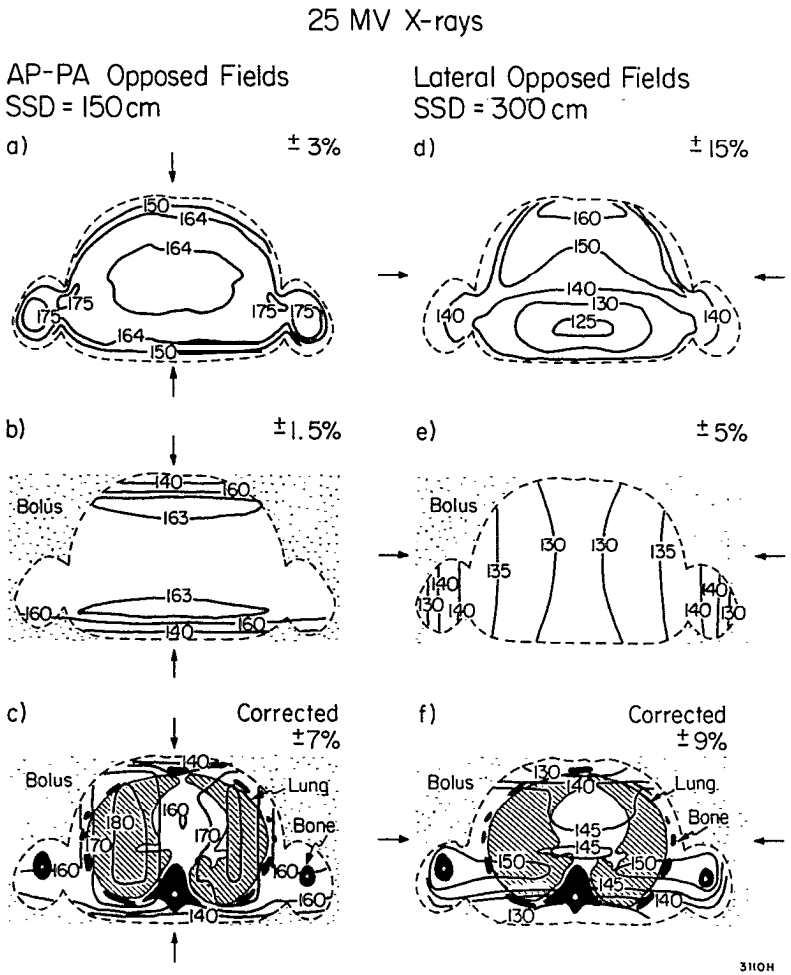

Figure 10: Dose distributions for large parallel opposed fields of 25 MV x-radiation.

representative. For patients treated on their side with horizontally directed AP/PA fields, radiographs and CT scans show a significant change of the lung geometry. Also, changes of the surface contour can be observed for heavy patients. In order to simulate patients positioned supine and treated with bilateral fields, some absorbing material must be placed alongside the phantom to account for arm absorption. It is difficult to duplicate the shape of the upper arms in the region of the shoulder in this way. The attenuation of the beam by the arms is important since the arms can be used to shield the lungs although the reproduction of arm positioning for both CT scanning and treatment is very difficult. If lung attenuators are to be used for lateral fields it may be better to position one arm over the head such that a more reproducible lung geometry is maintained.

A number of cooperative group studies have adopted protocols for treating children with TBI. Unfortunately, this pediatric population varies considerably in size and weight. Therefore, it is not feasible to construct a "standard" child phantom. However, it is possible to use a modular water phantom which approximates the shape of the body[29]. This technique uses a number of small (15 cm x 15 cm^2) water containers placed side-by-side. The depth of the water in each container is varied to correspond to changes in patient thickness. This type of phantom has been used to check the amount of absorbing material needed to compensate for variations of patient contour and for verifying the calculation of prescribed dose.

4.1 *In vivo* measurements

Limitations of the available input data and inherent problems with the calculation schemes, make accurate determination of TBI dose distributions very difficult using most computer-generated treatment planning systems. Furthermore, variations in patient position can alter the distributions dramatically. For this reason it is desirable to have an *in vivo* measurement technique available. One convenient method for finding midline doses employs entrance and exit thermoluminescent dosimeters (TLD's). However, this approach must be used with care in order to obtain acceptable results. Measurements on a single patient could yield less accurate midline doses than those determined by calculation. Systematic errors can result if the effects of lack of scatter on the exit surface are not considered (see Sections 3.2 and 4.2.3). Averaging the appropriate data derived from a large number of patients can be useful. Entrance and exit dosimeters, for example, can be used to monitor the midline dose to the lung. By averaging over a number of patients and separating into broad ranges of patient size, correction factors can be generated for predicting the lung dose for any situation. However, this methodology should be carefully verified with phantom experiments.

Various steps can be taken to minimize the errors associated with this approach. For instance, for the pediatric population of patients (thicknesses less than 20 cm), a simple average of the entrance and exit values will give a dose which is 1 to 2% lower than the true midline value. For thicker patients the shape of the depth dose curve must be taken into account. Usually, the dosimeters are surrounded with a build-up thickness of material to provide full electronic equilibrium. The dosimeters can be placed at the center of small plastic cylinders which have been cut along the major axis so that one

surface is flat. These cylinders can be taped to the patient with the flat surface toward the skin. For higher energy beams (6 MV or greater) it is possible to provide only partial build-up thickness in order to reduce the size of the equilibrium cap. In this case, the reading must be corrected before it is used to predict the midline dose value. Another problem with the use of entrance and exit dosimeters is the accurate placement of the exit dose measuring device. Again using measurements of lung dose as an example, erroneously low readings will result if the exit dosimeter is not properly placed and lies in the "shadow" of the diaphragmatic dome or mediastinal structures. It is for this reason that port filming is suggested as a means of checking the dosimeter position. Entrance and exit measurements have also been used to predict the dose for areas where the diameters of body parts vary significantly. For instance, this technique has been used to monitor the dose for the areas of the head, neck, shoulders and legs. Figure 7 shows entrance and exit measurements obtained for the purpose of checking compensator thicknesses. A simple average has been used to predict the midline value. Notice that some midline values have been measured directly; e.g., the dose in the mouth, in the rectum, between the legs and between the feet. For lateral treatments, dosimeters placed near the axillae can offer useful estimates of the lateral dose uniformity as well as an indication of lung dose.

Quast[74] in reviewing the discussion on *in vivo* dosimetry at the meeting of Leiden in 1982, indicated some potential problems with *in vivo* radiation detectors. Table 8 is similar to that of Quast[74] although different in detail and summarizes some of these concerns. Ionization chambers generally have unwieldy cables and a high voltage. Although the energy response can be a concern, newer chamber models exhibit a uniform response down to low energies. For pulsed radiations from high energy accelerators, corrections may have to be made for ion recombination. These corrections will not necessarily be the same as those used for conventional treatment geometries. Body temperatures can effect the ionization chamber readings although the instantaneous effect is difficult to assess. For open chambers, temperature correction factors change by 5% as the ionization chamber temperature increases from 22°C to 37°C. Although semiconductor diodes have smaller cables, they tend to be energy dependent. Furthermore, the response of diodes is dependent on the total dose received by the diode. Thermoluminescent dosimeters (TLD) have the advantage of being very small and, therefore, can be placed inside of body cavities. Although they might demonstrate a supralinear effect as a function of dose, proper care in calibration of TLD response with dose should account for this effect. However, the energy response of TLD can be a greater concern. A simple calculation for lithium flouride (LiF) using equation (2) of Section 3.3 indicates a 4% difference in response comparing the primary cobalt-60 spectrum to the large field spectrum of Figure 4. The LiF energy response data were taken from Porta1[73]. Most other thermoluminescent materials except for lithium borate, demonstrate a much larger energy response and therefore will be much more sensitive to variations in the photon spectra. Although TLD has the advantage of small spatial requirements, its major disadvantage is the time required for readout and dose determination.

Table 8

Problems with detectors for *in vivo* dosimetry

	Cable	High voltage	Delay in results	Response dependent on			
				Energy	Dose rate	Dose	Temperature
Ionization chambers	XX	XXX	0	X	X	0	XX
Semi-conductor	X	0	0	XXX	0	XX	X
Thermo-luminescent dosimetry (TLD)	0	0	XX	XX	0	X	0

0 no concern
X mild concern
XX moderate concern
XXX serious concern

5. Special Considerations

5.1 Junctions for HBI techniques

The problem of matching adjacent fields with a resultant uniform dose distribution is even more acute for large fields than it is for conventional field sizes due to the large divergence of the beam edge and the large patient volumes in the junction region. One means of minimizing the problem has been indicated by Leung et al.[56]. In their large field irradiator, the collimators along the longitudinal axis of the patient were designed to move independently. Closing one collimator to the central ray allows for a non-divergent beam edge. This can then be located at the junction for both the upper and lower halves resulting in uniform distribution. However, most institutions do not have such a specially-designed dedicated facility. Therefore, methods of minimizing dose inhomogeneity in the junction region must be found.

There are several considerations in dealing with junctions for half body irradiation. The first is a radiobiological question. Usually an upper and lower half body are given as single fractions with a time interval of about 6 weeks. What is the biological effect of summing two doses with a six week interval? Although a clear answer to this question is not available, it can be stated that a specified dose given in two fractions over six weeks will have a smaller biological effect than if the same dose were given in a single fraction. Hence, a higher dose might be allowable in the junction region. The second question relates to tissues at risk in the junction region. Normally the junction is taken at the level of the umbilicus. The intestines and spinal cord appear to be the tissues at risk. For the doses usually given, there may be a short term acute response for intestines but there does not appear to be a long term permanent toxic response. With these considerations, it appears that a higher dose might be allowable in the junction region compared to the rest of the target volume.

The standard method of matching two pairs of parallel opposed fields requires that a gap, S, be calculated at the skin surface between the adjacent fields to account for beam divergence:

$$S_1 = \frac{1}{2} W_1 \left(\frac{d}{SSD_1} \right) \tag{7}$$

$$S_2 = \frac{1}{2} W_2 \left(\frac{d}{SSD_2} \right) \tag{8}$$

$$S = S_1 + S_2 \tag{9}$$

where S_1 represents the distance between the beam edge at the surface to the beam edge at depth d for field 1.

W_1 is the width of field 1 at the surface.

SSD_1 is the source-to-surface distance for field 1.

S_2, W_2 and SSD_2 are the analogous quantities for field 2.

A sample distribution using a gap to match beam edges at midplane is shown in Figure 11(a) for cobalt-60 radiation at an SSD of 150 cm, patient separation of 22 cm and field sizes of 60 x 50 cm^2. With a

uniform dose at the midplane (-133%) an under dosage results towards the skin surfaces (-110%). However, without any gap, the distribution of Figure 11(b) indicates a maximum dose of more than -215% compared to a midplane dose of -140%. Which of these distributions is acceptable is dependent on the clinical situation. The following should be kept in mind when matching fields. First, beam penumbras should be reasonably large (2-5 cm) in the junction region. Small penumbras will cause high hot spots or low cold spots if there is any mismatch of the field edges. Second, the penumbras of the two matching fields should have similar shapes i.e. matching a small penumbra from one field with a large penumbra of the other will result in a dosage non-uniformity. Third, the light field and the 50% dose level should be coincident. Any deviation should be accounted for when matching fields.

A practical consideration is the marking of the patient such that a good junction match can be produced with a 6 week time interval. One method is to tattoo a small dot on the patient on both lateral sides at the mid-plane level using the divergent light field as a reference. These tattoos will be readily recognizable for the second treatment and the edge of the light field can again be used but now on the opposite half of the patient. For obsese patients such lateral marks are not very reproducible and more stable reference points may have to be used.

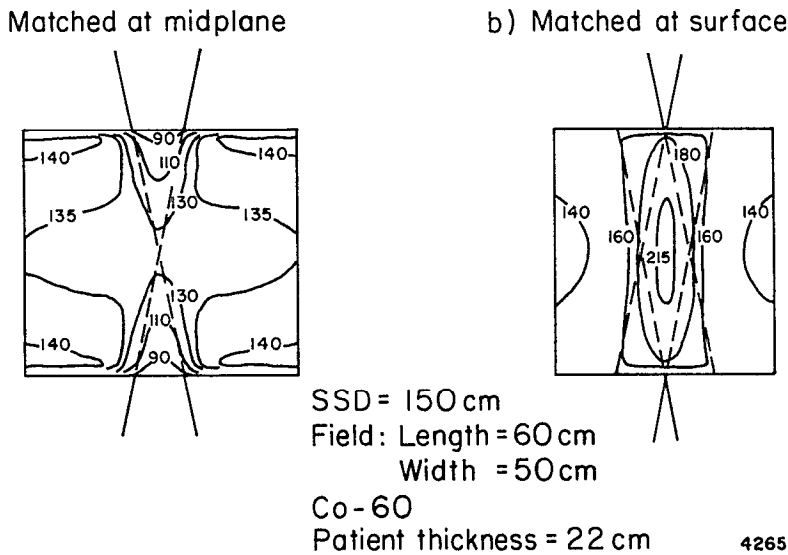

Figure 11: Dose distributions in the junction region for two half body fields for cobalt-60 radiation with an SSD of 150 cm. Patient thickness is 22 cm and the field size is 60 x 50 cm^2. (a) Field edges are matched at the patient midplane. (b) Field edges are matched at the skin surface.

5.2 Shielding

It is sometimes necessary to shield certain organs for patients receiving TBI, HBI or TLI. For example, patients who have been given large doses of Adriamycin prior to their-selection for bone marrow transplantation and TBI, may require complete protection of the heart. Adriamycin sensitizes heart tissue and limits the total body dose unless regional shielding is used. Also, at least one institution[80] has used full shielding blocks for protection of the lungs. In both such cases electron boost fields may be needed to fill in the rib areas surrounding these organs since the bone-marrow containing portions of the chest wall are part of the target volume.

Shielding of organs is usually straightforward for those institutions using specially designed TBI machines since the overall geometry for these units is similar to that used for treating Hodgkin's patients with a mantle field i.e. AP/PA fields with the patient supine and prone using a treatment distance of approximately 2 meters. For those institutions using a standard treatment unit with the beam directed toward a distant wall (4 to 5 meters away) control of the block positioning is much more difficult. The problems are similar to those described in section 4.3.2 for the use of compensators.

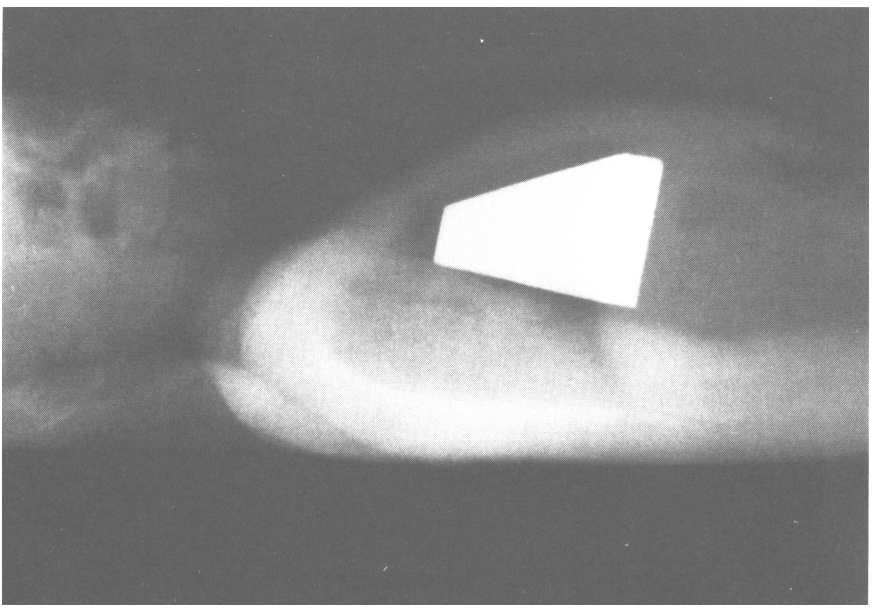

Figure 12: X ray port film for patient treated laterally. Notice the block shielding the heart.

The steps used to fashion and mount a heart block can be used as an illustration of the shielding procedure. For protection of this organ, especially for lateral fields where a radiograph is not particularly helpful, it is necessary- to determine the block size using transverse CT scans. These scans also provide the thickness of the chest wall as required for selection of the electron energy needed to boost the chest wall. Figure 12 shows a block covering the heart for a lateral field technique. The dimensions taken from the CT scans were reduced to account for placement of the block at a small distance (approximately 20 cm) from the patient. The block was placed on a Styrofoam base and secured to a small table positioned next to the patient support couch. The mask for the electron boost field is cut to agree with the size of the projected photon field block on the patient's skin. Obviously, for the block positioning to be accurate, frequent port filming is essential.

5.3 Port filming

There are two applications for port filming of large field patients. First, port films are helpful for guaranteeing that the full patient is within the useful portion of the radiation beam. Second, the positioning of shielding blocks, compensators and exit dosimeters can be checked.

Since many centers performing TBI are forced to use fields of less than adequate size due to restrictive treatment room dimensions, the legs must be drawn up and the head bent forward in order to fit within the beam. Figure 13 illustrates the troublesome areas for a typical treatment geometry. Under the conditions shown, port filming of the head and feet is essential especially for those treatment units where a circular primary collimator restricts the use of the beam diagonal. A simple method for determining the field coverage uses a few single-point film densitometer readings in the areas of the head and feet. It should be recognized that a non-coincidence of the x ray field relative to the light localizing field of as little as 3 mm at an isocenter of 80 cm will project to 1.5 cm at an extended treatment distance of 400 cm. A similar magnification of the entire penumbra region will occur. Thus, underdosing can easily occur for the large treatment distances used for TBI.

Figure 13 shows a typical port film for checking the placement of compensators and exit dosimeters. The leg compensator can be seen as a line projecting across the upper thigh. Though the edge of the one-dimensional compensator used for the head was just visible on the original x rays, it cannot be detected on the print shown. However, it is possible to tape a piece of lead solder along the compensator edge for better visualization. It is also possible to use markers to indicate the position of dosimeters placed on the patient's skin or within cavities.

The port film shown in Figure 13 also contains additional information in that the image (actually a set of two films) provides a record of the amount of lung shielded by the arms. There is no standard for arm positioning for TBI. Especially for the AP/PA technique, a wide variation in approach has been noted. These variations can lead to considerable differences in the dose distribution for the thorax region, so that the port film represents an important patient record.

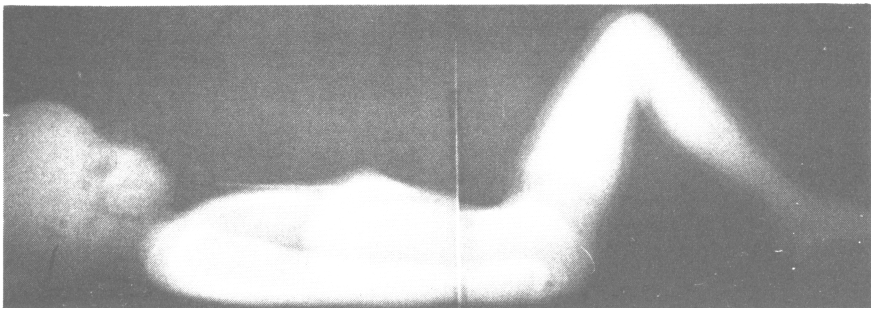

Figure 13: X ray port film for patient of Figure 12. Notice the edge of the leg compensator cutting across the upper part of the legs. Also note the outline of the lungs and the shielding of this organ by the arms.

6.0 Summary of Recommendations

1. Choice of irradiation method
   a) For the energy range between cobalt-60 and 25 MV x rays, the higher energies will give a more uniform. dose distribution (not considering the build-up region).
   b) AP/PA parallel opposed fields are preferred although under some conditions (e.g. pediatric cases, higher energies,, a ±10% uniformity can be achieved with bilateral fields.
   c) If the build-up region is of concern, beam spoilers or electron filters should be employed. For parallel opposed fields, the contribution of exit dose should not be neglected.

2. Basic phantom dosimetry
   a) Phantoms
      i) Water is the material of choice.
      ii) Minimum phantom size should be 30 x 30 x 30 cm^3. Larger phantoms are preferred and can be obtained by placing water equivalent phantom materials about the minimum phantom size.
      iii) Plastic phantoms will need corrections to convert to water as per TG21.
      iv) Smaller phantoms will need corrections for the lack of full scatter. These corrections are dependent on phantom size, field size and energy.
   b) Dosimeters
      i) Dosimeter response should be energy independent.
      ii) Stem and cable effects should be checked and must be minimal.
   c) Dose calibration
      i) In-air measurements should be avoided for the dose calibration.
      ii) Use procedures as recommended by TG21 with the following changes :
         A. Distance and field sizes should be representative of large field geometry.
         B. Large field chamber factors should be applied.
         C. A water or plastic phantom no smaller than 30 x 30 x 30 cm^3 should be used. Larger phantoms are preferred. For plastic phantoms appropriate corrections must be made to determine the dose to water.
         D. The measurements should be corrected to provide results representative of full scattering conditions.
         E. If trays for shielding or compensators or if beam spoilers are used routinely, then these should be in position when the dose calibration is performed.
   d) Central ray data
      i) "In phantom" dose ratios, such as percent depth doses, tissue-phantom ratios or tissue-maximum ratios should be determined. If limited phantom sizes are used, corrections should be made for the lack of full scatter.
      ii) Use cylindrical chamber at depth.
      iii) Use parallel plate chamber in build-up region.
   e) Inverse square test
      i) Inverse square should be tested over the range of possible treatment distances that may be used for large field treatment.

ii) If deviations from the inverse square law are greater than 2% then dose calibrations should be performed at a number of distances.

3. Output factors
   i) Output factors measured “in air” should be avoided unless additional scatter from the floor or walls can be accounted for.
   ii) If a range of field sizes are to be used, then dose calibrations should be performed for a number of field sizes.
   iii) If trays or filters are used routinely, then these should be in position for each field size.

4. Beam profiles
   i) Dose ratios as measured on the central ray should also be measured along several lines which intercept the long and short field axes and are parallel to the central ray. This will provide dose profiles at any depth along the principal planes assuming all the data are normalized to a reading at some depth along the central ray.
   ii) Flattening filters may have to be designed to provide a sufficient beam homogeneity.

5. Attenuation data
   i) Attenuation coefficients should be measured for at least three field sizes under treatment conditions. A plot of broad beam attenuation coefficient versus side of square field will allow interpolation for any field size.
   ii) The broad beam coefficient to be used for treatment calculations will depend on the size of the absorber or the field size if the absorber covers the entire beam.

6. Dose prescription
   i) The dose should be prescribed to one point.
   ii) Dose limits should be specified.
   iii) The dose limit to critical organs may have to be specified separately.
   iv) The therapy unit dose rate as well as the total duration of the treatment may need to be specified depending on the clinical requirements.

7. Corrections for patient size
   i) Equivalent patient dimensions should be determined and central ray data for full scattering conditions should be corrected to account for the patient dimensions both in the lateral and depth directions.
   ii) Contour corrections should be made using methods that are field size dependent i.e. ratio of tissue-air ratio method or ratio of tissue-phantom ratio method.
   iii) For treatment time or monitor unit calculations either the field size or the patient size should be used depending on which one is smaller.
   iv) The lack of backscatter can result in noticeable dose reduction especially for small patient thicknesses. The use of bolus or backscattering material will help to reduce this effect.

8. Compensators for missing tissue
   i) The use of tissue equivalent bolus is the easiest if the loss of skin sparing is not a problem.
   ii) If skin sparing is to be maintained compensators-should be built.

9. Inhomogeneities: Lung
   i) When the dose to lung is critical, inhomogeneity corrections should be made.
      a) For manual calculations, methods which are field size dependent should be used i.e. ratio of TAR's or ratio of TPR's (the power-law method should not be used with TAR's since it is in error for large fields).
   ii) Inhomogeneity corrections should be made using CT data to provide anatomical information.
      a) The best is to perform pixel based calculations.
      b) Alternately, an average density can be assumed.
      c) Without CT, transmission measurements can provide equivalent thickness data.
      d) Radiographs combined with an assumption about average density will provide an estimate of equivalent thickness.
      e) The simple relation between dose correction factor and patient thickness can be used if no other information is available.

10. Inhomogeneities: Bone
   i) The dose to bone can be calculated from equation (5) for lower energy photons in section 4.4.2 using the data of Table 7 or equation (6) for higher energy photons.

11. Methods of compensating for lung tissues
   i) Produce three dimensional compensator from full dose distributions with inhomogeneity corrections.
   ii) Use constant thickness lung attenuator.
   iii) Use shields for part of the treatment time.

12. Anthropomorphic phantom measurements
   i) TLD measurements should be performed in anthropomorphic phantoms using the treatment techniques and doses that are representative of patient treatments. The TLD response should be calibrated such that absolute doses can be determined.

13. *In vivo* measurements
   i) The treatment technique should be verified by performing TLD measurements on several representative patients. The TLD response should be calibrated to determine absolute doses. To obtain an indication of the dose at d_{max}, the dosimeters should have sufficient tissue equivalent material such that full build-up is achieved. Exit dosimeter readings should be corrected for the lack of backscatter if they are to be used for determination of midplane doses.

REFERENCES

1. Aget, H., Van Dyk, J., Leung, P.M.K. Utilization of a high energy photon beam for whole body irradiation. Radiol. 123, 747-751, 1977.

2. Beck, W.L., Stokes, T.R., Lushbaugh, C.C. Proceedings of the National Symposium on Natural and Man-made Radiation in space. NASA TMX - 2440, 1971.

3. Bortin, M.M., Gale, R.P., Rimm A.A. For the Advisory Committee of the International Bone Marrow Transplant Registry. Allogeneic bone marrow transplantation for 144 patients with severe aplastic anemia. JAMA 245, 1132-1139, 1981.

4. Bruce, W.R., Johns, H.E. The spectra of x ray scatter in low atomic number materials. Br. J. Radiol. Suppl. 9, 37-38, 1960.

5. Brucer, M . A total body irradiator.. Int. J. Appl. Radiat. Isotopes. 10, 99-105, 1961.

6. Cassady, J.R., Order, S., Camitta, B., Marck, A. Modification of gastro-intestinal symptoms following irradiation by low dose rate technique. Int. J. Radiat. Oncol. Biol. Phys. 1, 15-20, 1976.

7. Chaffey, J.T., Rosenthal, D.S., Moloney, W.C., Hellman, S. Total body irradiation as treatment for lymphosarcoma. Int. J. Radiat. Oncol. Biol. Phys. 1, 399-405, 1976.

8. Christ, G. The dosimetry of total body irradiation at the University of Tubingen. J. Eur. Radiother. 3, 211-212, 1982.

9. Conference on Dosimetry of Total Body Irradiation by External Photon Beams. Oak Ridge Associated Universities. February 1967. Cloutier, R.J., O'Foghludha, F., Comas, F.V. (Eds) Document CONF-670219, Health and Safety (TID-4500). Published by Oak Ridge Associated Universities.

10. Cunningham, J.R., Woo, M., Rogers, D.W.O., Bielajew, A.F. The dependence of mass energy absorption coefficient ratios on beam size in a phantom. Medical Physics 13, 496-502, 1986.

11. Cunningham, J.R., Wright, D.J. A simple facility for whole-body irradiation. Radiol. 78, 941-949, 1962

12. Cunningham, J.R., Scatter-air ratios. Phys. Med. Biol. 17, 42-51, 1972.

13. Cunningham J.R. Tissue inhomogeneity corrections in photon-beam treatment planning. In: Progress in Medical Radiation Physics Vol. 1, Orton, C.G. (Ed.) Plenum Publishing Corp. 103-131, 1982.

14. D'Angio, G.J., Evans, A.E. Cyclic low-dose total body irradiation for metastatic neuro-blastoma. Int. J. Radiat. Oncol. Biol. Phys. 9, 1961-1965, 1983.

15. Doppke, K.P. Treatment strategies - Hodgkin's disease. In: Advances in Radiation Therapy Treatment Planning. Wright, A.E., Boyer, A.L. (Eds.) AAPM Monography No. 9. American Institute of Physics, N.Y., 341-355, 1983.

16. Draeger, R.H., Lee, R.H., Shea, T.E., Whitten, F.I., Eicher, M. Research Report 11, Bethesda, MD. Naval Medical Research Institute, 1219, 1953.

17. Dutreix, J., Gluckman, E., Brule, J.M. Biological problems of total body irradiation. J. Eur. Radiother. 3, 165-173, 1982.

18. El-Khatib, E., Battista, J.J. Improved lung dose calculation using tissue-maximum ratios in the Batho correction. Med. Phys. 11, 279-286, 1984.

19. Engler, M.J., Feldman, M.I., Spira, J. Arc technique for total body irradiation by a 42 MV betatron. Med. Phys. 4, 524-525, 1977.

20. Evans, R.G. Radiobiological considerations in magna-field irradiation. Int. J. Radiat. Oncol. Biol. Phys. 9, 1907-1911, 1983.

21. Faw, F., Glenn, D. Dose distributions if the field size is larger than the patient. (Abstract). Proceedings of the Third International Congress of Medical Physics Sweden, 1972.

22. Fedoruk, S.O., Johns, H.E. Transmission dose measurement for cobalt-60 radiation. Br. J. Radiol. 30, 190-195, 1957.

23. Findley, D.O., Skov, D.D., Blume, K.G. Total body irradiation with a 10 MV linear accelerator in conjunction with bone marrow transplantation. Int. J. Radiat. Oncol. Biol. Phys. 6, 695-702, 1980.

24. Fitzpatrick, P.J., Rider, W.D. Half body radiotherapy of advanced cancer. J. Can. Assoc. Radiol. 27, 75-79, 1976.

25. Fitzpatrick, P.J., Rider, W.D. Half body radiotherapy. Int. J. Radiat. Oncol. Biol. Phys. 1, 197-207, 1976.

26. Fitzpatrick, P.J., Garrett, P.G. Metastatic breast cancer ovarian ablation with lower half-body irradiation. Int. J. Radiat. Oncol. Biol. Phys. 7, 1523-1526, 1981.

27. Gagnon, W.F., Horton, J.L. Physical factors effecting dose to the skin from cobalt-60 gamma rays and 25 MV x rays. Med. Phys. 6, 285-290, 1979.

28. Galvin, J.M., D'Angio, G.J., Walsh, G. Use of tissue compensators to improve the dose uniformity for total body irradiation. Int. J. Radiat. Oncol. Biol. Phys. 6, 767-771, 1980.

29. Galvin, J.M. Calculation and prescription of dose for total body irradiation. Int. J. Radiat. Oncol. Biol. Phys. 9, 1919-1924, 1983.

30. Geise, R.A., McCullough, E.C. The use of CT scanner in megavoltage photon beam therapy planning. Radiol. 124 133-141, 1977.

31. Glasgow, G.P., Mill, W.B. Cobalt-60 total body irradiation dosimetry at 220 cm source-axis distance. Int. J. Radiat. Oncol. Biol. Phys. 6, 773-777, 1980.

32. Glasgow, G.P., Mills, W.B., Phillips, G.L., Herzig, G.P. Comparative 60-Co total body irradiation (220 cm SAD) and 25 MV total body irradiation (370 cm SAD) dosimetry. Int. J. Radiat. Oncol. Biol. Phys. 6, 1243-1250, 1980.

33. Glasgow, G.P. The dosimetry of fixed, single source hemibody and total body irradiators. Med. Phys. 9, 311-323, 1982.

34. Goolden, A.W.G., Goldman, J.M., Kam, K.C., Dunn, P.A., Baughan, A.S.J., McCarthy, D.M., Worsley, A.M., Gordon-Smith, E.C., Samson, D., Catovsky, D., Galton, D.A.G. Fractionation of whole body irradiation before bone marrow transplantation. Brit. J. Radiol. 56, 245-250, 1983.

35. Halperin, E.C. Total lymphoid irradiation as an immunosuppressive agent for transplantation and the treatment of autoimmune disease: A review. Clin. Radiol. 36, 125-130, 1985.

36. Herring, D.F. Methods for extracting dose response curves from radiation therapy data - A unified approach. In: Patterns of Care Study. American College of Radiology, 1978.

37. Heublein, A.C. Preliminary report on continuous irradiation of the entire body. Radiol. 18, 1051-1062, 1932.

38. Hill, R.P. Response of mouse lung to irradiation at different dose rates. Int. J. Radiat. Oncol. Biol. Phys. 9, 1043-1047, 1983.

39. Hoppe, R.T., Kushlan, P., Kaplan, H.S., Rosenberg, S.A., Brown, B.W. The treatment of advanced stage favourable histology non-Hodgkin's lymphoma: A preliminary report of a randomized trial comparing single agent chemotherapy, combination chemotherapy and whole body irradiation. Blood, 58, 592-598, 1981.

40. International Commission on Radiation Units and Measurements (ICRU). Determination of absorbed dose in a patient irradiated by beams of x or gamma rays in radiotherapy procedures. ICRU Report 24, 1976. (International Commission on Radiation Units and Measurements, Washington).

41. International Commission on Radiation Units and Measurements. Dose specification for reporting external beam therapy with photons and electrons. ICRU Report 29, 1978 (International Commission on Radiation Units and Measurements, Washington).

42. Jacobs, M.L., Pape, L. A total body irradiation chamber and its uses. Int. J. App. Radiat. Isotopes, 9, 141-143, 1960.

43. Jenkin, R.D.T., Rider, W.D., Sonley, M.H. Ewing's sarcoma: A trial of adjuvant total body' irradiation. Radiology, 96, 151-155, 1970.

44. Jenkin, R.D.T., Berry, M.D. Sequential half-body irradiation. Int. J. Radiat. Oncol. Biol. Phys. 9, 1969-1971, 1983.

45. Johns, H.E., Cunningham, J.R. The Physics of Radiology. 4th Edition, Charles C. Thomas, Springfield, ILL, 1983, p 230.

46. Johns, H.E., Cunningham, J.R. The Physics of Radiology, 4th Ed. Charles C. Thomas, Springfield, Ill. 1983, p. 397-402.

47. Johnson, R.E. Radiation therapy of generalized lymphocytic lymphomas. Amer. J. Roentgen. 117, 50-53, 1973.

48. Keane, T.J., Van Dyk, J., Rider, W.D. Idiopathic interstitial pneumonia following bone marrow transplantation: The relationship with total body irradiation. Int. J. Radiat. Oncol. Biol. Phys. 7, 1365-1370, 1981.

49. Kereiakes, J.G., VandeRiet, W., Born, C., Ewing, C. Silberstein, E., Saenger, E. Active bone-marrow dose related to hematological changes in whole-body and partial-body cobalt-60 gamma radiation exposures Radiol. 103, 651-656, 1972.

50. Khan, F.M., Williamson, J.F., Sewchand, W., Kim, T.H. Basic data for dosage calculation and compensation. Int. J. Radiat. Oncol. Biol. Phys. 6, 745-751, 1980.

51. Kim, T.H., Khan, F.M., Galvin, J.M. A report of the work party: Comparison of total body irradiation techniques for bone marrow transplantation. Int. J. Radiat. Oncol. Biol. Phys. 6, 779-784, 1980.

52. Kirby, T.H., Hanson, W.F., Cates, D.A. Total body irradiation dosimetry (Abstract). Med. Phys. 12, 523, 1985.

53. Kolb, H.J., Rieder, I., Bodenberger, U., Netzel, B., Schaffer, E. Kolb, H., Thierfelder, S. Dose rate and dose fractionated studies in total body irradiation of dogs. Path. Biol. 27, 370-372, 1979.

54. Lam, W.C., Lindskoug, B.A., Order, S.E., Grant, D.G. The dosimetry of cobalt-60 total body irradiation. Int. J. Radiat. Oncol. Biol. Phys. 5, 905-911, 1979.

55. Lam, W.C., Order, S.E., Thomas, E.D. Uniformity and standardization of single and opposing cobalt-60 sources for total body irradiation. Int. J. Radiat. Oncol. Biol. Phys. 6, 245-250, 1980.

56. Leung, P.M.K., Rider, W.D., Webb, H.P., Aget, H., Johns, H.E. Cobalt-60 therapy unit for large field irradiation. Int. J. Radiat. Oncol. Biol. Phys. 7, 705-712, 1981.

57. Loeffler, R.K. Therapeutic use of fractionated total body and subtotal body irradiation. Cancer 47, 2253-2258, 1981.

58. Lutz, W.R., Bjarngard, B.E. Design of a versatile total body irradiation facility. (Abstract) Radiological Society of North America, 68th Scientific and Annual Meeting, Chicago, November, 1982.

59. Lutz, W.R., Svensson, G.K., Bjarngard, B.E. Design and beam properties of a 4 MV total body irradiator. (Abstract) Med. Phys. 10, 539, 1983.

60. Lutz, W.R., Larsen, R.D. The effect of flattening filter design in quality variations within an 8-MV primary x ray beam. Med. Phys. 11, 843-845, 1984.

61. Magna-field symposium. Florida. Sponsored by Children's Cancer Study Group in conjunction with ASTR. October, 1982. Int. J. Radiat. Oncol. Biol. Phys. 9, 1905-1973, 1983

62. Marinello, G., Barrie, A.M., LeBourgeois, J.P. Measurement and calculation of lung dose in total body irradiation performed with cobalt-60. J. Eur. Radiotherap. 3, 174-182, 1982.

63. Meeting of Leiden. May 1982. J. Eur. Radiother. 3, 157-264, 1982,

64. Miller, L.S., Fletcher, G.H., Berstner, H.B. Radiobiological observations on cancer patients treated with whole-body x-irradiation. Radiat. Research 4, 150-165, 1958.

65. Miller, R.J., Langdon, E.A., Tesler, A.S. Total body irradiation utilizing a single cobalt-60 source. Int. J. Radiat. Oncol. Biol. Phys. 1, 549-552, 1976.

66. Mulvey, P.J., Godlee, J.N. Technique and dosimetry for TBI at University College Hospital. London. J. Eur. Radiother. 3, 241-242, 1982

67. Payne, D.G., Yeoh, L., Feld; R., Pringle, J.F., Evans, W.K., Herman, J.G., Quirt, I.C. Upper half body irradiation (UHBI) for extensive small cell carcinoma of the lung. Int. J. Radiat. Oncol. Biol. Phys. 9, 1571-1574, 1983.

68. Peters, L.J., Withers, H.R., Cundiff, J.H., Dicke, K.A. Radiobiological considerations in the use of total-body irradiation for bone-marrow transplantation. Radiology, 131, 243-247, 1979.

69. Peters, V.G., Herer, A.S. Modification of a standard cobalt-60 unit for total body irradiation at 150 cm SSD. Int. J. Radiat. Oncol. Biol. Phys. 10, 927-932, 1984.

70. Phillips, G.L., Herzig, R.J., Lazarus, H.M., Fay, J.W., Wolff, S.N., Mill, W.B., Lin, H., Thomas, P.R.M., Glasgow, G.P., Shina, D.C., Herzig G.P. Treatment of resistant malignant lymphoma with cyclophosphamide, total body irradiation and transplantation of cryopreserved autologous marrow. New Eng. J. Med. 310, 1557-1561, 1984

71. Pla, M., Chenery, S.G., Podgorsak, E.B. Total body irradiation with a sweeping beam. Int. J. Radiat. Oncol. Biol. Phys. 9, 83-89, 1983.

72. Podgorsak, E.B., Pla, C., Evans, M.D.C., Pla, M. The influence of phantom size on output, peak-scatter factor and percentage depth dose in large field photon irradiation. Med. Phys. 12, 639-645, 1985.

73. Portal, G. Preparation and properties of principal TL products. In: Applied Thermoluminescence Dosimetry. Oberhofer, M., Scharmann, A. (Eds). Adam Hilger Ltd., Bristol, 1981, p. 97-122.

74. Quast, U. Summary of the discussion on *in vivo* dosimetry. J. Eur. Radioth. 3, 259-261, 1982.

75. Quast, U. Physical treatment planning of total body irradiation - Patient translation and beam zone method. Med. Phys. 12, 567-573, 1985.

76. Rawlinson, J.A. Internal Report, Ontario Cancer Institute, Toronto, Canada 1977.

77. Rider, W.D., Messner, H.A. Magna-field irradiation: Work in progress in bone marrow transplantation at the Princess Margaret Hospital, Toronto. Int. J. Radiat. Oncol. Biol. Phys. 9, 1967, 1983.

78. Rider, W.D., Van Dyk, J. Total and partial body irradiation. In: Radiation Therapy Treatment Planning, Bleehen, N.M., Glatstein, E., Haybittle, J.L. (Eds.), Marcel Dekker Inc., 559-594, 1983.

79. Sahler, O.D. Development of a room specifically designed for total body irradiation. Radiol. 72, 266-267, 1959.

80. Shank, B., Simpson, L. The role of total body irradiation in bone marrow transplantation for leukemia. Bull N.Y. Acad. Sci. 58, 763-777, 1982.

81. Shank B. Techniques of magna-field irradiation. Int. J. Radiat. Oncol. Biol. Phys. 9, 1925-1931, 1983.

82. Sontag, M.R., Battista, J.J., Bronskill, M.J., Cunningham, J.R. Implications of computed tomography for inhomogeneity corrections in photon beam dose calculations. Radiol. 124, 143-149, 1977

83. Sontag, M.R., Cunningham, J.R. The equivalent tissue-air ratio method for making absorbed dose calculations in a heterogeneous medium. Radiol. 129, 787-794, 1978.

84. Spokas, J.J., Meeker, R.D. Investigation of cables for ionization chambers. Med. Phys. 7, 135-140, 1980.

85. Strober, S., Kotzin, B.L., Hoppe, R.T., Slavin, S., Gottlieb, M., Calin, A., Fuks, Z., Kaplan, H.S. The treatment of intractable rheumatoid arthritis with lymphoid irradiation. Int. J. Radiat. Oncol. Biol. Phys. 7, l-7, 1981.

86. Surmont, J., Dutreix, A., Lalanne, C.M. Les irradiations in toto pour greffes de tissu ou transplantation d'organe chez l'homme. Problems techniques. J. Radiol. Electrol. 41, 679-689, 1960.

87. Svensson, G.K., Larson, R.D., Chen, T.S. The use of a 4 MV linear accelerator for whole body irradiation. Int. J. Radiat. Oncol. Biol.Phys. 6, 761-765, 1980

88. Symposium on Total Body Irradiation for Bone Marrow Transplantation. Paris, September, 1978. Pathologie Biologie, 27, 317-380 1979.

89. Task Group 21, Radiation Therapy Committee, American Association of Physicists in Medicine. A protocol for the determination of absorbed dose from high-energy photon and electron beams. Med. Phys. 10, 741-771, 1983.

90. The Physics and Clinical Aspects of Total Body Irradiation Used for Bone Marrow Transplantation. Workshop of the Children's Cancer Study Group, Montreal, June 1978. Int. J. Radiat. Oncol. Biol. Phys. 6, 743-787, 1980.

91. Thomas, E.D., Storb, R., Buckner, C.D. Total body irradiation in preparation for marrow engraftment. Transplantation Proc. 4, 591-593, 1976.

92. Thomas, E.D. The role of marrow transplantation in the eradication of malignant disease. Cancer 49, 1963-1969, 1982.

93. UCLA Bone Marrow Transplantation Team: Bone marrow transplantation for aplastic anemia: Conditioning with cyclophosphamide plus low-dose total body irradiation. In: Experimental Hematology Today 1979. Baum, S.J., Lenney, G.D., (Eds.) New York, Springer Verlag, 185-191, 1979.

94. Urtasun, R.C., Belch, A., Bodnar, D. Hemibody irradiation: An active therapeutic modality for the management of patients with small cell lung cancer. Int. J. Radiat. Oncol. Biol. Phys. 9, 1575-1577, 1983.

95. Van Dyk, J., Leung, P.M.K., Cunningham, J.R. Dosimetric considerations of very large cobalt-60 fields. Int. J. Radiat. oncol. Biol. Phys. 6, 753-759, 1980.

96. Van Dyk, J., Battista, J.J., Rider, W.D. Half body radiotherapy: The use of computed tomography to determine the dose to lung. Int. J. Radiat. Oncol. Biol. Phys. 6, 463-470, 1980.

97. Van Dyk, J., Keane, T.J., Kan S., Rider, W.D., Fryer, C.J.H. Radiation pneumonitis following large single dose irradiation: A re-evaluation based on absolute dose to lung. Int. J. Radiat. Oncol. Biol. Phys. 7, 461-467, 1981.

98. Van Dyk, J., Keane, T.J., Rider, W.D. Lung density as measured by computerized tomography: Implications for radiotherapy. Int. J. Radiat. Oncol. Biol. Phys. 8, 1363-1372, 1982.

99. Van Dyk, J. Magna-field irradiation: Physical considerations. Int. J. Radiat. Oncol. Biol. Phys. 9, 1913-1918, 1983.

100. Van Dyk, J. Whole and partial body radiotherapy: Physical considerations. In: Advances in Radiation Therapy Treatment Planning. Wright, A., Boyer, A.L. (Eds.). AAPM Monograph No. 9, American Institute of Physics. N.Y. 403-426, 1983.

101. Van Dyk, J. Lung Dose calculations using computerize tomography: Is there a need for pixel based procedures? Int. J. Radiat. Oncol Biol. Phys. 9, 1035-1041, 1983.

102. Van Dyk, J. Broad Beam attenuation of cobalt-60 gamma rays and 6, 18 and 25 MV x rays by lead. Med. Phys. 13, 105-110, 1986.

103. Webster, E.W. Physical considerations in the design of facilities for the uniform whole-body irradiation of man. Radiol. 75, 19-32, 1960.

104. Wong, J.W., Henkelman, R.M. Reconsideration of the power-law (Batho) equation for inhomogeneity-corrections. Med. Phys. 9, 521-530, 1982.