

RADIATION DOSIMETRY
FOR FAST ELECTRONS

A THESIS
SUBMITTED TO THE FACULTY OF GRADUATE STUDIES
UNIVERSITY OF MANITOBA

IN PARTIAL FULFILMENT OF THE REQUIREMENTS FOR THE DEGREE
OF MASTER OF SCIENCE

by

MAHMOUD M. H. HAMMOUDAH

1974

ACKNOWLEDGEMENTS

Thanks to my supervisor Dr. A. F. Holloway for his advice, continuous encouragement and help since 1972 and to Dr. D. V. Cormack for useful discussions and private communications. Thanks also to Mr. J. Legal for his assistance in designing the phantoms which were necessary for the measurements, and to Mr. E. Cartwright for his help in the repair and the maintenance of the equipment and machines used.

The financial assistance of the Manitoba Cancer Treatment and Research Foundation and the International Atomic Energy Agency is much appreciated.

RADIATION DOSIMETRY

FOR FAST ELECTRONS

MAHMOUD M. H. HAMMOUDAH

ABSTRACT

The experimental determination of dose in rads was performed using a Fricke ferrous sulfate dosimeter.

The dose was determined for fast electrons with initial energies 15, 20, 25, 30 and 34 MeV at a depth 4.44 gm/cm^2 in a tissue equivalent material phantom. The dose so determined was used to obtain an overall conversion factor (C_E) to convert the readings of a field ionization chamber, corrected for temperature and pressure and calibrated to read exposure for ^{60}Co γ -rays, to the dose in rads for electrons in the same medium. The source of electrons was the Manitoba Cancer Treatment and Research Foundation betatron.

The dose and conversion factor (C_E) were also obtained for ^{60}Co γ -rays using a "Theratron F" radiotherapy unit to check the accuracy and precision of the Fricke dosimeter.

Measurements were also performed using three different collimating arrangements to study their effect on the relative central axis depth dose. The relative central axis dose for 10, 20, and 34 MeV as a function of area from 10 to 200 cm^2 for rectangular, square, and circular field sizes was studied at 2.6 and 4.8 gm/cm^2 depth in a tissue equivalent phantom.

To study the effect of replacement of the original lucite localizer supplied by the manufacturer and a localizer made of metal (brass), isodose distributions were obtained using photographic films placed in a tissue equivalent phantom.

TABLE OF CONTENTS

CHAPTER	PAGE
I. INTRODUCTION	1
II. LITERATURE REVIEW	5
A. Calorimeters	5
(1) Adiabatic Calorimetry	6
(2) Quasi-adiabatic Systems	7
B. Chemical Dosimeters	9
C. Ionization Chamber Dosimetry	15
D. Solid State Dosimeters	23
(1) Solid state integrating dosimeters	23
(2) Solid state electrical conductivity dosimeters	23
E. Photographic Film Dosimeters	23
F. Phantoms	24
G. Practical Factors Influencing the Absorbed Dose	
Distribution for High Energy Electron Beams	25
III. MATERIALS AND METHODS	28
A. Radiation Sources	28
(1) Electrons	28
(2) Cobalt-60 γ -rays source	29
B. Phantoms Used	32
C. Chemical Dosimetry	37
D. Experimental Determination of C_E	38
E. Collimating Devices	39
F. Effect of Replacement of Localizer on Shape of	
Isodose Distribution	40
G. Experimental Arrangements	42
IV. RESULTS	43
A. Determination of the Overall Conversion Factor (C_E)	43

(1) C_E for Cobalt-60 radiation	43
(2) C_E for high energy electron beams 10 - 34 MeV	44
B. Intercomparison Between M.C.F. C_E Values and ICRU C_E Values	46
C. Intercomparison Between Dose Determined by N.R.C. and M.C.F.	46
D. Calibration with Different Collimating Systems	54
E. Influence of Collimator Design on the Shape of the Isodose Distribution	62
V. DISCUSSION	67
VI. CONCLUSIONS	75
BIBLIOGRAPHY	79

LIST OF FIGURES

FIGURE	PAGE
1. Theoretical C_E Values vs. Mean Electron Energy from ICRU 21 (1972)	21
2. Theoretical and Experimental C_E Values vs. Mean Electron Energy	22
3. Electron Collimating System (Treatment and Master Collimators)	31
4. Irradiation Vials Lucite Layer Unclamped	34
5. Irradiation Vials Lucite Layer Clamped	35
6. Ionization Chamber Lucite Layer	36
7. Brass Localizer	41
8. M.C.F. and ICRU C_E Values vs. Mean Electron Energy	52
9. rr/mr vs. Irradiation Area Defined by Lead Plates Measured at 2.6 gm/cm^2 Depth in a Lucite Phantom	56
10. rr/mr vs. Irradiation Area Defined by Lead Plates Measured at 4.8 gm/cm^2 Depth in a Lucite Phantom	57
11. rr/mr vs. Irradiation Area Defined by Lead Plates and Base Sandwich Measured at 2.6 gm/cm^2 Depth in a Lucite Phantom	58
12. rr/mr vs. Irradiation Area Defined by Lead Plates and Base Sandwich Measured at 4.8 gm/cm^2 Depth in a Lucite Phantom	59
13. rr/mr vs. Irradiation Area Defined by Original Collimator Measured at 2.6 gm/cm^2 Depth in a Lucite Phantom	60
14. rr/mr vs. Irradiation Area Defined by Original Collimator Measured at 4.8 gm/cm^2 Depth in a Lucite Phantom	61
15. 8 ϕ Isodose Distribution, 20 MeV B_1 , Lucite Localizer Measured in a Rubber Phantom using Bare X-Ray Film	63

FIGURE	PAGE
16. 8 ϕ cm Isodose Distribution, 20 MeV B_1 , Brass Localizer Measured in a Rubber Phantom using Bare X-Ray Film	64
17. 8 ϕ cm Isodose Distribution, 34 MeV B_2 , Lucite Localizer Measured in a Rubber Phantom using Bare X-Ray Film	65
18. 8 ϕ cm Isodose Distribution, 34 MeV B_2 , Brass Localizer Measured in a Rubber Phantom using Bare X-Ray Film	66
19. Geometry of the Position of the Ionization Chamber relative to the Position of the Scattering Foil in the Betatron	71(a)

LIST OF TABLES

TABLE	PAGE
1. Mass Stopping Power Ration between Water and Dosimeter Solution	14
2. Ferrous Sulfate Dosimetry Results for ^{60}Co γ -rays	48
3. Ionization Dosimetry Results for ^{60}Co γ -rays	48
4. Ferrous Sulfate Dosimetry Results for Electrons	49
5. Ion Chamber Dosimetry Results for Electrons	50
6. Comparison between M.C.F. and ICRU C_E Value for ^{60}Co γ -rays	51
7. Comparison between M.C.F. and ICRU C_E Values for Fast Electrons	51
8. Dose Intercomparison between M.C.F. and N.R.C.	53
9. Uncertainty in the Determination of the Dose at a Point in the Phantom using the Chemical Dosimeter	76

CHAPTER I

INTRODUCTION

Recently electron beams have been used extensively for therapeutic purposes. There are several physical reasons for using electrons in certain clinical situations.

For high energy electrons, above 10 MeV, the surface dose is within 20% of the maximum dose for most field sizes. Consequently the marked decrease in relative dose in the distal surface of internal air cavities such as larynx which can sometimes be a problem for ^{60}Co γ -rays and high energy x-ray therapy is not present when an electron beam is used. The plan for electron treatment of supra-clavicular and axillary nodes by contiguous fields of different energy has unique features not readily achievable with x-radiation. This illustrates the usefulness of electron energies in excess of 30 MeV for special conditions.

Because of the finite depth of the penetration of electrons there are no primary electrons beyond the maximum range. This abrupt termination is unique with charged particles in contrast to gradual attenuation of photons. As a result the adjacent underlying tissues are spared in electron therapy.

The depth dose patterns with electrons offer rapid and simple treatment setups, with only one field in many cases. Electron accelerators have the potential for high dose rates which can shorten the treatment time. The isodose contour can be made reasonably flat by use of appropriate scattering and collimating systems to provide a uniform radiation field of defined size. The depth of penetration of electrons is proportional to their incident energy which can be conveniently controlled.

The RBE (relative biological effectiveness) is found to

be similar to that determined for high energy x-rays (Dutreix, et al. {1}, Wambersie, et al. {2}, Robinson and Ervin {3}, and Kim et al. {4}).

In medical applications of high energy electrons the long term objective of a physics department is to determine the dose delivered at any point in any irradiated patient. It is highly desirable that standard methods for measurements of output and absorbed dose be explicitly described in order to facilitate uniformity of dosimetry. It is necessary to establish standard methods for monitoring of output, measurements of dose and procedures for correlation of output with the dose delivered in the patient.

Electrons for patient treatment from the Manitoba Cancer Treatment and Research Foundation (M.C.F.) Brown Boveri betatron are monitored by an ionization chamber of the transmission type which is sufficiently large in cross-section to include about half of the entire beam. The monitoring system indicates both ionization rate and integrated ionization.

Unlike high energy x-rays and γ -rays, high energy electrons are easily scattered and absorbed and in both processes undergo changes of energy. Small changes in scattering foil thickness or in collimator design considerably affect the intensity, the energy and the shape of the electron beam.

The specific objectives of the work reported in this thesis are:

- I. To determine the dose per unit monitor reading at a point within an irradiated homogeneous material. The word point is used here with a special meaning. One ascribes to a point all of the events taking place within a small volume (e.g. a sphere or

a cylinder) centered at the point. The size of the volume is usually not specified, but is taken to be small enough so that the number of interactions per unit mass contained in the volume is independent of point to point variations in characteristics of both the radiation field and the material, but on the other hand, large enough to contain a sufficient number of interactions during the time interval of observation so that the measured quantity can be obtained with the desired precision.

The materials used for the determination of dose are tissue equivalent phantoms, homogeneous and well defined. They are chosen because their radiation properties are rather similar to those of most biological tissues. Their composition is defined and they are readily available. Phantoms have been used to determine a considerable body of information on the spatial distribution of absorbed dose such as central axis depth dose data and isodose curves.

II. To determine the dose at other points in the irradiated phantom relative to the point of measurement in I.

III. To study how the relative dose distribution in II changes with energy, field size, and nature of the collimating devices.

The standard electron beam dose distributions that are used in treatment planning for high energy electron beam radiotherapy are generally measured in homogeneous tissue equivalent phantoms.

The effect of tissue inhomogeneities on dose distribution such as bone, lung, chest wall and air cavities have been discussed and demonstrated by several authors (Pohlitz {5}, Dahler, et al. {6}, Boone et al. {7}, and Brenner, et al. {8}), and will not be discussed in this work.

Chapter II of this dissertation will discuss the methods reported in the literature to determine the dose. It will also give a literature review of the factors affecting the dose distribution for fast electrons. Chapter III will describe the materials and methods used in this thesis. The main experimental results will be presented in Chapter IV, and will be compared to practical and theoretical results obtained by other authors. Chapter V will present a discussion of the experimental work and Chapter VI the conclusions that may be drawn from it.

CHAPTER II

LITERATURE REVIEW

For the determination of absorbed dose there are many different methods in use already. These include calorimeters, chemical dosimeters, ionization chambers, photographic film dosimeters and solid state dosimeters.

A method is regarded as "absolute" if it can be used to measure radiation without the necessity of calibrating its response in a known field of radiation. A device that is absolute in this sense may not necessarily be highly accurate, it needs only be inherently accurate enough to give reliable results within specified accuracy limits, without the need of calibration. However, primary radiation standards are not only absolute but usually quite accurate as well.

At the present time there are three devices that are absolute and generally accepted as being accurate enough to serve as dosimetry standards: calorimeters, the Fricke chemical dosimeter and ionization chamber dosimeters. All other devices available at the present time require calibration in known radiation fields for the attainment of accuracies within acceptable limits.

A. Calorimeters

Calorimetry is a basic method for the determination of dose in a small volume of an irradiated medium.

If a small volume of the medium is thermally isolated from the remainder, the dose in that volume (the absorber) is given by

$$D_1 = \frac{dE}{dm} = \frac{dE_h}{dm} + \frac{dE_s}{dm} \dots\dots\dots(1)$$

where D_1 is the dose in the absorber material 1, of mass dm , dE is the energy imparted to the absorber by ionizing radiation, dE_h is the

energy appearing as heat and dE_s is the heat defect which may be positive or negative. An example of dE_s is the energy produced or absorbed in induced radiochemical reactions. If there is no change of state

$$\frac{dE_h}{dm} = C_p \times \Delta T \dots\dots\dots(2)$$

where C_p is the specific heat at constant pressure and ΔT the change in temperature.

The rise in temperature produced by one rad in a medium such as water may be calculated directly as follows

$$1 \text{ rad} = 100 \frac{\text{ergs}}{\text{gm}} = 10^{-5} \frac{\text{joules}}{\text{gm}} \dots\dots\dots(3)$$

$$1 \text{ calorie} = 4.18 \text{ joules}$$

$$\text{Hence } 1 \text{ rad} = 2.39 \times 10^{-6} \text{ calories} \dots\dots\dots(4)$$

If the medium is water the specific heat is 1 calorie/gm/°C so the rise in temperature is small and equal to $(2.39 \times 10^{-6})^\circ\text{C/rad}$. Although this rise in temperature is small, sensitive methods are available to make its measurement possible.

An absorbed dose calorimeter generally consists of

1. an "absorber": the thermally isolated mass in which the energy dissipated into heat is measured.
2. one or more surrounding jackets provided for thermal isolation of the "absorber" and for temperature control, and
3. surrounding vacuum chamber.

Polystyrene and carbon are recommended as the preferred material for construction of the absorber and its surroundings in order to keep the heat defect as small as possible.

(1) Adiabatic Calorimetry

By thermodynamic definition an object is in adiabatic

equilibrium if there is no net heat exchange with its environment. In principle, a situation whereby there is no net heat exchange can be achieved if the temperature of the object and its surroundings are equal all times. At best, the impossibility of perfect temperature control makes the adiabatic system only an approximation. The term adiabatic can be applied to any calorimetric system in which the rate of temperature change due to heat loss is considered negligible relative to the rate of temperature change to be measured. More precisely, however, the term is usually applied only to those systems in which the jacket is controlled manually or through automatic feedback, to match the temperature of the calorimeter absorber. Although many dose measurements employ adiabatic systems these are usually designed for measurements of high dose rates.

(2) Quasi-adiabatic Systems

In the quasi-adiabatic system, heat losses are reduced by constraining the temperature of the jacket so as to maintain a small heat loss at all times. However, with this arrangement the small heat losses can be evaluated from an observation of the time-temperature curve of the calorimeter absorber. In such systems the absorber is surrounded by its jacket. Constraint of the jacket temperature is made possible by means of a surrounding mantle, the temperature of which is controlled by an external mechanism. Such devices have been employed for the measurement of dose rates commonly encountered in radiological practice.

In a dose calorimeter an imbedded temperature transducer measures a quantity that is proportional to the temperature change in the absorber. The quantity measured may for example be the fractional resistance change in a thermistor or the change in the

electromotive force in a thermocouple. This quantity is also proportional to the heat dissipated per unit mass of "absorber". Ordinarily a comparative method of calibration is employed in which the proportionality constant is determined from a separate measurement when a known amount of electrical energy is dissipated in the "absorber". This method of calibration assumes that the measurement of heat input by electrical means is equivalent to its measurement in a radiation field, i.e. that systematic errors arising from temperature gradients in the calorimeter are negligible or equal in both cases and thus cancel each other.

The use of calorimeters for the determination of absorbed dose in photon beams has been fully discussed in ICRU Report 14 {9} and ICRU Report 17 {10}.

There have been fewer reports on the use of calorimeters to measure absorbed dose from high energy electron beams than those from x-ray beams discussed in ICRU Report 21 {11}.

To transfer the absorbed dose in calorimeters to other dosimeters it has been recommended by ICRU Report 14 {9} that an ionization chamber be used as the transfer instrument for photons owing to its high stability and precision. With high energy electron beams, however, there are other considerations. In general, the absorbed dose will be needed at a point where the electron energy is not the same as at the calorimeter absorber. Ionization chambers show a greater variation of sensitivity with electron energy than do other dosimeters.

Although field calorimeters have been developed, this method is still not generally available. Moreover problems such as heat defect due to the structure of the absorber, thermal

isolation, electrical calibration and structural inhomogeneities in the region occupied by the "absorber" may arise. Therefore, these dosimeters will not be further discussed or used.

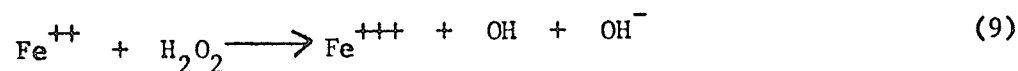
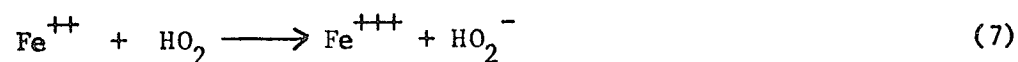
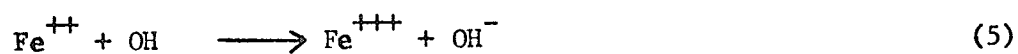
B. Chemical Dosimeters

One of the most common systems available for absolute dosimetry is the Fricke chemical dosimeter which was developed some 40 years ago (Fricke and Morse {12} and {13}). It is based on the oxidation of an aerated ferrous sulfate solution.

Some of the more pertinent references are listed in ICRU Reports 14 {9} , 17 {10} and 21 {11} as well as in H P A Report Series No. 4 {38} .

The basic principle for the determination of the dose by chemical method is that radiation produces chemical changes in aqueous solutions e.g. ferrous or ferrous ammonium sulfate by the oxidation of ferrous ion to ferric ion. The reactions oxidizing ferrous to ferric ion in a medium of sulfuric acid are quantitative and well established.

The simple reactions are



Detailed treatment of mechanisms of chemical reactions have been published by Fricke and Edwin {14} . The quantity of interest, namely the number of ferric ions produced can be determined by spectrophotometry by measuring the absorbance of the solution at the absorption

peaks at 224 nm or 304 nm for ferric ion.

Suppose a sample of ferrous sulfate is given a dose D_s rads and in the process a concentration ΔM molar of ferric ions are produced. The produced molarity ΔM has been found to be proportional to the dose D_s and the proportionality factor may be determined as follows:

$$D_s \text{ (rads)} = D_s \times 100 \frac{\text{ergs}}{\text{gm}} \dots\dots\dots(10)$$

$$1 \text{ ev} = 1.602 \times 10^{-12} \text{ ergs}$$

$$D_s \text{ (rads)} = \frac{D_s \times 100}{1.602 \times 10^{-12}} \frac{\text{ev}}{\text{gm}} \dots\dots\dots(11)$$

The number of molecules converted to ferric ions in the solution = $\Delta M \times \text{Avogadros constant}$

$$= \Delta M \times 6.023 \times 10^{23} \frac{\text{molecules}}{\text{litre}} \dots\dots\dots(12)$$

$$= \frac{\Delta M \times 6.023 \times 10^{20}}{\rho} \frac{\text{molecules}}{\text{gm}} \dots\dots\dots(13)$$

where ρ is the density of the solution in units of $\frac{\text{gm}}{\text{cm}^3}$.

The G-value is defined as the number of molecules produced per 100 ev of energy absorbed.

$$\text{Thus } G = \frac{\Delta M \times 6.023 \times 10^{20}}{\rho} \times \frac{1.602 \times 10^{-12}}{D_s} \frac{\text{molecules}}{100 \text{ ev}} \dots\dots\dots(14)$$

$$= \frac{\Delta M \times 9.65 \times 10^8}{\rho D_s} \frac{\text{molecules}}{100 \text{ ev}} \dots\dots\dots(15)$$

Hence the dose D_s can be obtained from equation (15).

$$D_s = \frac{\Delta M \times 9.65 \times 10^8}{\rho G} \text{ rads} \dots\dots\dots(16)$$

The ferric ions produced by radiation are determined spectrophotometrically by measuring the optical density of the irradiated solution by comparison with an unirradiated control.

The absorbance $\Delta(\text{OD})$ which is the difference in the optical density between the irradiated solution and unirradiated control is defined as

$$\Delta(\text{OD}) = \log \frac{I_o}{I_t} \dots\dots\dots(17)$$

where I_o is the intensity of the incident light and I_t that of the transmitted light.

The $\Delta(\text{OD})$ is proportional to the amount of ferric ions produced and the path length of the cell

$$\begin{aligned} \Delta(\text{OD}) &\propto \Delta M \times L \\ &= \text{constant} \times \Delta M \times L \dots\dots\dots(18) \end{aligned}$$

$$= \Delta E \times \Delta M \times L \dots\dots\dots(19)$$

where ΔM is the concentration expressed in mole/litre

L is the path length of the cell in cm

ΔE is the constant of proportionality and is called the molar extinction coefficient in units of $\text{l mol}^{-1} \text{cm}^{-1}$ (and it is also the difference in molar extinction coefficient between ferric and ferrous ions at the wave length used for the optical density measurements)

$$\Delta M = \frac{\Delta(\text{OD})}{\Delta E \times L} \dots\dots\dots(20)$$

From equation (16) and equation (20)

$$D_s = \frac{\Delta(\text{OD}) \times 9.65 \times 10^8}{\Delta E \times L \times \rho \times G} \text{ rads} \dots\dots\dots(21)$$

The above equation expresses the dose in the dosimeter solution.

The absorbed dose in water or water equivalent materials D_w and the absorbed dose in the solution D_s are related by an average value $s_{w,s}$ of the ratio of the mass stopping power $(S/\rho)_{\text{col}}$ between the water and solution.

This relation is expressed in the form

$$D_w = D_s \times \frac{(S/\rho)_{\text{col},w}}{(S/\rho)_{\text{col},s}} = D_s \times s_{w,s} \text{ rads} \dots\dots\dots(22)$$

The Fricke chemical dosimeter usually consists of 1 mmol/l ferrous or ferrous ammonium sulfate and 1 mmol/l sodium chloride in aerated 0.4 mol/l sulfuric acid.

The ferric ion has absorption maxima at wave lengths of 224 and 304 nm. The molar extinction coefficient is about twice as large at 224 nm as at 304 nm and the sensitivity of the ferrous sulfate dosimeter is thus approximately doubled when measurements are carried at 224 nm. However, impurities from containers are reported to be more troublesome at 224 than at 304 by Pettersen and Hettinger {23}.

A G-value of (15.5 ± 0.2) per 100 eV for ^{60}Co γ -rays has been recommended by ICRU Report 14 {9}. For electron beams a value for G of (15.7 ± 0.6) has been recommended by ICRU Report 21 {11} for energies in the range 1 to 30 MeV when 0.4 mol/l sulfuric acid is used in the dosimeter and a value of $(2205 \pm 3) \text{ l mol}^{-1} \text{ cm}^{-1}$ for the molar extinction coefficient at 25°C and 304 nm. This molar extinction value is based on a review of 83 determinations by Broszkiewicz and Bulhak {32}.

The temperature of the solution during irradiation will influence the G-value however, Petterson and Hettinger {23} recommend a correction coefficient of 0.15% per $^\circ\text{C}$ which is a mean value of temperature coefficients obtained by themselves and temperature coefficients published by several authors. Such correction factors can be applied if the temperature of the irradiation conditions differ from 25°C .

The molar extinction coefficient is more dependent on

temperature than the G-value.

If the temperature during optical measurements differ from 25°C the temperature coefficient is 0.7% per °C at 304 nm (ICRU Report 14 {9}).

Although most work with the Fricke dosimeter has been carried out with 0.4 mol/l sulfuric acid, a reduction in the acid concentration to 0.05 mol/l has been shown to reduce the G-value by a few percent (ICRU Report 14 {9}).

The response of the dosimeter, if initially in equilibrium with atmospheric oxygen, is linear with absorbed dose up to 35000 rads but thereafter the deficiency of oxygen results in a decreased production of HO_2 radicals and the G-value is lowered. Lack of linearity below 35000 rads points to the presence of impurities which may compete with ferrous ions for the radiation induced radicals. It is to counteract this effect that 1 mmol/l sodium chloride is used in the dosimeter solution.

Influence of impurities even in analytic-reagent-grade sulfuric acid has been noted by Davies and Law {33} who overcame their problem either by pre-irradiation of the acid or a suitable treatment with H_2O_2 .

For high energy electron beams it has been reported by ICRU Report 21 {11} that in experienced hand, measurements of absorbed doses of 10,000 rads show a standard deviation of the mean (S.D.M.) about 0.4% when glass irradiation vessels are used.

The cells in which the dosimeter solution is contained may influence the G-value. Erroneously high yields have occurred when different kinds of containers have been used and several methods of treating them in order to minimize these effects have been discussed

by ICRU Report 14 {9} and Fricke and Edwin {14}. These methods, however, appear to be useful only in limited circumstances and are not such as to ensure reproducibility when any container is used. Thus it is necessary for each investigator to find the most suitable irradiation vessel for his purpose.

The shape of the irradiation vessel is also of importance since it has been found that the surface to volume quotient of the irradiation cell may have an effect on the G-value (Svensson et al. {34}), which tends to increase with increase of this quotient.

Recommended values of $s_{w,s}$ for use in equation 22 for ^{60}Co γ -rays have been given by ICRU Report 17 {10} and for high energy electrons by Petterssen and Hettinger {23} and Svensson {27} Table 1.

TABLE 1

MASS STOPPING POWER RATIO BETWEEN WATER AND DOSIMETER SOLUTION

Radiation	$s_{w,s}$ (Fricke Solution)
^{60}Co γ -rays	1.003
10 - 35 MeV electrons	1.004

From equation 21 and 22 the absorbed dose in water or water equivalent phantom is

$$D_w = \frac{\Delta(\text{OD}) \times 9.65 \times 10^8}{\Delta E \times L \times \rho \times G} \times s_{w,s} \dots\dots\dots(23)$$

For 0.4 mol/l sulfuric acid, 1 mmol/l ferrous ammonium sulfate and 1 mmol NaCl

$$\Delta E = 2205 \text{ at } 25^\circ\text{C } 1 \text{ mol}^{-1} \text{cm}^{-1} \text{ (ICRU Report 21 {11})}$$

$\rho = 1.024 \text{ gm/cm}^3$ (Fricke and Edwin {14} and Petterssen and Hettinger {23})

$G = 15.5 \text{ molecules/100 ev at } 25^\circ\text{C for } ^{60}\text{Co } \gamma\text{-rays}$
(ICRU Report 14 {9} and ICRU Report 17 {10})

$G = 15.7 \text{ molecules/100 ev at } 25^\circ\text{C for high energy electron beams}$ (ICRU Report 21 {11})

$L = 1 \text{ cm}$

$s_{w,s} = 1.003$ for $^{60}\text{Co } \gamma\text{-rays}$

$s_{w,s} = 1.004$ for high energy electron beams 5 - 35 MeV

Inserting the temperature correction for both G-values and the molar extinction coefficient in equation (23) we get

(a) For $^{60}\text{Co } \gamma\text{-radiation}$

$$D_w = \frac{2.77 \times 10^4 \times \Delta(\text{OD})}{[1 + 0.007(t - 25)] \times [1 + 0.0015(t - 25)]} \dots\dots\dots(24)$$

(b) For high energy electron beam 5 - 35 MeV

$$D_w = \frac{2.73 \times 10^4 \times \Delta(\text{OD})}{[1 + 0.007(t' - 25)] \times [1 + 0.007(t - 25)]} \dots\dots\dots(25)$$

where t' is the temperature of measurement of absorbance and t is the temperature of irradiation.

C. Ionization Chamber Dosimetry

The absorbed dose D_w , in an irradiated medium can be determined from the absorbed dose D_g in a gaseous material by the Bragg-Grey relation:

$$D_w = D_g \times s_{w,g} \dots\dots\dots(26)$$

where $s_{w,g}$ is the average value of the ratio of the mass stopping power $(S/\rho)_{\text{col}}$ of the medium and the gas

$$s_{w,g} = \frac{(S/\rho)_{\text{col},w}}{(S/\rho)_{\text{col},g}} \dots\dots\dots(27)$$

$(S/\rho)_{\text{col}}$ has to be taken at the mean energy of the primary electrons at the point of interest.

However in practical cases the Bragg-Grey condition of undisturbed particle fluence is not met completely and a factor $P_{w,g}$ is introduced to correct for this:

$$D_w = D_g \times s_{w,g} \times P_{w,g} \dots\dots\dots(28)$$

$P_{w,g}$ is the perturbation correction factor to correct for disturbances to the fluence, which may be produced by the presence of the probe g , within the medium, w .

$P_{w,g}$ would be only unity for an infinitesimal cavity

$$D_w = \frac{\bar{W}}{e} \times J_g \times P_{w,g} \times s_{w,g} \dots\dots\dots(29)$$

J_g is the quotient of the ionization charge by the mass of the gas g

$\bar{W}$ is the average energy expended by electron in the gas per ion pair formed

e is the charge of the electron

$$D_w = \frac{Q_g}{\rho_g V_g} \times s_{w,g} \times P_{w,g} \times \frac{\bar{W}}{e} \dots\dots\dots(30)$$

Q_g is the charge produced in a volume V_g filled with a gas of density ρ_g

Air is the gas normally used in ionization chambers for electron or photon dosimetry.

The dose can be precisely determined using a suitable reference instrument such as an extrapolation chamber. Such chambers have been described in ICRU Report 21 {11} .

It is recommended that ionization chambers used as field

instruments in high energy electron beams should have small dimensions especially in the direction of the beam. Therefore flat chambers are most suitable for measurements of depth ionization curves or depth absorbed curves. However in most centers equipped with high energy accelerators an ionization chamber calibrated as an exposure meter for ^{60}Co or 2 MV x-rays is available.

Almond {35} and {36} , Wambersie et al. {24} , Kartha and Macdonald {25} , H P A Report Series No. 4 {38} , and ICRU Report 21 {11} have suggested that the dose at the point of measurements in a water phantom using an ionization chamber is given by

$$D_w = M \times N_c \times C_E \dots\dots\dots(31)$$

where D_w is the dose in water at the position of the chamber, when the chamber system is replaced by water and an identical monitor exposure is made

M is the instrument reading corrected for temperature and pressure.

N_c is the exposure calibration factor given by a standardizing laboratory for ^{60}Co γ -rays or 2 MV x-rays

C_E is the overall conversion factor to give the dose in water and it includes:

1. The correction factor for attenuation of photons in the chamber wall for ^{60}Co or 2 MV x-rays during exposure calibration.
2. The ratio of mass stopping powers for water and air which is valid for the mean energy of the primary electron at the point of measurement.
3. The perturbation correction factor.

Theoretical values of the overall conversion factor C_E

have been published in ICRU Report 21 {11} as a function of initial energies and depths (Table 6.2 page 43).

For any initial energy and any depth not specified in the table the overall conversion factor can be derived as follows:

For a gas filled probe in water the absorbed dose can be determined by equation (28).

$$D_w = D_g \times s_{w,g} \times P_{w,g}$$

$$D_g = J_g \times \frac{\bar{W}}{e} \dots\dots\dots(32)$$

$$\text{where } J_g = A x M x N_c \dots\dots\dots(33)$$

A is the attenuation factor in exposure calibration of the ionization chamber.

$$\text{Therefore } C_E = A \times s_{w,g} \times P_{w,g} \times \frac{\bar{W}}{e} \dots\dots\dots(34)$$

The factor $s_{w,g}$ entering this equation may be calculated from equation (27).

$$s_{w,g} = \frac{(S/\rho)_{col,w}}{(S/\rho)_{col,g}}$$

where $(S/\rho)_{col}$ has to be taken at the mean energy of the primary electrons $(E)_d$ at the point of interest (depth d) weighted by the distribution of the electron fluence in energy.

The mean energy of the electrons at the point of interest can be calculated by

$$(E)_d = E_o (1 - \frac{d}{R_p}) (\text{Harder } \{37\}) \dots\dots\dots(35)$$

$(E)_d$ - the mean electron energy of the spectrum in MeV

E_o - the initial incident energy in MeV

d - the depth of medium to point of interest in gm/cm^2

R_p is the practical range of the electron.

R_p can be calculated from a well established empirical

formula (ICRU Report 21 {11})

$$R_p = (0.52 E_0 - 0.4) \text{ gm/cm}^2 \dots\dots\dots(36)$$

The theoretical values of C_E given by ICRU Report 21 {11} apply to a flat extrapolation chamber with the air gap at right angles to the beam ($P_{w,g} = 1$). For cylindrical chambers used with their axis perpendicular to the direction of the incident electrons, values of $P_{w,g}$ the perturbation factor are published in ICRU Report 21 {11} .

Experimental values for C_E can be obtained directly by comparison of the dose determined by absolute dosimeters and the measurements obtained by the calibrated ionization chamber.

Intercomparison between the theoretical values of C_E published by ICRU Report 21 {11} has been carried out by plotting the C_E values as a function of the mean electron energy, instead of the initial energy and depth for 20,30, and 35 MeV initial electron energies, Fig. 1. It is seen that where more than one value of C_E is obtained by this means for any energy they differ by no more than 3%.

Theoretical and experimental values of C_E have been determined by a number of authors, HPA Report Series No. 4 {38} , Wambersie, et al. {24} , and Kartha and Macdonald {25} . These together with the values recommended by ICRU Report 21 {11} are plotted as a function of mean electron energy in Fig. 2.

The intercomparison shows that the C_E values of HPA {38} which are based on theoretical and experimental measurements differ from those given by ICRU by no more than 2% at any energy. Also the

experimental values of Wambersie, et al. {24} differ by no more than 3% from the values given by ICRU {11} . The experimental values of Kartha and Macdonald {25} however differ by as much as 6% from those recommended by ICRU {11} .

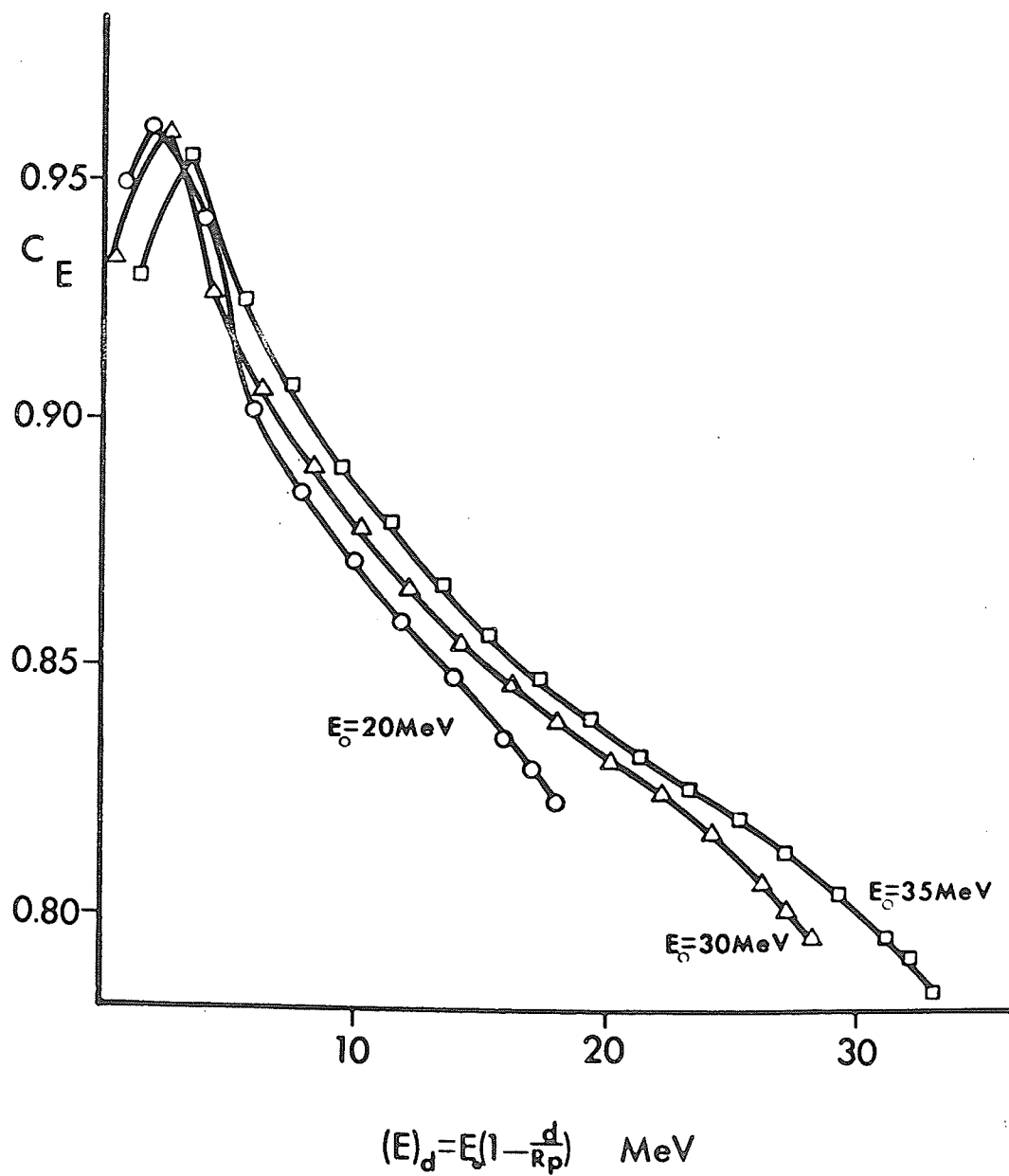


FIGURE 1

THEORETICAL C_E VALUES vs. MEAN ELECTRON ENERGY
FROM ICRU 21 (1972)

D. Solid State Dosimeters

There are now several solid state systems available for electron dosimetry. These dosimeters can be divided into two classes:

(1) Solid state integrating dosimeters (radiophotoluminescent glasses, thermoluminescent crystals, and glass and plastic optical density dosimeters).

(2) Solid state electrical conductivity dosimeters (semi conductor junction detectors, induced conductivity in insulating materials).

At the present time there is no solid state dosimeter that is absolute in the sense of not requiring calibration of response in a known field of radiation. The reason a solid state dosimeter must in general be calibrated in a known field is that there is not yet sufficiently good control, or in most cases, understanding of the important parameters that affect the radiation sensitivity of the material. These dosimeters therefore will not be further discussed or used.

E. Photographic Film Dosimeters

Photographic dosimetry is a convenient and rapid method of obtaining a set of relative isodose curves.

The sensitive material known as the emulsion consists of microscopic silver halide crystals (grains) dispersed in gelatine. The emulsion is generally coated as a thin layer on one or both sides of a cellulose acetate or polyester film, or sometimes on a glass plate. Radiation absorbed in an individual grain forms a "latent image" which facilitates the reduction of the grain to silver under the chemical action of development. Additional chemical stages are necessary to remove the undeveloped grains

and to make the pattern of developed grains permanent. Dutreix and Dutreix {39} and Hettinger and Svenssen {40} reported that isodose curves for electron beams in the energy range 10 - 24 MeV, measured by film, ferrous sulfate, or ionization dosimetry (corrected for density effect) showed good agreement. However, for high energies 25, 30 and 34.5 MeV, Hettinger and Svenssen {40} observed some discrepancy ($\pm 3\%$), the film giving higher relative values at a depth than the ferrous sulfate dosimeter.

The response curve of the film (density versus dose) should be measured before using a given type of film for measurement of isodose distributions. Many films show a linear relationship between density and absorbed dose up to a density of 2. Others show a curvature in the response curve. The linear relationship is not necessary, but it is more convenient since the optical density can be used directly to give the relative absorbed dose. During irradiation the film should be clamped in unit density phantom material leaving no gaps along the direction in which the electrons are incident. The film may be used in a paper envelope or unwrapped. In the latter case the phantom should be of opaque materials.

F. Phantoms

It has been recommended by ICRU Report 21 {11} that water can be used for ionization measurements and for films wrapped in evacuated waterproof envelopes or for bare film if the water is made opaque (Loevinger et al. {41}).

Slabs of solid materials are convenient for any kind of dosimeters. These should have a mass density and electron density close to those of water and low atomic number. Some suitable materials are commercially available. They allow the construction of phantoms of irregular shapes and the incorporation of inhomogeneities

if desired.

A dosimeter reading at a given depth in a phantom made of lucite say, may be used to determine the absorbed dose at an equivalent depth in water by multiplying by the ratio of mean mass stopping powers of water to dosimeter solution (Almond {35}).

G. Practical Factors Influencing The Dose Distribution For High Energy Electron Beams.

Published depth dose data show some differences (Schulz {43}). Electrons scattered by the collimator may contribute very substantially to the surface absorbed dose. This contribution may amount to 0.4 of the maximum absorbed dose (Loevinger et al. {41}). These scattered electrons are of lower energy than the primary electrons and enter the medium at relatively large angles and thus influence the depth dose curves. On the other hand, in some designs the inner surface of the collimator has been aligned in such a way as to use these electrons to square off the isodose curves by raising the absorbed dose at the periphery of the field at a certain depth in the patient.

The contribution of the electrons scattered by collimators appears to arise principally from the inner surface of the collimator. Alignment of the inner surface of the collimator with the source is usually not sufficient to prevent a number of electrons from striking this surface since the source is not a point, particularly in betatrons. The easiest way to reduce the amount of scatter is to reduce the thickness of the collimating device by using a material of high density (ICRU Report 21 {11}). An equivalent method is to place a ring of high density material at the aperture of a plastic collimator (Loevinger et al {41} , Dahler {44} , Svenssen and Hettinger {45}, and {46}). For adjustable diaphragms, metallic plates are more

convenient than plastic blocks (Robinson and McDougall {47}). The gain in beam purity by reduction of the scattered electron contribution is more important than the increase in bremsstrahlung, resulting from high energy electrons striking the high density material used for collimation. Bremsstrahlung can be kept to minimum by using aluminum or sandwiches of aluminum and steel or aluminum and brass.

The amplitude of the initial build up is not very dependent on energy in the range from 10 to 50 MeV. The depth dose curve and particularly its initial part are very dependent on the contribution of secondary electrons arising from the collimating device (Svenssen and Hettinger {46}). The actual absorbed dose at the surface of water or water equivalent medium usually is about 0.8 of the maximum in the absence of contamination of the beam.

The effect of field size on central axis depth dose values is usually negligible if the beam diameter is larger than the range, R_p , of the electrons for energies up to 20 MeV. Above this energy, values of depth dose are independent of field size for fields greater than 10 cm in diameter (ICRU Report 21 {11}). When the beam diameter is decreased below these values, it is observed that:

1. The maximum moves towards the entry surface.
2. The slope of the descending part of the depth absorbed dose curve is reduced.
3. The extrapolated range does not vary much but becomes difficult to specify when the field size gets very small.
4. The influence of electrons scattered by the collimating system usually becomes more and more important and depth dose curves vary more from one machine to another for narrow beams than for broad beam.

In electron beam treatment planning when the desired size treatment cone is not available, it is often necessary to reduce the field size by placing plates of absorbing materials in the path of the beam. Some institutes use metals such as lead, brass or steel, others use perspex and/or aluminum.

The absorbed dose and central axis percent depth dose for the particular field and energy are not easily available and additional measurements have to be made.

For x or γ -rays the absorbed dose and central axis depth dose can be frequently estimated by graphical interpolation from the relevant dose and percent depth dose curves applying the equivalent square technique (Batho et al. {48} and Clarkson {49}) but in electron dosimetry the equivalent square technique applicability has not yet been extensively examined or verified. F. Bagne {50} , found that the applicability of the equivalent square technique to electron dosimetry did not automatically follow. She reported that the equivalent square method can be applicable to rectangular cones as well as square cones in the energy range 10 - 45 MeV, however, the absorbed dose she found to be higher by as much as 10% for circular cones than for rectangular cones of equal equivalent square.

CHAPTER III

MATERIALS AND METHODS

Materials and methods used in this work.

A. Radiation Sources

(1) Electrons

The source of electrons was the Asklepitrone 35 MeV betatron manufactured by Brown Boveri Company and installed at the M.C.F.

The betatron is a circular accelerator in which electrons can be brought to a high level of energy.

The accelerating transformer is excited by means of an alternating current, fed to the primary windings from the main supply. An alternating magnetic field reacting around the central core is thus obtained. An evacuated, torroidal accelerating tube is mounted in the field, between the magnet poles of the accelerating transformer. Electrons are injected into the accelerating tube by means of an electron gun and are made to travel along the circular path by the guiding field between the magnet pole, the electrons gaining kinetic energy at each revolution because of the increasing magnetic flux through the orbit.

The central core and the magnet poles of the transformer are both excited by the same primary current.

The betatron is capable of producing either electron beams or x-rays up to a maximum energy of 34 MeV.

For electron radiation a large beam uniform to $\pm 10\%$ is obtained by scattering the primary beam through thin copper foils.

Different scattering foils are used, one for low energy levels (β_1 : 10 MeV to approximately 25 MeV) and one for high energy levels (β_2 : approximately 25 to 34 MeV).

For x-radiation the natural distribution of intensity, within a circular field is equalized by means of an absorber shaped

so that the resultant beam will have a uniform absorbed dose rate across its diameter. The betatron is provided by the manufacturer with collimators to define the electron beam at the beam exit by removable treatment collimators of various sizes.

For each field size, a separate treatment collimator is used consisting of:

(a) A diaphragm made of several layers of metals consisting of 18 mm Al, 10 mm Fe, 5 mm Pb and 5 mm Fe forming the base of the collimator called a sandwich.

(b) A lucite tube (localizer) fixed to the diaphragm base by screws.

Both components define the field size at the exit of the beam at a virtual distance from the point of exit from the accelerating tube equal to 110 cm.

These treatment collimators can be fitted into a master collimator at the beam exit window. This master collimator also has a slot for inserting the electron transmission ionization chamber, Fig. 3.

The electrons pass through the transmission chamber on their way out of the betatron. When the charge due to ionization accumulates to a given level a counter device connected to the transmission chamber is caused to trigger, and in the process recharges the ionization chamber. This counting device indicates the monitor reading by a mechanical register and each unit can be called a kick or a count. The number of kicks or counts registered on the counter is determined by the ionization delivered to the transmission ionization chamber.

(2) Cobalt-60 γ -ray source

The theratron "F" ⁶⁰Co teletherapy unit, manufactured by

Atomic Energy of Canada Limited installed at the M.C.F. was used during these experiments.

The head of the unit, which contains the radioactive source, is mounted at one end of a curved arm. Opposite the head is mounted a counterweight. The mounting is such that the head can rotate about a horizontal axis 75 cm from the source. The source was always directed towards this axis during these experiments.

The diaphragm system is fastened to the bottom of the head and is used to define the radiation beam to the desired size and shape. The cross-section of the irradiated field is illuminated when desired by the field size illuminating system. The field size is also indicated by a scale on the side of the diaphragm housing, and can vary from 3 x 3 cm to 20 x 20 cm at 75 cm source surface distance (SSD).

The amount of radiation delivered is controlled by a timer which indicates in minutes and seconds the elapsed time of exposure, and is connected to the shutter of the unit so that when a given time has elapsed the shutter is automatically closed.

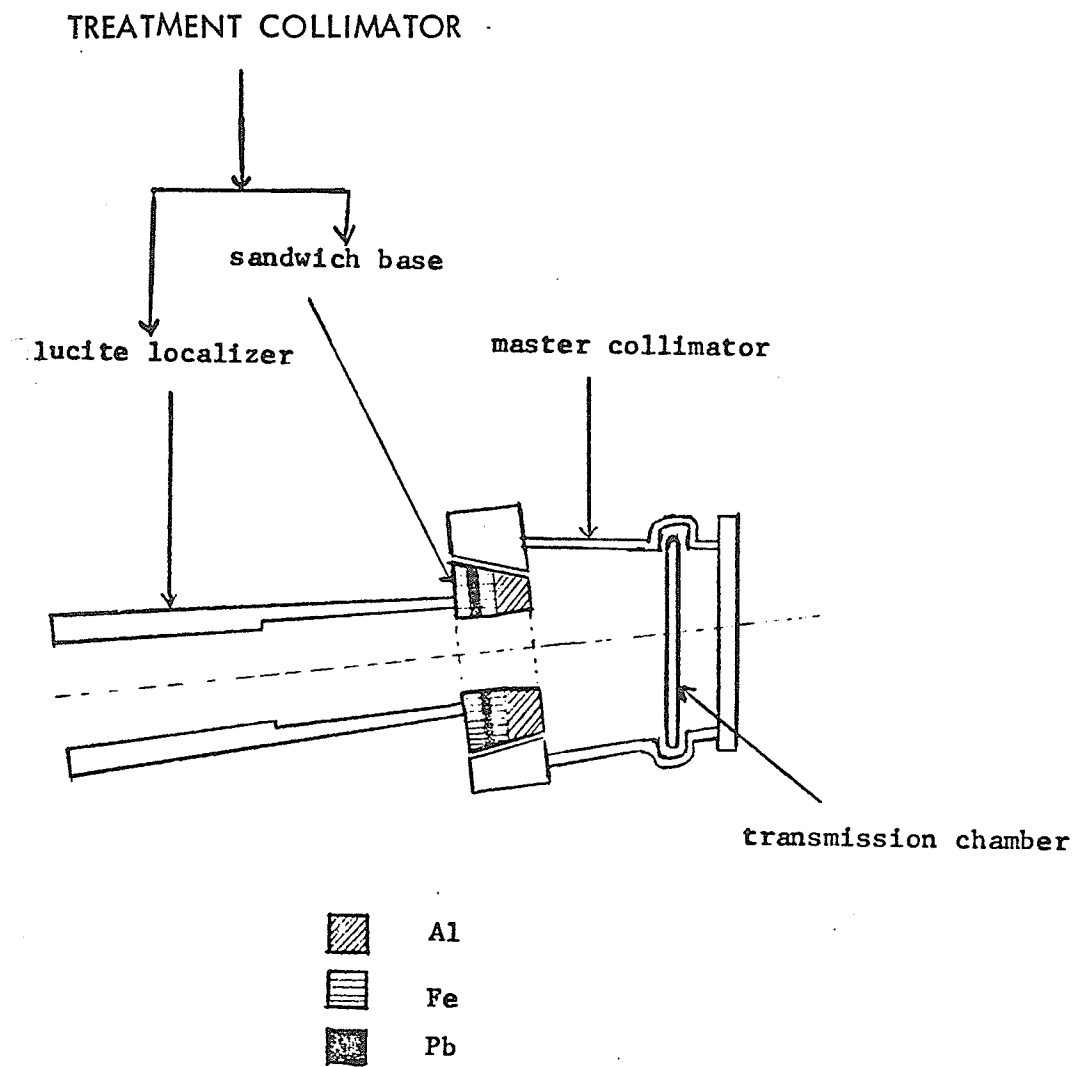


FIGURE 3

ELECTRON COLLIMATING SYSTEM (TREATMENT
AND MASTER COLLIMATORS)

B. Phantoms Used

For chemical dosimetry a layer of lucite 25 x 20 x 2.5 cm was divided into two parts, arranged such that the irradiation vial could be fitted in a central hole of the layer. Two similar holes on each side of the central hole were provided for additional dosimeter vials to check the symmetry of the beam along the central axis parallel to the layer and normal to the beam direction, Fig. 4 and Fig. 5.

For ionization measurements the lucite blocks used for chemical dosimetry were replaced by a single lucite block with the same overall dimension. It has a central hole just large enough to accept the ionization chamber. The block was placed so that the center of the ionization chamber was in the center of the beam. On each side of the central hole there is additional hole of the same dimension so that measurements could be made to check the symmetry of the beam along the central axis parallel to the layer and normal to the beam direction, Fig. 6. The dosimeter readings with the ionization chamber placed in the central hole were used for ionization dosimetry measurements.

Rods of the same material were used to fill unoccupied holes so that air gaps could be avoided. Other layers of lucite of any thickness could be added to the dosimeter layer so that measurements could be made at various depths in the phantom.

A tissue equivalent rubber phantom was used for film dosimetry. The phantom is composed of layers of opaque rubber 2 cm thick in which the film can be held by the rubber layers. The rubber layers are fixed in a bakelite box which has a cover designed to press the rubber layers firmly to avoid air gaps. The rubber excludes light so that unwrapped film would be used.

A water phantom 30 x 30 x 40 cm, was used for inter-comparison between the National Research Council of Canada (N.R.C.) chemical dosimetry system and M.C.F. chemical system. The inter-comparison was carried out for a 30 MeV electron beam at 3 cm depth in water and also for 32 MV photons at 10 cm depth.

IRRADIATION VIAL

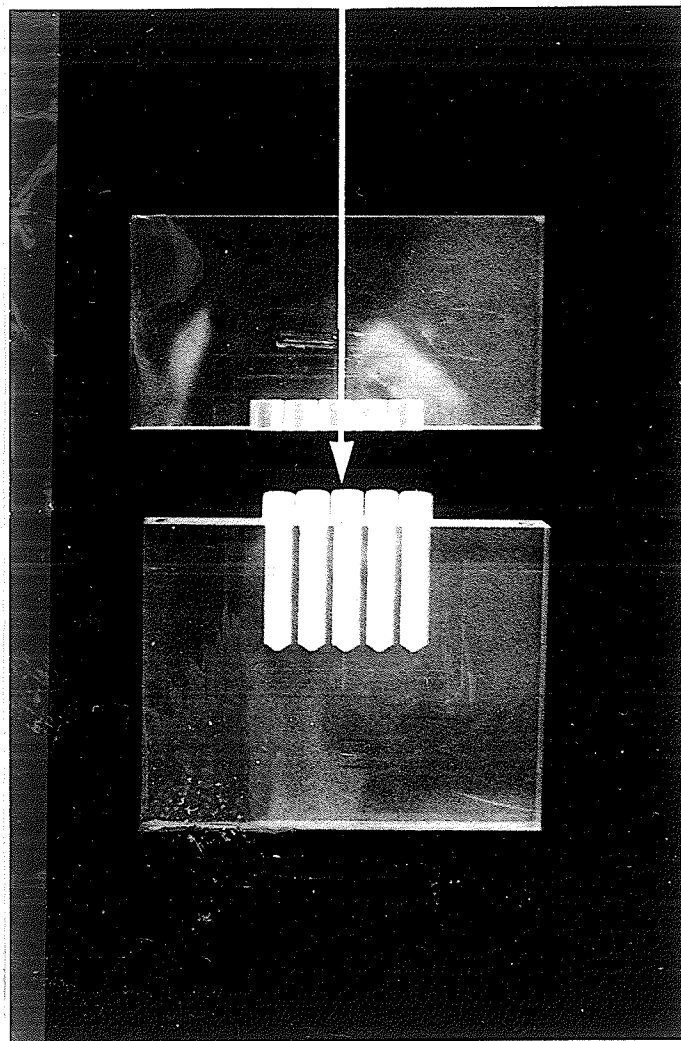


FIGURE 4

IRRADIATION VIALS LUCITE LAYER UNCLAMPED

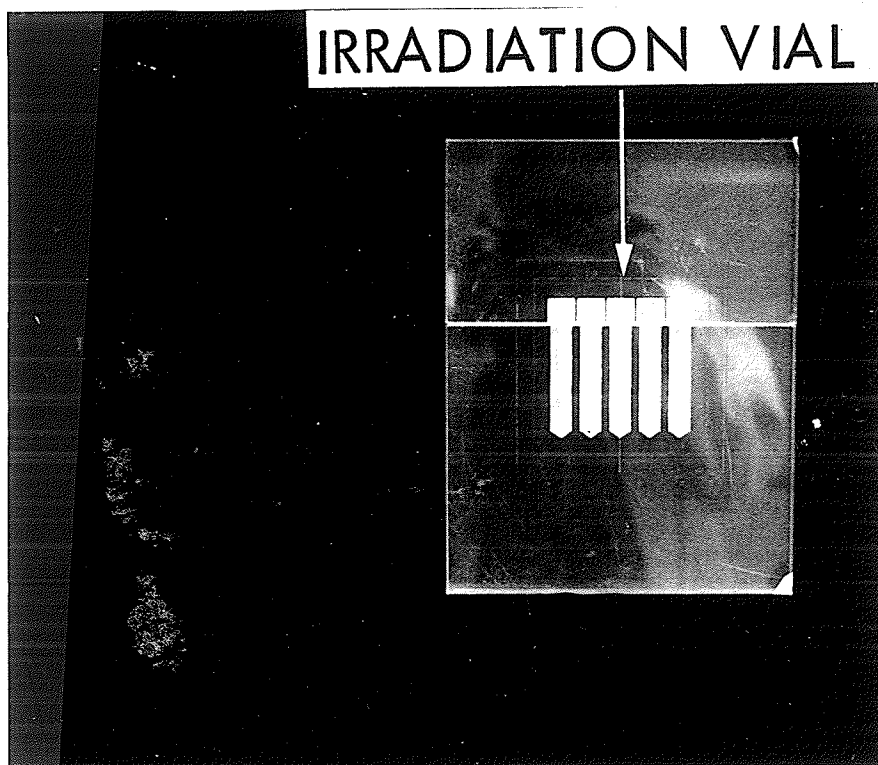


FIGURE 5

IRRADIATION VIALS LUCITE LAYER CLAMPED

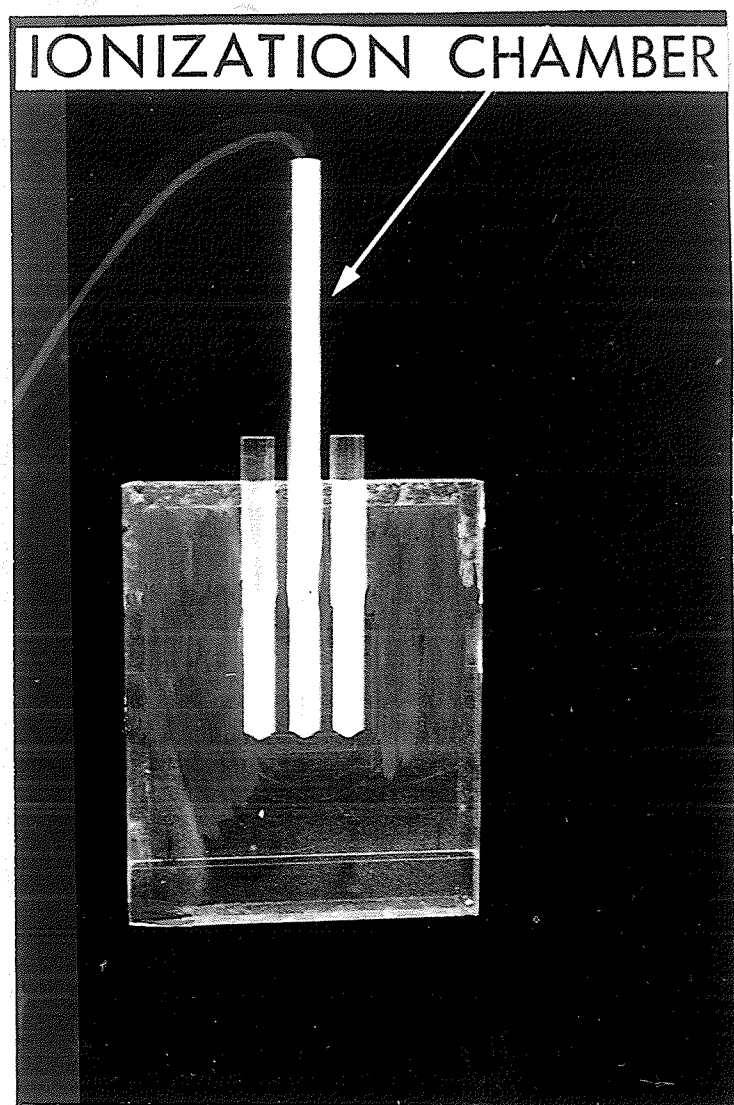


FIGURE 6

IONIZATION CHAMBER LUCITE LAYER

C. Chemical Dosimetry

The ferrous sulphate dosimeter can determine the absorbed dose absolutely and precisely at reference points along the central axis.

The dosimeter solution contains 0.4 mol/l sulfuric acid, 1 mmol/l ferrous ammonium sulphate and 1 mmol/l sodium chloride.

The equations used for the determination of absorbed dose in water or water equivalent phantom are (24) and (25) previously discussed in the literature review.

For ^{60}Co γ -rays

$$D_w = \frac{2.77 \times 10^4 \times \Delta(\text{OD})}{[1 + 0.007(t' - 25)] \times [1 + 0.0015(t - 25)]}$$

For electrons of energy from 10 to 34 MeV

$$D_w = \frac{2.73 \times 10^4 \times \Delta(\text{OD})}{[1 + 0.007(t' - 25)] \times [1 + 0.0015(t - 25)]}$$

After the ferrous sulphate solution was irradiated, the optical density of the solution was measured and used to determine the dose. The irradiation temperature was recorded and used to correct the G-value as indicated in the equations for D_w .

The instrument used to measure the optical density $\Delta(\text{OD})$ is a Beckman Model DU spectrophotometer with a Model 73600 power supply.

The hydrogen lamp was used for measurements in the ultraviolet range, the wavelength used for the determination of ferric ion content being 304 nm.

The cell compartment temperature during measurement of optical density was adjusted to $25 \pm 0.5^\circ\text{C}$ by circulation of water adjusted to the appropriate temperature.

For preparation of the dosimeter solution, pyrex containers,

double distilled water, Fisher SO-F-47 certified standard ferrous ammonium sulphate, Fisher A-300 reagent A.C.S. sulfuric acid, Baker analyzed reagent crystalline sodium chloride and falcon plastics disposable pipets type 7529 were used.

The irradiation was carried out using several types of cells of plastic or glass and filled with dosimetric solution, both with and without pre-treatment as mentioned in the literature review. The results obtained showed a variation in dose measurement ranging from 10 - 40%, but after further several trials with other types of containers for irradiation of Fricke solution, the author found that the most satisfactory containers were plastic test tubes manufactured by Falcon Plastics, Code No. 2054 made of polystyrene. These containers were subsequently used with no form of pre-treatment for all experiments.

A set of UV (Silica 170-2500 nm) rectangular measurement cells were used for optical density $\Delta(OD)$ measurements. Care was taken to arrange the cells in the same way each time they were used.

D. Experimental Determination Of C_E

The doses obtained by chemical dosimetry were used to determine the overall conversion factors (C_E) for an ionization chamber which has an exposure calibration factor (N_C) for Cobalt-60 γ -rays. Equation (31) was used for this determination of C_E

$$D_w = M \times N_C \times C_E$$

Hence

$$C_E = \frac{D_w}{M \times N_C} \dots\dots\dots(37)$$

E. Collimating Devices

The ideal collimating device should completely stop the electrons that hit it but should not contaminate the electron beam emerging through the collimator by scattered radiation of reduced energy. In practice this cannot be realized and the collimator will contaminate the electron beams by scattered radiation as reported in the literature review.

Experiments were performed to investigate the effect of electron scattering from different collimating devices on relative absorbed dose along the central axis, and its variation with field size and energy of the electron beam.

Measurements were performed using three different collimator systems for various field sizes.

(a) The treatment collimators supplied with the betatron were replaced by metal plates of lead which could be adjusted to various cross-sections. The lead plates were placed on the surface of the lucite phantom for the required field size in two layers, the first layer directly on the phantom surface and the second layer on the top of the first layer.

(b) The collimator base was fixed to the master collimator and the lead plates were placed on the surface of the phantom as in (a). The lucite localizers were removed in this arrangement.

(c) The original treatment collimator was fixed to the master collimator of the betatron without the presence of lead plates. The lucite tubes were brought into contact with the phantom as normally used in patient treatment.

The lead plates used in (a) and (b) above were sets of 4 similar plates 30 x 15 x 2.5 cm which could be adjusted to give

various field sizes. The lead material was chosen because of its density and availability.

F. Effect Of Replacement Of Localizer On Shape Of Isodose Distribution

Families of isodose curves were obtained using photographic film dosimetry for fields of circular cross-section to examine the effect of metal devices substituted for the lucite tube of the original collimator. This system was chosen in an attempt to reduce the number of low energy electrons scattered by the lucite localizer. The metal devices, Fig. 7, are made of brass ring, of about 2 cm thickness. This choice was based on experimental findings that the absorption of 2 cm brass is nearly equivalent to that obtained by the same thickness of lead for 34 MeV electron beam $\approx 96\%$. The density of brass (8.5 gm/cm^3) is less than the density of Pb (11.3 gm/cm^3) which could reduce the weight of collimator.

The brass rings were held in position by 3 aluminum rods fixed to an aluminum plate which was connected to the base sandwich by the same screws used for the lucite localizer. The brass rings were held to give the same virtual source to phantom distance of 110 cm. Such metal devices were made for circular field sizes of 4 cm and 8 cm diameter. The lucite tube and the brass device can be easily exchanged for measurements.

Kodak type M-54 films were used to obtain the isodose curves and developed by hand in liquid Kodak developer and fixer under standard conditions of temperature. Isodose curves were obtained from these films using an automatic isodose plotting system manufactured by Artronix Instrumentation Company, Saint Louis, MO 63108, Model 3304.

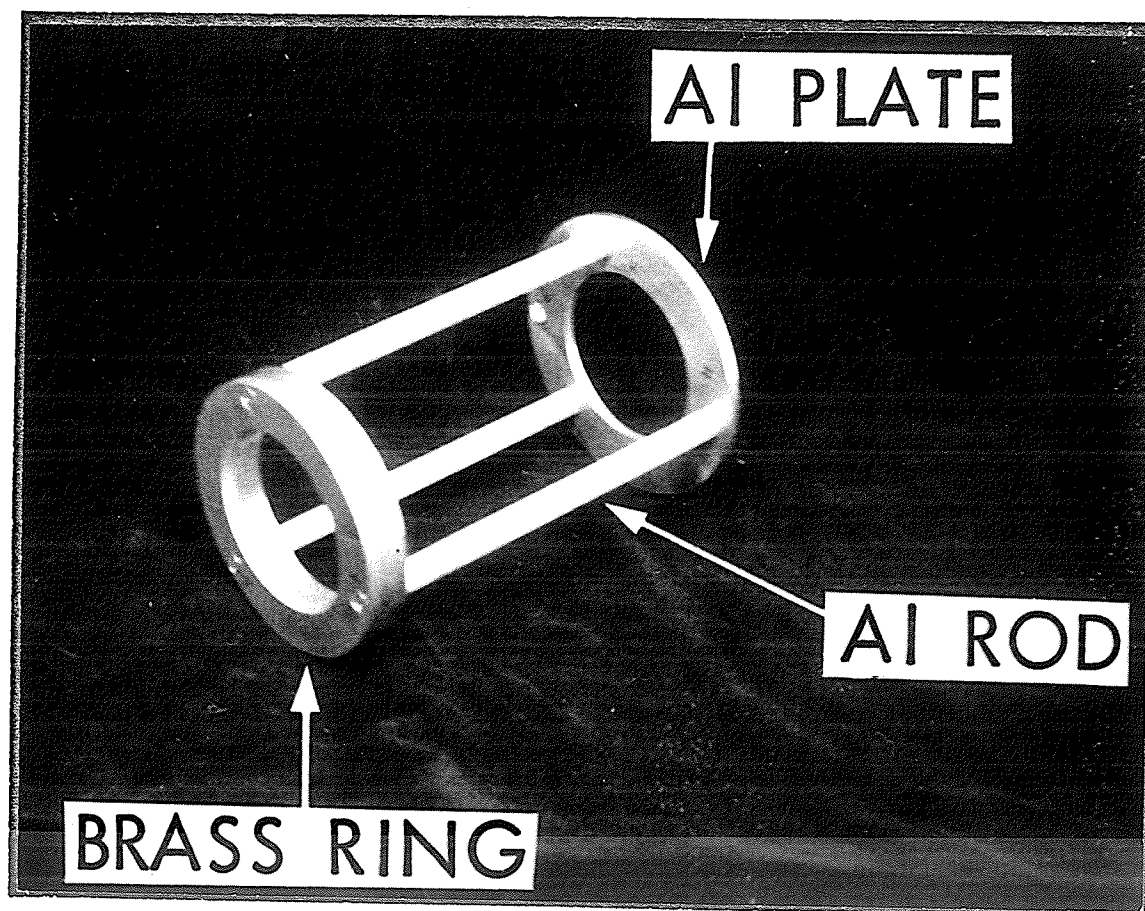


FIGURE 7

BRASS LOCALIZER

G. Experimental Arrangement

The betatron was adjusted horizontally for measurements with chemical dosimetry, calibration of the ionization chamber and for photographic film dosimetry. The betatron was adjusted vertically in experiments performed to examine the effect of electron scattering with different collimating devices. In all cases the phantoms were arranged so that the beam entered the surface normally and so that the virtual target distance was 110 cm.

CHAPTER IV

RESULTS

A. Determination Of The Overall Conversion Factor (C_E)

The ionization dosimeter consisted of a Baldwin Farmer ionization chamber no. 533702, connected to a Baldwin Ionex Instrument MK₃ No. 498105.

The exposure calibration factor(N_c)for this dosimeter was given as 1.05 by Dr. Cormack and Dr. Holloway (September 1973 - private communication) as a result of measurements carried out by them in which the response of this dosimeter was compared with that of a calibrated system, irradiated under the same conditions in a ^{60}Co beam. This calibrated system, which used the same ionization chamber in conjunction with a Townsend balance system had previously been calibrated at N.R.C. in Ottawa, Report number PXNR-2316, 1972.

(1) C_E for cobalt-60 radiation

C_E for ^{60}Co γ -rays is the ratio of the dose to the exposure at a particular point in an irradiated phantom assuming electronic equilibrium. To derive these two quantities a chemical and a calibrated ion chamber dosimeter were used respectively.

The dosimeter was placed at depth 4.44 gm/cm² in the lucite phantom. The source to phantom distance (SSD) was 75 cm and the field size was 20 x 20 cm on the surface of the phantom. The central axis of the beam was normal to the surface of the phantom and directed towards the center of the irradiation vial or the ionization chamber.

The measurements were performed for ionization and chemical dosimetry in succession by replacing the ionization chamber layer in the phantom by the chemical layer. In order to determine

that the dosimeter could be replaced in the same position the measurements were performed in the following order:

Ionization dosimetry, chemical dosimetry, ionization dosimetry and chemical dosimetry.

In each replacement care was taken to be sure that the center of the beam was directed to the center of the ionization chamber, or the irradiation vial, and perpendicular to the phantom surface at a source phantom distance (SSD) 75 cm. The difference between the maximum and minimum readings of ionization or optical density after replacement did not exceed 1%, and it was concluded that the consistency was satisfactory.

The optical density $\Delta(OD)$ of the irradiated solution was measured and the dose D_w in rads was calculated using equation (24). The overall conversion factor for ^{60}Co γ -rays to convert the exposure (MxN_c) to dose in rads is analogous to that used for high energy electron beam and here has the same abbreviation C_E . This conversion factor could be determined using equation (37).

The results of chemical dosimetry for the determination of absorbed dose D_w in rads are shown in table 2. Table 3 shows the result of ionization dosimetry. In this table the dose rate, D_w per minute, from table 2 was used to determine the overall conversion factor (C_E).

(2) C_E for high energy electron beams 10 - 34 MeV

The measurements were performed with the same lucite phantom used for ^{60}Co γ -rays measurements. The depth of measurement in the phantom was 4.44 gm/cm^2 and the virtual source to phantom distance was 110 cm. The field size was 12 x 12 cm defined by the

collimator supplied by the manufacturer. For both chemical and ionization dosimetry the center of the electron beam was directed to the center of the vial or the ionization chamber and normal to the phantom surface as for ^{60}Co γ -rays. The measurements were performed on several days. Each day the measurements were carried out for both ionization and chemical dosimetry by replacing the ionization chamber layer of the lucite phantom by the irradiation vial layer as for ^{60}Co γ -rays. The betatron tends to heat up during use causing uncertain heating effects on the transmission chamber monitor. The effect of heat and phantom replacement on the measurements was checked by exposing the ionization chamber first, then the chemical dosimeter and so on alternating from one to the other.

In each replacement care was taken to be sure that the center of the beam was directed to the center of the ionization chamber or the irradiation vial and perpendicular to the phantom surface, and that the end of the lucite localizer was in contact with the surface of the phantom at a virtual distance from the electron source of 110 cm. The differences between the maximum and minimum readings of ion chamber or optical density after replacement and from day to day were within 2%. Therefore the consistency of the measurements was satisfactory.

The optical density $\Delta(\text{OD})$ of the irradiated Fricke solution was measured and the dose D_w in rads was obtained using equation (25). These results are shown in table 4. Table 5 shows the results of the ionization chamber dosimetry. In this table the dose results D_w per count from table 4 were used to determine the overall conversion factor (C_E) using equation (37).

B. Intercomparison Between M.C.F. C_E Values and ICRU C_E Values

Table 6 shows the intercomparison between the M.C.F. experimental result for C_E and the value given by ICRU Report 14 {9}, HPA Report Series No. 4 {38} and ICRU Report 23 {51} for ^{60}Co γ -rays radiation. Table 7 shows the intercomparison between experimentally determined values of C_E and values derived from ICRU Report 21 {11} for high energy electron beams. The mean electron energy $(E)_d$, at the point of measurements was calculated using equation (35). This intercomparison is also shown in Fig. 8.

C. Intercomparison Between Dose Determined by N.R.C. and M.C.F.

Fricke dosimeters which were supplied by N.R.C. consisted of sealed quartz vials. The vials were disc shaped about 1 cm thick and 3 cm in diameter, they were irradiated with the beam normal to the flat face and returned unopened to Ottawa where the absorbance measurements were performed by Dr. R  al Levesque of the N.R.C. The M.C.F. vials were Falcon plastic test tubes filled with dosimeter solution prepared as described in materials and methods.

Each vial was irradiated in a water phantom at 3 cm depth to the center of the vial for 30 MeV electrons or 10 cm depth for 32 MV x-rays. For each modality two vials from N.R.C. and one from M.C.F. were irradiated one at a time. Two additional vials of each type were left unirradiated to serve as unirradiated controls. The exposure time particularly for x-rays was long (approximately 100 minutes or more) and the betatron became warmer as time went on. Because of this the M.C.F. vials were exposed once for each energy and the Fricke dosimeter irradiations were preceded and followed by an ionization measurement.

Ion chamber measurements were made by the Baldwin Farmer ionization chamber in conjunction with the Townsend balance system at 3 cm depth for 30 MeV electron beams and 10 cm depth for 32 MV x-ray beams in the water phantom.

The reproducibility during replacement for ion chamber measurements for both electrons and x-ray beams was satisfactory since it was found that the difference between the maximum and minimum readings did not exceed 2% (private communication with Dr. Cormack).

Each vial from the N.R.C. or the M.C.F. filled with the chemical solution was irradiated for 2500 counts for both electron beams and x-ray beams. The ionization system was exposed each time for 50 counts for both electrons and x-rays.

The dose was obtained from the ionization measurements by applying the N.R.C. exposure calibration factor for ^{60}Co γ -rays and the C_E values published by ICRU Reports {9} and {11}. The dose was also obtained using the M.C.F. chemical system. The intercomparison is summarized in table 8. This table also includes values of dose given by Dr. Levesque for the N.R.C. measurements.

TABLE 2
FERROUS SULFATE DOSIMETRY RESULTS
FOR ^{60}Co γ -RAYS

Time	Optical Density $\Delta(\text{OD})$	D_w rads	D_w rads/minute	Mean D_w rads/minute	(1) S.D.M.
55 min	0.1875	5201.6	94.57	94.14	$\pm 0.4\%$
45 min	0.1520	4216.7	93.71		

TABLE 3
IONIZATION DOSIMETRY RESULTS
FOR ^{60}Co γ -RAYS

Time	No. of Readings	Mean value of M	(1) S.D.M.	N_c	Mean D_w rads/min (11)	$C_E = \frac{D_w}{M \times N_c}$
1 min	7	93.97	$\pm 0.1\%$	1.05	94.14	0.954

(1) The S.D.M. of the optical density or ionization measurements were obtained by using the Health Sciences Computer System programme ST41 (one-way analysis of variance).

(11) Values obtained from Table 2.

TABLE 4

FERROUS SULFATE DOSIMETRY RESULTS FOR ELECTRONS

Energy E_o MeV	No. of Monitor Counts	No. of Experiments	Mean Value of $\Delta(OD)$	S.D.M.	D_w rads	D_w rads/ count
34 β_2	2997	10	0.1378	$\pm 0.2\%$	3767	1.257
30 β_2	2997	4	0.1300	$\pm 0.5\%$	3555	1.186
25 β_2	2997	4	0.1205	$\pm 0.5\%$	3296	1.100
20 β_1	2997	2	0.1260	$< \pm 0.1\%$	3446	1.150
15 β_1	2997	2	0.1185	$\pm 0.4\%$	3241	1.081

TABLE 5

ION CHAMBER DOSIMETRY RESULTS FOR ELECTRONS

Energy E_0 MeV	No. of Monitor Counts	No. of Readings	Mean Value of M	S.D.M.	Mean Value of M/count	N c	D_w rads/ count (1)	$C_E = \frac{D_w}{M \times N_c}$
34 β_2	60	75	89.51	$\pm 0.1\%$	1.492	1.05	1.257	0.802
30 β_2	60	25	82.51	$\pm 0.1\%$	1.375	1.05	1.186	0.821
25 β_2	60	30	76.53	$\pm 0.1\%$	1.276	1.05	1.100	0.821
20 β_1	60	10	78.01	$\pm 0.1\%$	1.300	1.05	1.150	0.842
15 β_1	60	10	70.20	$\pm 0.1\%$	1.170	1.05	1.081	0.880

(1) Values obtained from Table 4.

TABLE 6

COMPARISON BETWEEN M.C.F. AND ICRU C_E VALUE
FOR ^{60}Co γ -RAYS

Type of Radiation Source	Depth in the Phantom gm/cm^2	M.C.F. C_E Value	ICRU C_E Value
^{60}Co γ -rays	4.44	0.954	0.95

TABLE 7

COMPARISON BETWEEN M.C.F. AND ICRU C_E VALUES
FOR FAST ELECTRONS

Incident Electron Energy E_0 MeV	Depth in the Phantom gm/cm^2	Mean Electron Energy $(E)_d$ MeV	M.C.F. C_E Values	ICRU 21 C_E Values
34 β_2	4.44	25.3	0.802	0.817
30 β_2	4.44	21.2	0.821	0.827
25 β_2	4.44	16.2	0.821	0.843
20 β_1	4.44	11.1	0.842	0.864
15 β_1	4.44	6.0	0.880	0.892

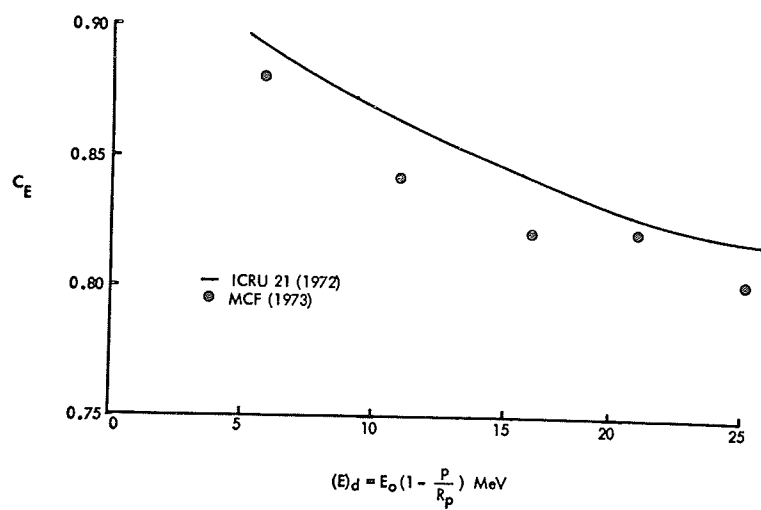


FIGURE 8

M.C.F. AND ICRU C_E VALUES vs. MEAN ELECTRON ENERGY

TABLE 8

DOSE INTERCOMPARISON BETWEEN M.C.F. AND N.R.C.

Energy	Depth in the Phantom gm/cm ²	N.R.C. Dose rads/count chemical system	M.C.F. Dose rads/count chemical system	M.C.F. Dose rads/count Using ICRU C _E values Ionization dosimetry
30 MeV Electrons Beam	3	1.17	1.19	1.21
32 MV X-Rays Beam	10	0.91	0.92	0.935

D. Calibration With Different Collimating Systems

The effect of different collimating systems on the relative rads per monitor reading (rr/mr) at points 2.6 and 4.8 gm/cm^2 depth on the central axis was investigated.

Measurements were made with 10, 20 and 34 MeV electron beams, using Baldwin Farmer ionization chamber no. 398408 connected to the Baldwin MK₃ Instrument No. 498105. The measurements were performed at depths 2.6 and 4.8 gm/cm^2 in the lucite phantom. The virtual source to phantom distance was adjusted to 110 cm. The electron beam was always directed to the center of the sensitive volume of the ion chamber placed in the central hole normal to phantom surface. One or two readings were recorded for each field size.

Ionization measurements were recorded for various field sizes from 4 x 4 cm to 14 x 14 cm for rectangular and square fields, and 4 cm to 8 cm diameter for circular fields.

The rr/mr was obtained from the ionization measurements by multiplying the readings with C_E factors derived from the values given by ICRU Report 21 {11}.

For circular field sizes in arrangements (a) and (b) described in the "materials and methods chapter" brass rings were placed on the top of the second layer of lead for measurements at depth 2.6 gm/cm^2 and lead rings were placed on the surface of the first layer of lead next to the phantom for measurements at depth 4.8 gm/cm^2 in the phantom.

When the collimation was by means of lead plates on the surface of the phantom (arrangement (a)) with no base sandwich it was found (Fig. 9 and 10) that the rr/mr was independent of field

size for both depths when 10 MeV and 20 MeV electrons were used except for the smallest field sizes used (4 x 4 cm and 4 cm diameter). With 34 MeV electrons, field size had no effect on rr/mr at 4.8 gm/cm² depth but at shallower depth it was noticed that as the area decreased from 60 cm² the rr/mr increased by about 7% for a field size of area 12.6 cm².

When the base sandwich was used together with the lead defining plates (arrangement (b)) it was found (Fig. 11 and 12) that the calibration (rr/mr) was independent of area for the lower energies above 60 cm². For smaller area the value of rr/mr decreased until for about 13 cm² area it was reduced by about 40% at 10 MeV and 15% at 20 MeV. At 34 MeV the value of rr/mr is independent of area except possibly for the smallest field size at 4.8 gm/cm² depth.

When the original collimator consisting of the lucite localizer and base sandwich (arrangement (c)) was used results in (Fig. 13 and 14) similar to those with (arrangement (b)) was obtained except that the value of rr/mr was slightly higher for all field sizes presumably due to radiation scattered off the lucite localizer at 10 and 20 MeV. With 34 MeV electrons the rr/mr dependence of field size was approximately similar to that obtained with (arrangement (a)).

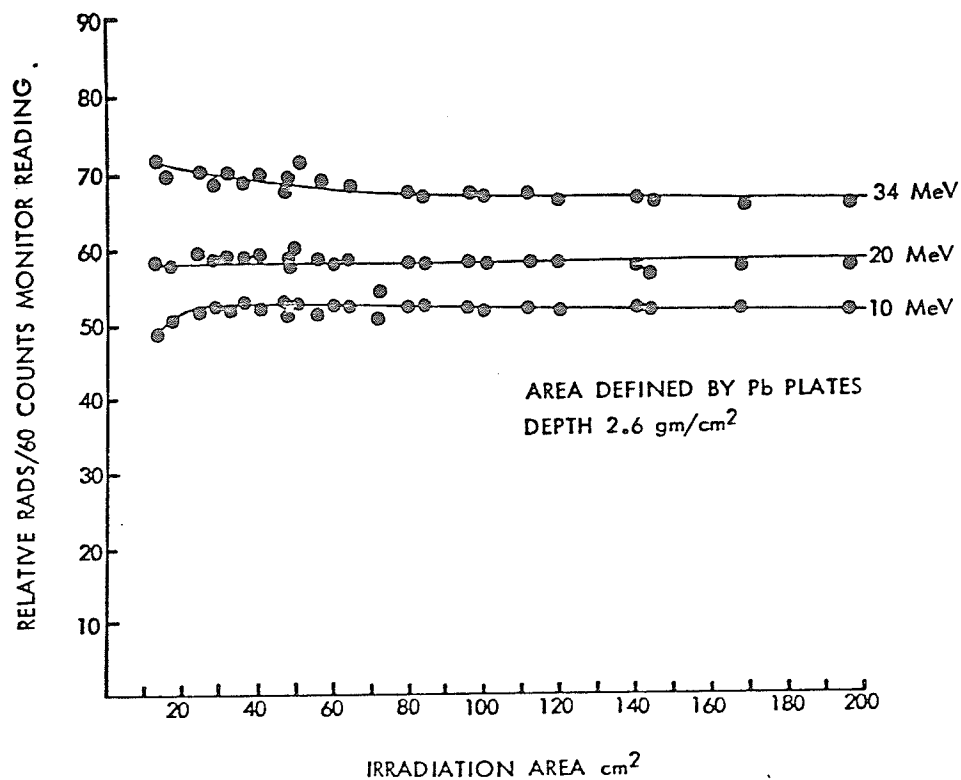


FIGURE 9

rr/mr vs. IRRADIATION AREA DEFINED BY LEAD PLATES
MEASURED AT 2.6 gm/cm^2 DEPTH IN A LUCITE PHANTOM

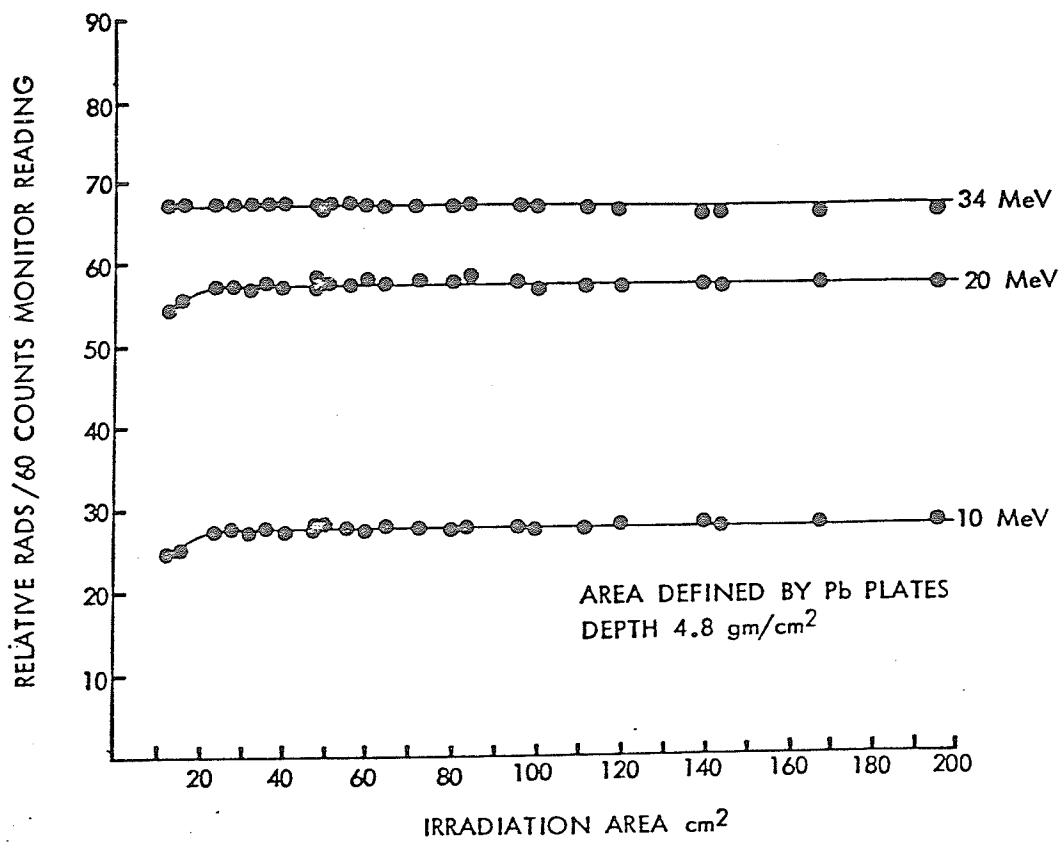


FIGURE 10

rr/mr vs. IRRADIATION AREA DEFINED BY LEAD PLATES
MEASURED AT 4.8 gm/cm^2 DEPTH IN A LUCITE PHANTOM

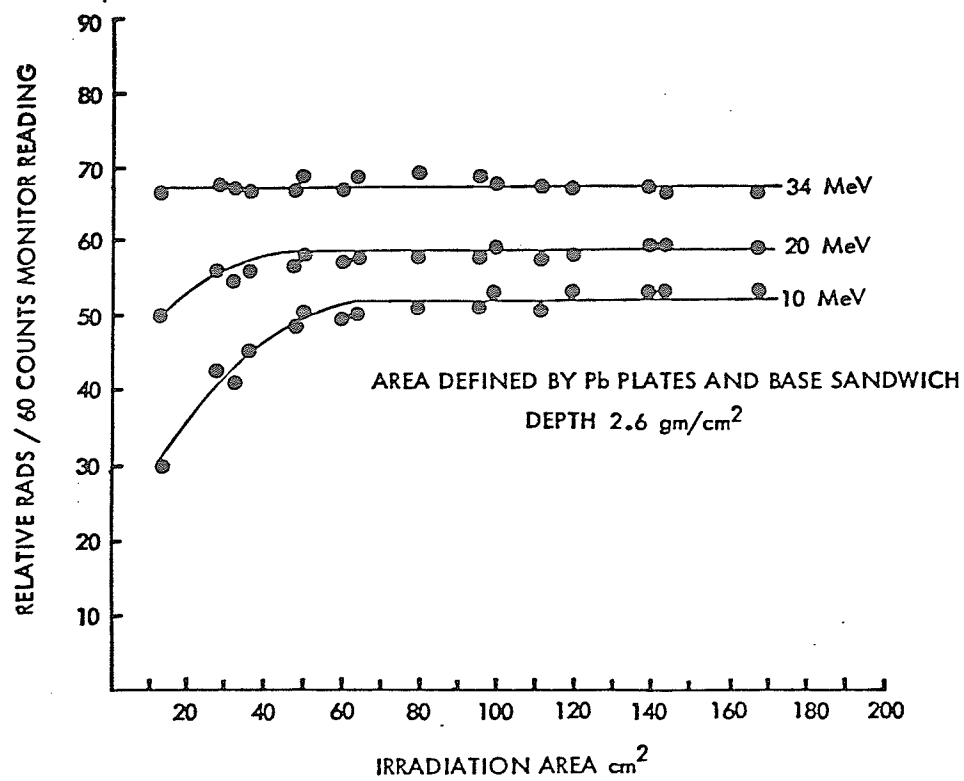


FIGURE 11

rr/mr vs. IRRADIATION AREA DEFINED BY LEAD PLATES
AND BASE SANDWICH MEASURED AT 2.6 gm/cm^2 DEPTH
IN A LUCITE PHANTOM

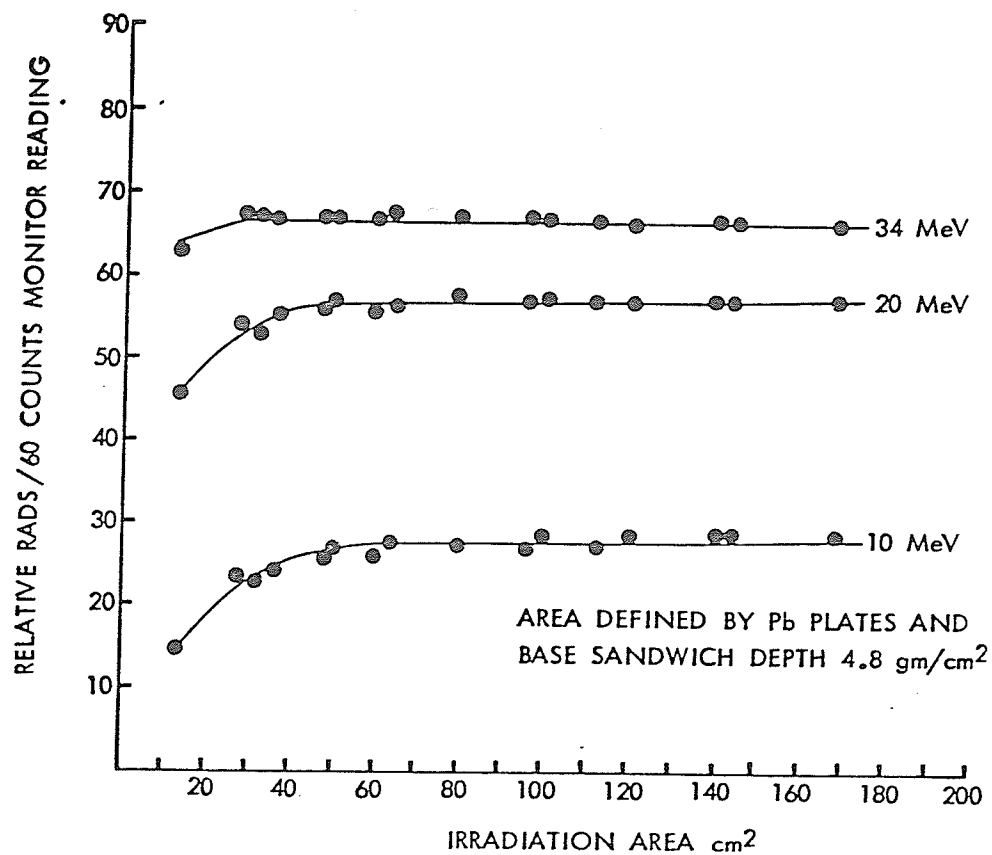


FIGURE 12

rr/mr vs. IRRADIATION ARE DEFINED BY LEAD PLATES
AND BASE SANDWICH MEASURED AT DEPTH 4.8 gm/cm^2
IN A LUCITE PHANTOM

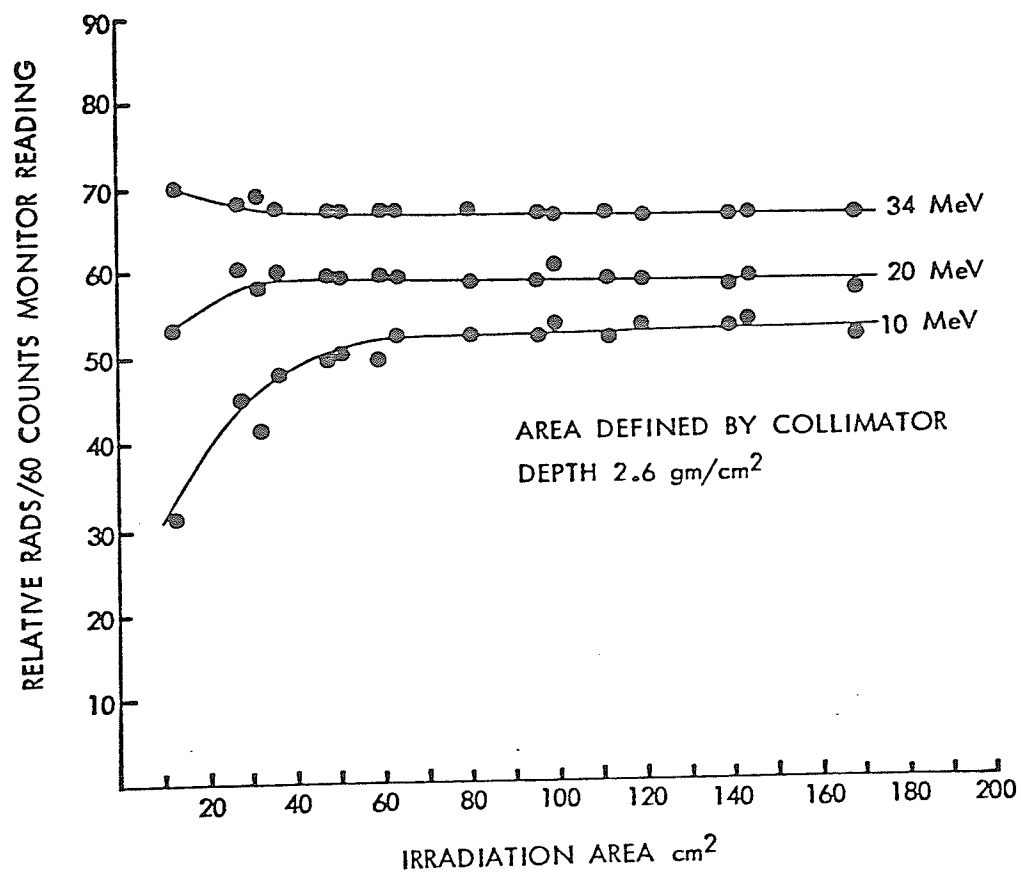


FIGURE 13

rr/mr vs. IRRADIATION AREA DEFINED BY ORIGINAL
COLLIMATOR MEASURED AT 2.6 gm/cm^2 DEPTH IN A
LUCITE PHANTOM

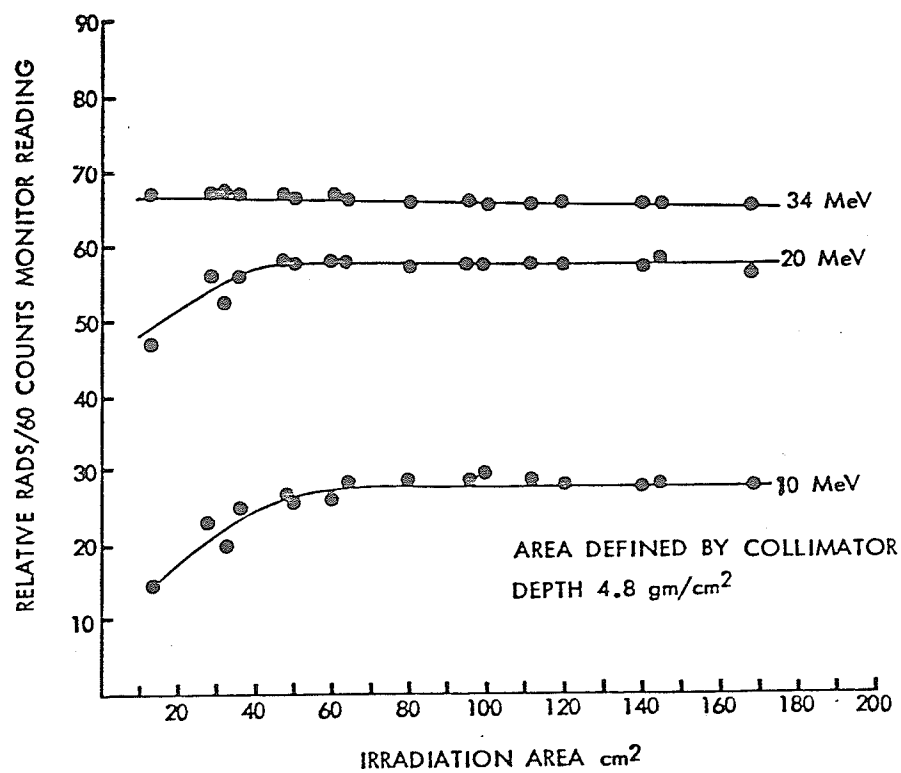


FIGURE 14

rr/mr vs. IRRADIATION AREA DEFINED BY
ORIGINAL COLLIMATOR MEASURED AT
 4.8 gm/cm^2 DEPTH IN A LUCITE PHANTOM

E. Influence Of Collimator Design On The Shape Of The Isodose Distribution.

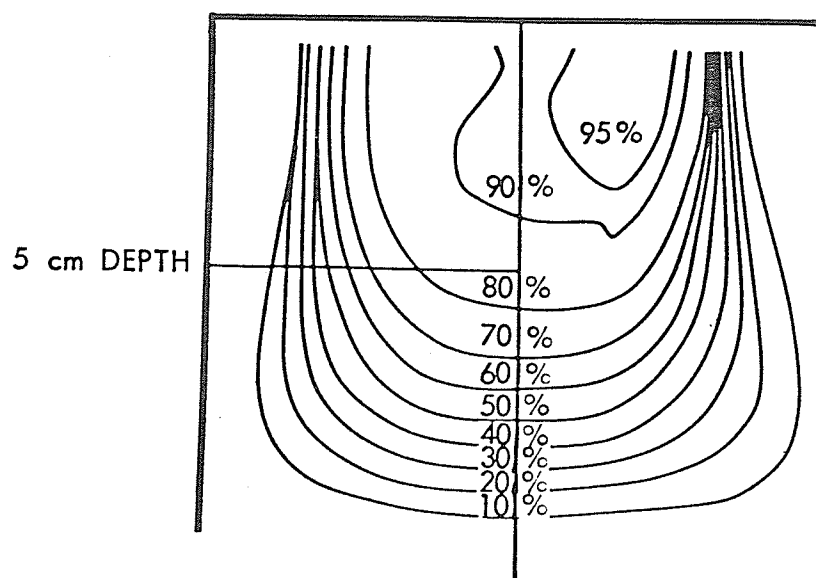
In section D. of this chapter results were presented which showed the influence of the collimating system on the dose delivered to two points on the central axis. In this section, the effect of the collimating system on the distribution of radiation throughout the irradiated volume will be investigated. The method chosen for this purpose makes use of photographic film in the rubber phantom. The electron beam was directed horizontally in such a way as to hit the surface of the phantom normally and parallel to the plane of the film. The edge of the film was placed in contact with the central diameter of the lucite or the brass localizer described in Chapter III on materials and methods.

The virtual source to phantom distance was 110 cm. Each film was irradiated by the electron beam for an exposure of 14 counts on the monitor.

Fig. 15 and Fig. 16 show the comparison of isodose curves obtained for field size 8 cm circular using a 20 MeV electron beam for lucite and brass collimator respectively.

Fig. 17 and Fig. 18 show the isodose curves obtained for an electron beam of energy 34 MeV.

When one examines the results shown in Fig. 15 to Fig. 18, it is evident that with the brass localizer the penumbra region near the surface of the phantom is somewhat reduced while the symmetry of the distribution has improved compared with similar distribution for the lucite localizer. Moreover the metal localizer increased the percentage depth dose by about 5%. The measurements were repeated using 4 cm circular field and similar results were obtained.



8 ϕ cm ISODOSE DISTRIBUTION
20 MeV B₁

LUCITE LOCALIZER

FIGURE 15

8 ϕ cm ISODOSE DISTRIBUTION , 20 MeV B₁, LUCITE
LOCALIZER MEASURED IN A RUBBER PHANTOM USING
BARE X-RAY FILM

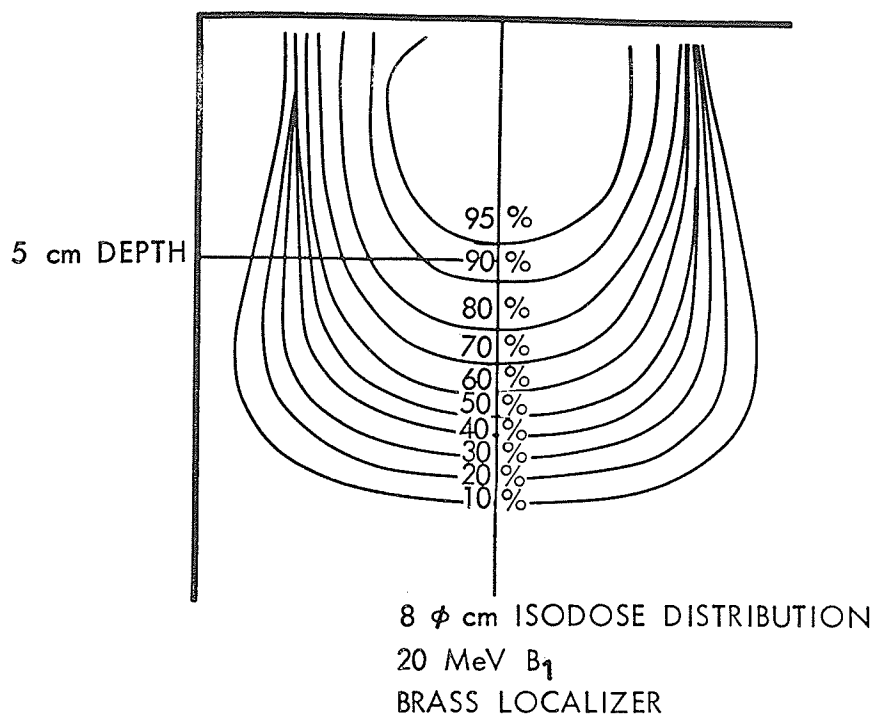


FIGURE 16

8 ϕ cm ISODOSE DISTRIBUTION , 20 MeV B₁ , BRASS
LOCALIZER MEASURED IN A RUBBER PHANTOM USING
BARE X-RAY FILM

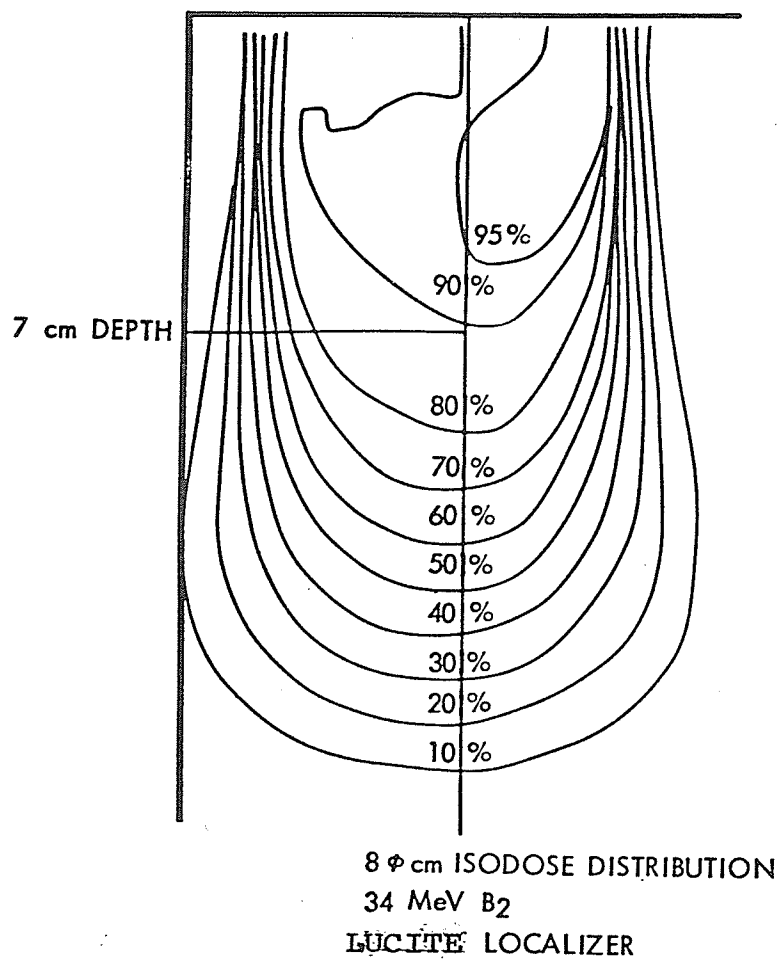


FIGURE 17

8 ϕ cm ISODOSE DISTRIBUTION, 34 MeV B₂, LUCITE
LOCALIZER MEASURED IN A RUBBER PHANTOM USING
BARE X-RAY FILM

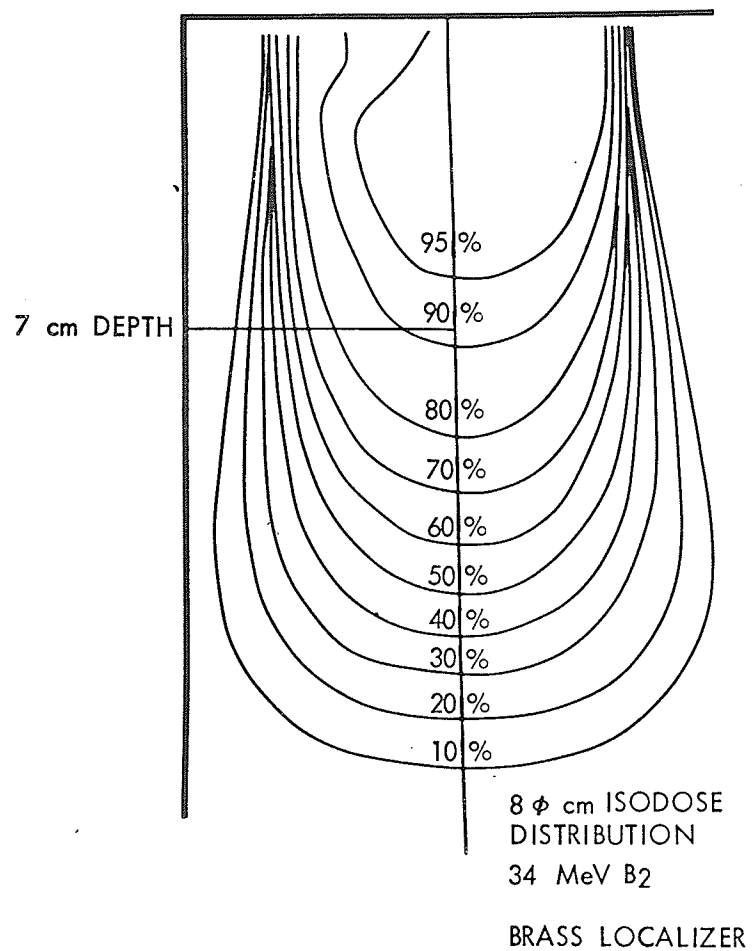


FIGURE 18

8 cm ISODOSE DISTRIBUTION, 34 MeV B₂ BRASS LOCALIZER MEASURED IN A RUBBER PHANTOM USING BARE X-RAY FILM

CHAPTER V

DISCUSSION

For the therapeutic use of radiation the most important parameter is the dose delivered to the patient.

Most institutes depend on field ionization chambers for which exposure calibration factors for ^{60}Co γ -rays or 2 MV x-rays have been determined at a standard laboratory. These calibrated chambers can be used to determine the dose at points in water or water equivalent material well beyond the maximum of the "build up" curve for a ^{60}Co γ -ray beam, because the quotient of dose and exposure is approximately independent of depth. This quotient, called the overall conversion factor (C_E) has been extensively studied and well established empirically and theoretically for ^{60}Co γ -rays (ICRU Report 14 (9), ICRU Report 23 (51) and HPA Report No. 4 (38)) and is equal to 0.95.

For fast electrons an analogous factor has been suggested but in this case the conversion factor has to be determined for each energy and depth because of the magnitude of the polarization effect in water compared with that for air. The magnitude of the factor (C_E) decreases as the electron energy increases. Theoretical values have been published by ICRU Report 21 (11) and a few measured values have been published by some authors.

The comparison between some practical and theoretical values has been illustrated in the literature review Chapter II. Thus ion chambers may be used as a basis for the determination of the dose delivered to a patient. However, field ionization chambers require calibration at standard laboratories usually situated at a considerable distance from the institution in which

they are required for use. Consequently events may occur which cause a change of the calibration factor such as the possibility of damage in shipment or replacement of some parts of the electronic circuits during maintenance or repair. So even with an ionization chamber dosimeter calibrated at a standard laboratory, there is no security for the determination of dose at the local institute. Moreover the only calibration factor given by most standard laboratories is for Cobalt 60 γ -rays or 2 MV x-rays. Therefore, at institutes using accelerators to produce electrons of energy greater than 3 MeV for radiotherapy, standard methods such as the Fricke chemical dosimeter, calorimeters or standard extrapolation ionization chambers have to be developed for absolute determination of dose.

The Fricke chemical dosimeter is considered one of the most suitable standard dosimeters for x-rays and gamma rays with maximum photon energies between 0.6 and 50 MeV (ICRU Report 14 (9)).

Ferrous sulfate dosimeter, the basis of the Fricke dosimeter, has many features which makes it suitable for absolute dosimetry

(a) It requires relatively simple equipment for chemical analysis

(b) The dose response is independent of dose rates up to 10^8 rads/second

(c) The conversion factor required to convert the dose determined for the dosimeter solution to that in water at the same location is close to unity for radiation considered here, and has much less dependence on radiation quality than for gaseous detectors.

(d) The dosimeter solution has nearly the same linear attenuation coefficient for tissue-like materials as water or water

equivalent phantoms, hence the dosimeter introduces small or negligible perturbation in the radiation field.

Irradiation cells and methods of removing impurities were discussed in the literature review. It has been reported that these effects are greater with high energy electron beams so that the dosimeter is less accurate than other dosimeters. Impurities from plastic containers are reported to give rise to an S.D.M. $\pm 1.5\%$ or more for measurements around 10,000 rads. Glass containers were recommended to yield better results. However, only limited success has been obtained with the chemical dosimeter system for high energy electron beams (ICRU Report 21 (11)).

Different collimating systems used with high energy electron beams were found to have a great influence on the rr/mr . If the field size is defined entirely by a metal collimator on the surface of the phantom it is seen from the results in Fig. 9 and Fig. 10 that the rr/mr is independent on the area of the field size, whether it be a circular, square, or rectangular field, at a depth of measurement 2.6 gm/cm^2 or 4.8 gm/cm^2 for energies 10 and 20 MeV. For the field size 4 cm circular and 4 x 4 cm square it appears in Fig. 9 and Fig. 10 that the rr/mr at either depth is less than for larger fields. It was found, however, that the size of the ionization chamber which covers about 2.5 cm perpendicular to the central axis of the field may be the cause of this reduction in ionization measurements. To examine this an experiment was done using the metal collimator on the surface of the phantom defining an area $4 \times 10 \text{ cm}^2$ with the 4 cm dimension perpendicular and then parallel to the ionization chamber.

In the former case the ionization measurements were

reduced while in the latter case no significant changes were detected. With 34 MeV electrons, however, the rr/mr increased gradually by about 7% as the area of the field sizes decreased from 60 to 13 cm² at depth 2.6 gm/cm². This may be due to the production of x-rays or scattered electrons from the collimating system for small field sizes.

When the collimator base sandwich was used in conjunction with previous arrangement or with the lucite collimator as supplied by the manufacturer it was found Fig. 11 to Fig. 14 that as the area decreased below 60 cm² the rr/mr also decreased for 10 and 20 MeV electrons. This decrease might be due to any or a combination of the following possibilities:

1. The electrons emerge from the scattering foil in a large solid angle due to scattering from the window and from the scattering foil. Some of these electrons will be scattered through angles large enough that they will strike the lead or steel structures lining the direct pathway from the target to the end of the collimator. When the base sandwich is inserted, some of the electrons scattered from these structures will be intercepted, thus reducing the dose delivered to the phantom axis per count on the monitor. The reduction will be expected to be more pronounced as the aperture in the base sandwich is reduced in agreement with our findings. This is a reasonable explanation for finding that the number of rads per count is less with a field of small area than with a field of large area, especially at low primary electron energies, where the cross-section for the scattering process is high.

2. The electrons may be backscattered from the base

sandwich increasing the ionization in the transmission chamber which is connected to the monitor device of the betatron and as a result the Baldwin Farmer ionization chamber would read less relative to the transmission chamber. The number of backscattered electrons would be expected to increase as the field size decreases since the base sandwich blocks a larger cross-section of the beam. An experiment was performed by irradiating the Baldwin Farmer chamber in the path of the electron beam in air with and without a back-scatterer made up about 1 cm of aluminum and 1 cm of lead which approximates the construction of the base sandwich. The back-scatterer was placed 10 cm farther from the Baldwin Farmer chamber. This distance is approximately similar to the distance between the base sandwich and the transmission chamber. The experiment was repeated for a closer distance 7 cm from the Baldwin Farmer chamber. The results obtained for ionization measurements for both distances with and without backscatterer indicates that there was no significant contribution to the ionization in the Baldwin Farmer chamber, because of backscatter. Therefore, it is unlikely that the backscattered electrons would contribute to the dose in the electron monitoring chamber.

3. The Baldwin Farmer ionization chamber may be of such size that the source of the electrons coming from the edges of the scattering foil may be intercepted by the diaphragm. This reason was also excluded because it was found that the Baldwin Farmer ionization chamber does not miss any part of the scattering foil when the smallest field size, 4 cm diameter, was used. Fig. 19 shows approximately the geometry of the position of the ionization chamber relative to the position of the scattering foil for 4 cm diameter field in the betatron.

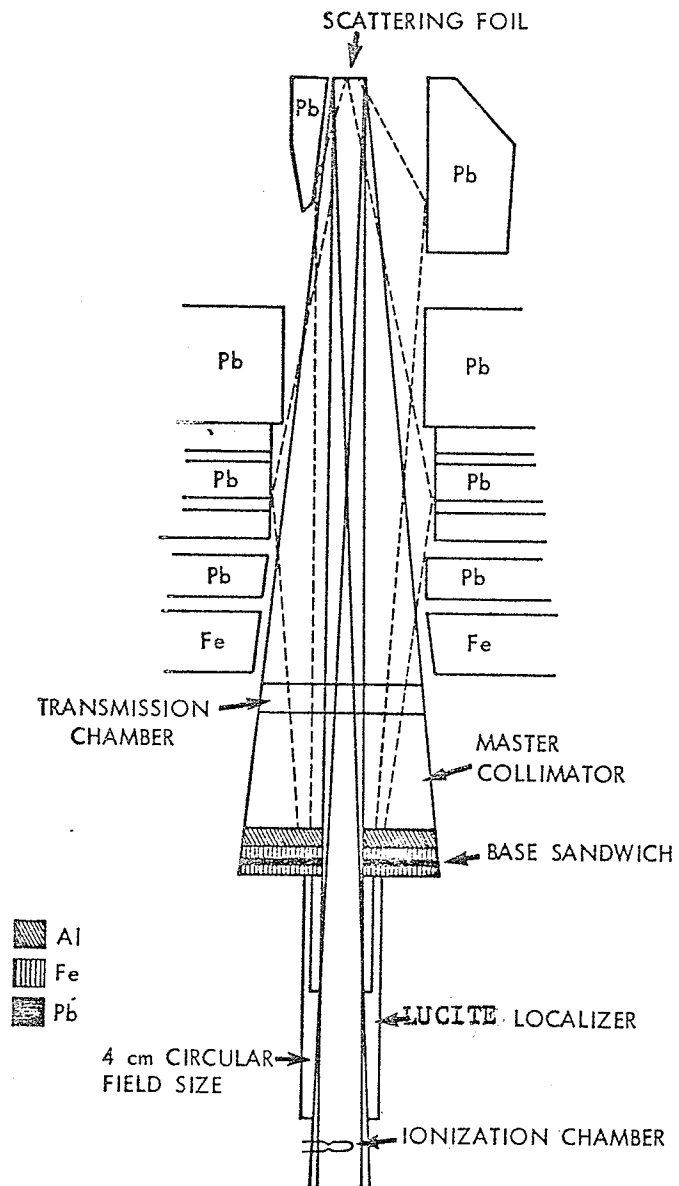


FIGURE 19

GEOMETRY OF THE POSITION OF THE IONIZATION CHAMBER RELATIVE TO THE POSITION OF THE SCATTERING FOIL IN THE BETATRON

Therefore, the most likely explanation for the decrease in the measurements for field sizes with area less than 60 cm^2 is the first reason. No evidence has yet been found to disprove the first reason given and the author proposes to accept this explanation.

When the original treatment collimator was used the rr/mr increases to about 6% for field sizes when the area decreases from 60 to 13 cm^2 at small depth 2.6 gm/cm^2 . This increase appears to be due to scattered electrons from the lucite tube at 34 MeV electron beams.

As previously mentioned if the field size is entirely defined by a lead collimator on the surface of the phantom (arrangement (a)) the rr/mr is approximately independent of the area of the field size at a depth of measurements 2.6 or 4.8 gm/cm^2 along the central axis in the lucite phantom with 10 and 20 MeV electrons, and at a depth 4.8 gm/cm^2 with 34 MeV electrons. This independence can be accounted for on the basis of either of the following assumptions.

(1) There is no scattering contribution to the points of measurements either because of the presence of lead plates or the phantom itself, thus the value of the rr/mr will be independent of area.

(2) There is a scattering contribution from the lead plates and the phantom. At larger field sizes the small scattering contribution coming from the lead is supplemented by a large contribution of scattering coming from the phantom and as the field size is reduced the scattering from the phantom to the points of measurements also is reduced and compensated by a larger contribution coming from the lead. Thus for both small and large field sizes the

sum of the two contributions is constant.

If the first assumption is correct, then when metal blocks are inserted inside the lucite collimator (arrangement (c)) to reduce it to a required field size or shape one would not expect a change in the value of rr/mr . If however the second assumption is correct one could not assume that the rr/mr value for the altered field size would remain the same. An experiment was performed by inserting a 1 x 6 cm brass plate of thickness about 2 cm, in both sides of the lucite localizer of a collimator defining a field size 6 x 8 cm to reduce it to 6 x 6 cm field size. Ionization measurements were made using the 6 x 8 cm, 6 x 6 cm, and 6 x 6 cm reduced from the 6 x 8 cm field sizes. The measurements were carried out at depths 2.6 and 4.8 gm/cm² along the central axis in the lucite phantom with 10, 20 and 34 MeV electrons. The value of rr/mr obtained from the reduced collimator was found to be greater than those obtained for either the 6 x 6 cm or 6 x 8 cm field size, except at 10 MeV where it was similar to that obtained for the 6 x 8 cm field size. This increase was presumably due to scattered radiation coming from the brass plates inserted inside the lucite localizer. Therefore, it appears that scattered electrons originating in the phantom and from the defining metal blocks do contribute significantly to the dose delivered to the patient. Consequently a new calibration must be performed when materials are inserted inside the lucite localizer to alter the shape of the field size.

The effect of insertion of metal materials inside the lucite localizer on the shape of the isodose distribution was also investigated by measurements performed by photographic film with 34 MeV electrons for a 6 x 8 cm field size reduced to a 6 x 5.5 cm.

This reduction was achieved by placing a 2.5 x 6 cm brass plate of thickness about 2 cm in one side of the 6 x 8 cm field size. The isodose distribution was found to be less distorted by scattered radiation and the penumbra was reduced in that half of the isodose distribution which was blocked by brass compared to the other side. This result is in agreement with our findings when the lucite localizer was replaced by brass localizer in the results Chapter IV, Section E.

CHAPTER VI

CONCLUSIONS

In this work the Fricke chemical dosimeter was used after some modifications. These modifications were in the irradiation containers. After several trials using different types of containers for irradiation of Fricke solution plastic test tubes manufactured by Falcon Plastics type 2054 made of polystyrene materials were found the most satisfactory. This choice was made when it was found that the reproducibility of the results was at least as good as for glass containers recommended by ICRU Report 21 (11) and HPA Report Series No. 4 (38).

This choice has the additional benefit that

1. The atomic number of the wall material is most nearly water equivalent.
2. No physical or chemical treatment of container was used before the measurements were performed.
3. Measurements around 3,000 to 4,000 rads show a S.D.M. of $\pm 0.2\%$.
4. The containers are commercially available, inexpensive, disposable and only used once for each measurement.

Cobalt-60 is used in most radiotherapy centres and therefore serves as an excellent standard beam for irradiating and calibrating dosimeters. Its conversion factor (C_E) to determine the dose is well established as previously mentioned. Therefore experiments using Cobalt-60 γ -rays were performed to check the chemical system used in this thesis and its precision.

The experimental results for the determination of the overall conversion factor (C_E) for ^{60}Co γ -rays show excellent agreement with the published value and suggest that there is no

systematic error in the use of chemical dosimetry as described in Chapter III "Materials and Methods".

The feasibility of the use of mailed chemical irradiation containers has been examined between the M.C.F. and the N.R.C. in Canada. The determination of dose using Fricke chemical dosimeter by the two institutes has been inter-compared and the results show maximum difference of 2%. This is considered to be satisfactory for the purpose of comparing clinical results of treatment in different centres.

The estimated uncertainties in the determination of dose in the tissue equivalent phantom using the ferrous sulfate system are given in Table 9 for electrons in the range 10 to 34 MeV, with the predominant contribution coming from the uncertainty in the G-value.

TABLE 9
UNCERTAINTY IN THE DETERMINATION OF THE DOSE AT A
POINT IN THE PHANTOM USING THE CHEMICAL DOSIMETER^a

Source	Uncertainty
Measurement of V(OD) (for 3 to 4 kilorads) ^b	0.4%
G-value	4.0%
Molar extinction coefficient	0.5%
Other effects discussed in literature review	0.2%
Determination of D_w from D_s	0.2%
Total	4.06%

^aThe overall uncertainty was determined by taking the square root of the sum of the squares of the separate uncertainties.

^bThe uncertainty of the optical density was taken twice its S.D.M. and includes the uncertainty due to positioning of the dosimeter.

A disadvantage of the Fricke dosimeter system is its relative insensitivity. This makes for great difficulty if a radiation distribution is to be obtained as opposed to the measurement of dose at one point.

With ^{60}Co γ -rays, or high energy x-rays, ion chambers have been shown to be extremely useful because of their sensitivity, ruggedness and small size. In such beams, the sensitivity of the chamber is independent of depth since the mean energy of the beam varies little with depth beyond the initial build-up region.

One difficulty with the use of ionization chambers for the determination of absorbed dose in media irradiated by high energy electrons is the dependence of their calibration on the energy of the electrons at the point of measurement. In order to use these instruments even for measurements of relative dose rate; such as when measuring depth dose data, it is therefore essential to know two things:

(1) The energy response of the ion chamber for the medium in question (C_E).

(2) The average energy of the electrons at the depth in question.

Methods have been proposed in ICRU 21 (11) for determining each of these parameters.

It may be seen from the results shown in Tables 6,7 and 8, as well as Figure 8, that the use of these methods give dose estimates in satisfactory agreement with those obtained independently by the Fricke dosimeter.

Thus one may conclude that the use of C_E values given in ICRU 21 (11) for electrons of average energy determined at the point

of measurement by methods given in the same publication, will permit the determination of absorbed dose with a precision satisfactory for clinical work.

The finding that the NRC ferrous sulphate dosimeter gave results in good agreement with ours suggests that such dosimeters, mailed from a central standardizing laboratory will be of great value in ensuring that all participating institutions have measured their absorbed dose in such a way as to yield comparable results.

In summary it was concluded from this work that

(1) The dose can be measured absolutely at any point in a phantom using a chemical dosimeter with a total uncertainty (95% confidence limited) of about 4%.

(2) Ion chambers may also be used to obtain absorbed dose at any point in a phantom irradiated by electrons provided that the procedures given in ICRU 21 (11) are followed for obtaining the average electron energy at the point in question and that the appropriate value for C_E is used.

(3) The isodose distribution can be improved both in symmetry and reduction of the penumbral region by replacing the lucite localizer by a localizer using brass defining rings at the patient's skin.

(4) No simple method can be used to predict the dose when materials are inserted inside the localizers to alter the shape of the field size and in each case a new calibration must be performed.

BIBLIOGRAPHY

- {1} A. Dutreix, J. Dutreix, and A. Wambersie, Ann. New York Acad. Sci. 161, 272 (1969).
- {2} A. Wambersie, A. Dutreix, J. Dutreix, and M. Tubiana, Symposium on High Energy Electrons: (editors, A. Zuppinger and G. Poretti) p140 (1965) Springer, Berlin.
- {3} J. Eugene Robinson and T. L. Ervin, Ann. New York Acad. Sci. 161, 301 (1969).
- {4} J. H. Kim, A. Pinkerton, and J. S. Laughlin, Ann. New York Acad. Sci. 161, 310 (1969).
- {5} W. Pohlit, Ann. New York, Acad. Sci. 161, 189 (1969).
- {6} Arne Dahler, Alan S. Baker, and J. S. Laughlin, Ann. New York Acad. Sci. 161, 198 (1969).
- {7} Max L. M. Boone, Peter R. Almond, and Ann E. Wright, Ann. New York Acad. Sci. 161, 214 (1969).
- {8} M. Brenner, P. Karjalainen, A. Ryttilä, and H. Jungar, Ann. New York Acad. Sci. 161, 233 (1969).
- {9} International Commission on Radiation Units and Measurements ICRU Report 14 (1969).
- {10} International Commission on Radiation Units and Measurements ICRU Report 17 (1970).
- {11} International Commission on Radiation Units and Measurements ICRU Report 21 (1972).
- {12} H. Fricke and S. Morse, Am. J. Roentgenol. Radium Therapy. Nucl. Med. 18, 430 (1927).
- {13} H. Fricke and S. Morse, Phil. Mag. [7] 7, 129 (1929)
- {14} H. Fricke and J. Edwin, Radiation Dosimetry Vol. 11, 167, F. H. Attix, and W. C. Roesch eds. (Academic Press Inc. New York)(1966).

- {15} A. O. Allen and W. G. Rothschild, Radiation Res. 7,
591 (1957).
- {16} A. O. Allen, V. D. Hogan, and W. G. Rothschild,
Radiation Res. 7, 603 (1957).
- {17} J. Baarli and P. Borg, Acta Radiol. 47, 203 (1957).
- {18} J. V. Davies, D. Green, J. P. Keene, J. Law, and
J. B. Massey, Phys. Med. Biol. 8, 97 (1963).
- {19} N. A. Frigerio, Radiation Res. 16, 606 (1962).
- {20} J. Geisselsoder, K. Koepke, and J. S. Laughlin,
Radiation Res. 20, 423 (1963).
- {21} E. J. Hall and R. Oliver, Brit. J. Radiol. 34, 397 (1961)
- {22} E. J. Hart, Radiation Res. 2, 33 (1955).
- {23} C. Petterson and G. Hettinger, Acta Radiol. 6, 160 (1967).
- {24} A. Wambersie, A. Dutreix, M. Prignot, and J. Dutreix,
Biophysik 3, 167 (1971).
- {25} Mukund Kartha and John C. F. MacDonald,
Brit. J. Radiol. 43, 549 (1970).
- {26} A. O. Fregene, Radiation Res. 31, 256 (1967).
- {27} H. Svensson, Acta Radiol. 10, 631 (1971).
- {28} H. Svensson, G. Hettinger, D. Frost, and W. Pohlitz,
Acta Radiol. 10, 54 (1971).
- {29} T. H. E. Bryant and T. P. Ridler, Health Physics 15,
263 (1968).
- {30} J. Law, Phys. Med. Biol. 14, 607 (1969).
- {31} Dieter Frankenberg, Phys. Med. Biol. 14, 597 (1969).
- {32} R. K. Broszkiewicz, and Z. Bulhak, Phys. Med. Biol. 15,
549 (1970).
- {33} J. V. Davies and J. Law, Phys. Med. Biol. 8, 91 (1963).

- {34} H. Svensson, C. Patterssen, and G. Hettinger,
Symposium on solid state and chemical radiation dosimetry
in medicine and biology (I.A.E.A., Vienna) p. 251 (1967).
- {35} P. R. Almond, Phys. Med. Biol. 12, 13 (1967).
- {36} P. R. Almond, Int. J. Appl. Rad. 21, 1 (1970).
- {37} D. Harder, Symposium on High Energy Electrons: (editors, A. Zuppinger
and G. Poratti) p.26 (1965) Springer, Berlin.
- {38} The Hospital Physicists Association. A practical guide
to electron dosimetry, HPA Report Series No. 4 (1971).
- {39} J. Dutreix and A. Dutreix, Ann. New York Acad. Sci. 161, 33 (1969).
- {40} G. Hettinger and H. Svensson, Acta Radiol. 6, 74 (1967).
- {41} R. Loevinger, C. J. Karzmark, and M. Weissbluth,
Radiology 77, 906 (1961).
- {42} Hospital Physicist's Association. "A code of practice for
the dosimetry of 2 to 35 MVX-ray and cesium-137 and Cobalt-60
gamma-ray beams". Phys. Med. Biol. 14, 1 (1969).
- {43} R. J. Schulz, Ann. New York Acad. Sci. 161, 176 (1969).
- {44} A. Dahler, Symposium on High Energy Electrons: (editors, A. Zuppinger
and G. Poratti) p.98 (1965) Springer, Berlin.
- {45} H. Svensson and G. Hettinger, Acta Radiol. 6, 289 (1967).
- {46} H. Svensson and G. Hettinger, Acta. Radiol. 6, 404 (1967a).
- {47} J. E. Robinson and R. S. McDougall, Phys. Med. Biol. 12, 315 (1967a).
- {48} A. F. Batho, O. Theimer, and R. Theimer, J. Canad. Ass. Radiol.
7, 51 (1956).
- {49} J. R. Clarkson, Brit. J. Radiol. 14, 265 (1941).
- {50} F. Bagne. Section of Medical Physics, Norris Cotton Cancer
Center Dartmouth-Hitchcock Medical Center, Hanover, N.H., U.S.A.
Research was supported by contract #NIH-70-2087, DHEW (1973).

{51} International Commission on Radiation Units and Measurements
ICRU Report 23 (1973).