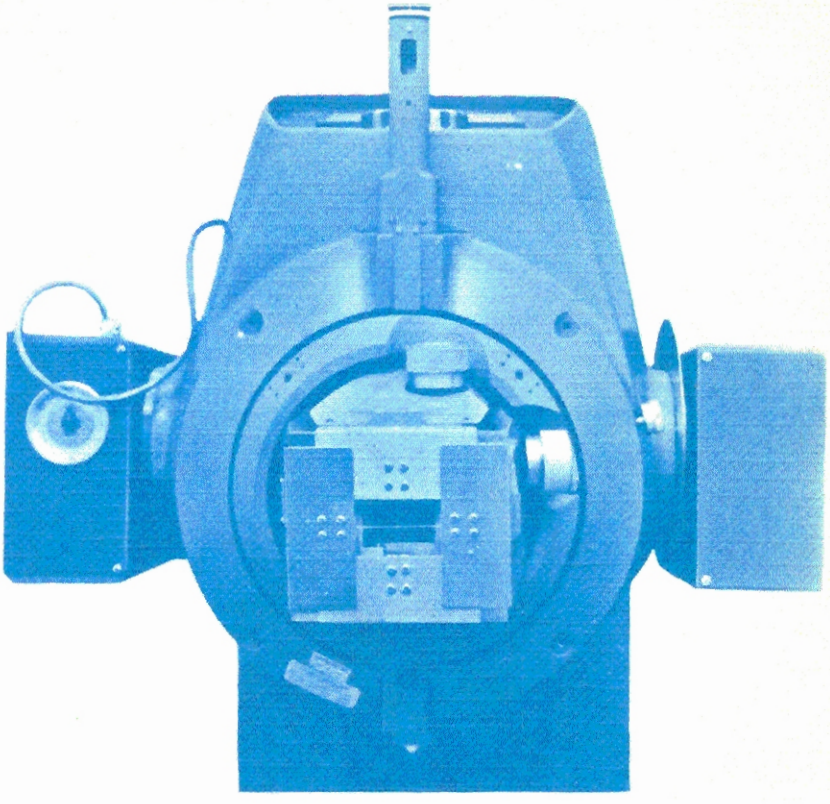

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Manual of Dosimetry in Radiotherapy

John B. Massey

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MANUAL OF DOSIMETRY IN RADIOTHERAPY

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A practical guide for testing and calibrating equipment used in external beam treatments

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and the
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FOREWORD

For radiotherapy to be safe and effective a number of technical conditions have to be met to ensure that the radiation beam is appropriate for each individual patient and that the dosage received by the patient is as planned. While it is the function of the radiotherapist to decide the characteristics of the treatment, he may not have the technical background to ensure that the equipment is operated in such a way that the desired results are achieved. This responsibility belongs to the medical radiation physicist.

Unfortunately, in many countries there is a shortage of qualified physicists experienced in radiotherapy. So although these countries may have many hospitals equipped with teletherapy and X-ray units, they are not able to use these facilities to the best advantage.

With this situation in mind, the International Atomic Energy Agency, with the active participation of the World Health Organization and the Pan American Health Organization, convened a panel of experts to study the dosimetric requirements of radiotherapy centres. The panel met in Caracas, Venezuela, from 22 to 26 April 1968. Although calling for the urgent establishment of training programs for hospital physicists, the panel considered it unlikely that enough full-time qualified experts would be available in the near future to meet the shortage. For several years to come, therefore, radiotherapists would have to rely on part-time physicists to advise them on practical dosimetry questions or would have to do the best they could without such advice. The available literature on this subject was directed mainly to the qualified physicist rather than to the practising radiotherapist. The panel therefore considered it desirable that a manual should be prepared, setting out clearly the basic principles of dosimetry in radiotherapy and providing a practical guide to the procedures that should be adopted to achieve accuracy of dosage.

Mr. John Massey undertook to prepare such a guide and early in 1969 submitted a first draft which was circulated to a number of experts for their comments. In April 1969 the text was reviewed in the light of these comments at a meeting in Geneva arranged by the IAEA and WHO and attended by the author. Mr. Massey then prepared a revised draft which was again circulated for comments and subsequently reviewed at a joint IAEA/WHO/PAHO meeting in Washington in August 1969, attended by four consultants. A final version was thus arrived at for publication.

The three sponsoring organizations, IAEA, WHO and PAHO, are greatly indebted to the author for his work.

AUTHOR'S ACKNOWLEDGEMENTS

The author is indebted to his present and former colleagues, both senior and junior, without whose joint experience the ideas embodied in this manual would never have been formulated. In particular, the continuous encouragement and guidance given by Dr. W. J. Meredith, Director of Physics at the Christie Hospital, have been invaluable. The relevance of this manual to clinical practice stems from the full and willing collaboration, extending over many years, of all the author's radiotherapist colleagues. This collaboration has been made possible and fully encouraged by Professor Ralston Paterson and Dr. Eric Easson, the former and present Directors of Radiotherapy of the Christie Hospital, to whom the author wishes to acknowledge his gratitude.

The many valuable comments on the first draft of the manual were made by the author's Departmental colleagues and by Drs. F. Behounek, P. Bird, H. Blatz, C. Braestrup, H. Eisenlohr, S. O. Fedoruk, P. Frost, K. Koren, R. Loevinger, S. Maudal, W. Mauderli, W. Minder, S. B. Osborn, P. M. Pfalzner, C. Pomroy, J. Reboul, W. Seelentag, B. Waldeskog and M. M. D. Williams. The consultants who discussed the revised draft in detail with the author at the Washington meeting were Drs. O. Machado, H. Mugliaroli, P. M. Pfalzner, M. Zaharia, in co-operation with Drs. H. Eisenlohr (IAEA), D. Joly (PAHO), J. Litvak (PAHO) and B. Waldeskog (WHO). Dr. Russell Morgan also attended this meeting on behalf of the International Commission on Radiological Units and Measurement.

Published literature is specifically referred to only where such reference either provides data required for implementing the techniques described here or gives supplementary information. No attempt has been made to refer to the very large volume of published work on which the author has drawn freely. It is sincerely hoped that the incorporation of their ideas will be regarded as sufficient acknowledgement by the authors of these publications. A bibliography of associated and further reading is given.

In spite of the help from many sources that the author has received in writing this manual, he wishes to stress that the responsibility for any ambiguities, omissions or errors is his alone.

CONTENTS

CHAPTER I

INTRODUCTION

This manual is intended as a practical guide to the procedures necessary for achieving good physical accuracy of irradiated volume and dosage in external beam therapy.

In planning a radiotherapy treatment the radiotherapist has to determine the location and to decide the volume of tissue to be irradiated, the dose to be delivered to this volume, the fractionation and the overall time of the treatment. The radiotherapist may have great difficulty in making these decisions but this manual is not concerned with these problems. There is no doubt, however, that, having made his choice, the radiotherapist needs to be confident that the chosen volume in the patient has, in fact, received the chosen dosage.

There are many radiotherapy centres at which the desired accuracy is already substantially achieved and for such centres this manual probably contains little that is unfamiliar. It is hoped, however, that even for these centres it can act as a useful reminder and source of reference. Unfortunately, there also exist many radiotherapy installations at which attention to the physical and technical aspects of radiotherapy described here is at best cursory and all too often insufficient. This may be because suitably qualified scientific or technical staff is difficult or impossible to obtain or, worse, because there is little local realization of what is required.

The preparation of this manual was recommended by the participants of an IAEA panel on 'Dosimetric Requirements of Radiotherapy Centres' held in Caracas in 1968. Its implementation is the result of the co-ordinated action of the IAEA, WHO and PAHO. Although the emphasis of the panel discussion was on the needs of the countries of Latin America, there can be little doubt that the contents are of importance and relevance to many radiotherapy centres in other parts of the world.

It is very well recognized that the best radiotherapy is possible only in those centres which have available a well-equipped, well-staffed and experienced full-time physics service. Unfortunately it is unrealistic not to recognize that in too many parts of the world it is not possible to have such a service and that radiotherapy will be carried out using a part-time physicist or even no physicist at all. It is imperative to realize that it is essential to give proper attention to the physical aspects of the treatment. This manual is therefore particularly directed to the graduate in a physical science who is not yet experienced in the technical and physical aspects of radiotherapy and to the radiotherapist who is acting also as a physicist.

The instructions, advice and recommendations given in this manual are, as far as possible, straightforward and, it is hoped, unambiguous. This must not be interpreted as meaning that there is only one proper way of doing the work but merely that this is one reasonable and correct way. If the reader is already acquainted with an alternative method of achieving the same end and is confident that it is at least as good as that presented here, then there is no reason to change. In that case it is most important to be sure that any such alternative methods do not conflict with any of the procedures given in the manual which are adopted. Since this can be very difficult to ensure, it is firmly recommended that, unless there are good

reasons to do otherwise, the procedures set out here should be adopted in full for all radiotherapy installations.

Care has been taken that the instructions given and the recommendations made in this manual do not conflict with established practice, descriptions of which are published in various journals, reports, monographs, books, etc.

Since this manual is intended only as a 'code of practice', or protocol, little attempt is made to justify or explain the detailed background of the technique described. Clearly it is easier and more satisfactory when there is a full understanding of the problems and for this reason some explanatory remarks are included. Nevertheless this manual can never (and is not intended to) replace either the qualified, well-instructed and experienced physicist or the formal text-book. A small bibliography of publications that can assist in the understanding of the problem is given in Appendix I.

There is no doubt that much help can be obtained from the published literature, yet a more personal kind of help is desirable. It is strongly recommended that any small centre of limited experience, and especially any new radiotherapy centre, should try to enlist the co-operation of a larger, more experienced centre which will be able to give invaluable help, even by correspondence.

The main emphasis in this manual has been placed on cobalt teletherapy equipment and on 200-400 kV X-ray equipment. Many of the instructions apply equally well to megavoltage X-ray equipment (e.g. linear accelerators and betatrons) but for such equipment it is even more important, in fact imperative, to have available a full-time trained physics and maintenance service. Attention is also paid to the special requirements of low kV X-ray equipment and techniques (i.e. the so-called 'contact' and 'Grenz ray' therapies).

Implementing the recommendations given here will necessitate spending a certain amount of money in the purchase of some equipment – if this is not already available – and for its maintenance. The amounts involved are not large and represent a very small fraction of the capital expenditure and running costs of a radiotherapy centre. All too often, however, the need to spend this money is overlooked with the result that the success of the radiation treatment given to the patients may be seriously compromised.

Some of the remarks and recommendations are, by intention, made at more than one place in this manual. Nevertheless the full import of some of the recommendations and text will not be apparent until the whole document has been read. It cannot be overemphasized that the information given here and the recommendations made are to be regarded as a complete scheme. As stated above, it may be extremely hazardous to select parts of these recommendations and to mix them with other methods. It is important to note also that the order of presentation is not the order in which the work is best done. In particular, testing and checking the dosemeter and the equipment described in Chapters 3 and 6 and the protection survey described in Chapter 7 need to precede the detailed measurements described in Chapters 4 and 5. The recommended sequence of working is briefly as follows:

(a) Check that the dose measuring equipment is in good order
(b) Check that the teletherapy equipment is in good order
(c) Carry out the protection survey
(d) Measure the quality and select the depth dose data
(e) Measure the radiation output
(f) Carry out the routine testing of equipment and dosemeters.

Fuller details of the recommended sequence of working are set out in Appendix VII, together with references to the parts of the manual where the required detailed information is given.

A comprehensive list of the definitions of terms used in radiotherapy has been compiled by the International Commission on Radiological Units (ICRU), see Refs [1] and [2] in Appendix I. These terms have well-established and precise meanings and it is important to conform to this usage. Definitions of selected terms (Table I) which are of major and immediate relevance to this manual are given in alphabetical order in the Glossary (Appendix II). Where appropriate and in the interests of clarity the definition given there is an excerpt, amplification or summary of that given in the original document.

TABLE I. RADIOLOGICAL TERMS DEFINED IN APPENDIX II

Exposure	Percentage depth dose (%DD)
The röntgen	Isodose surface
Absorbed dose	Isodose curve (or contour)
The rad	Isodose chart
Radiation quality	Source-surface distance (SSD)
Half-value thickness (HVT)	Central ray
Quality factor	Geometrical edges of the beam
	Geometric field (beam) size
Output	Wedge isodose angle
Surface absorbed dose	Treatment session, fractionation
Peak absorbed dose	Tissue-equivalent medium
Given (or applied) dose	Phantom
	Build-up
Tissue-air ratio	Surface backscatter factor
Dose distribution	

The ICRU recommends the concept of 'absorbed dose' (energy absorbed per gram of material) and its unit, the rad, for the statement of radiotherapy treatment doses and it is strongly urged that this recommendation should be adopted universally. However a great deal of radiotherapeutic experience is based on the use of the older concept of 'exposure' and its unit, the röntgen. In this manual, therefore, sufficient information will be given to enable the user to employ either rad or röntgen. It is hoped, however, that statements about radiotherapy treatment in terms of exposure (röntgen) will gradually be superseded by the use of absorbed dose (rad).

When stating the dose in rad it is necessary also to state the kind of material being irradiated. The material most commonly encountered in radiotherapy is, of course, muscle or soft tissue and it is common practice therefore to state the dose in 'rad in soft tissue'. In this manual when the term 'rad' is used without qualification 'rad in soft tissue' is intended.

CHAPTER 2

THE OBJECTIVE

It has already been stated that the purpose of this manual is to describe what must be done to ensure that, the radiotherapist having selected the appropriate external beam treatment for any individual patient, the desired dose pattern and dosage level are achieved at the chosen position within the patient. This requires not only that the appropriate radiation beams shall be correctly positioned and directed but also that the radiotherapy equipment shall be switched on for the correct length of time.

It is not intended that this manual should describe the process of planning a treatment. For accounts of treatment planning the reader is referred to appropriate text books, some of which are listed in the bibliography. It will therefore be assumed that the reader is familiar with, at least, the broad outline of what is involved and with the specialized terms used. The details of the way in which a treatment is planned can vary from centre to centre and from country to country and the choice is largely a personal one. It is, however, necessary to select one particular method on which to base the contents of this manual. Only in this way can recommendations that are specific and unambiguous be made. Since direct measurement of the radiation on or in the patient is seldom possible and is always fraught with great experimental and practical difficulty, indirect methods are usually preferred. A brief account of the particular method selected is given below.

2.1. TREATMENT PLANNING

The problem is to ensure that (a) the desired pattern of radiation (dose distribution) within the patient is achieved; and (b) the desired amount of radiation (dose) is delivered to the tumour.

2.1.1. Dose distribution

The dose distribution resulting from a combination of several beams is determined by summing the separate contributions from each beam. As a first approximation it is usual to regard the patient's tissue as being equivalent to water. Information about the radiation pattern that exists inside a large homogeneous water phantom irradiated by a single beam is presented in the form of isodose charts or central axis percentage depth dose tables. The detailed characteristics of the isodose surfaces and values of percentage depth dose are determined by such parameters as the radiation quality, beam size and source-surface distance. Central axis % depth doses and isodose charts are usually specific to the particular radiation source being used since both are affected by the design of the treatment cone or collimating system. The methods of dosimetry recommended in this manual are such as to minimize the effects of this dependence, at least as far as the central axis %DD is concerned. Well-established values of percentage depth dose data are given in 'Depth Dose Tables for Use in Radiotherapy' [3]. Typical examples of single-field isodose charts and of combined dose distribution resulting from various combinations of fixed and moving fields are given in

volumes I, II and III respectively of the IAEA Atlas of Radiation Dose Distributions [4]. It is worth noting that the introductions to each of the three volumes of that Atlas provide useful and interesting descriptions of the parameters which control the details of the dose distribution produced by single, multiple and moving fields respectively.

For any particular patient the combined dose distribution may be obtained by adding the contributions from the separate beams at an array of points within the patient. The result is that the combined dose at every chosen point of the array is known in terms of the relative 'given dose' (i.e. surface or peak dose) on each of the individual fields. Alternatively, if the treatment is being specified by exposure (röntgen), the pattern will be known in terms of the relative surface or peak exposures.

In practice, of course, the treatment is designed by first determining the relative values of the given dose (or given exposure) on each field required to produce the desired pattern of dose (or exposure) in the region of the tumour.

2.1.2. Dose

The actual magnitude of the given doses (or exposures) required to produce a chosen dose (or exposure) at the tumour is then calculated by simple proportion.

For those readers who are not familiar with this method of treatment prescription some examples are given in Appendix IV. In addition they are strongly urged to study the problem of treatment design separately. Descriptions of this are given in the radiotherapy texts listed in the bibliography (Appendix I). This list must not be regarded as comprehensive; many other suitable texts exist.

2.2. CONTROL OF DOSE DELIVERY TO THE PATIENT

Control of dose delivery is usually based on either a timer or a monitor dosemeter. The timer is appropriate to telecobalt or telecaesium equipment where the dose rate during an exposure is constant and to conventional X-ray equipment where the dose rate can be kept constant by keeping the tube voltage (kV) and current (mA) fixed. A monitor dosemeter is necessary where the dose rate cannot be kept constant. Even when the dose rate can be kept constant a monitor system is very useful and is strongly recommended.

On many occasions in this manual, and in practice, it is necessary to refer, in general terms, to the amount of radiation being directed from radiotherapy equipment towards the patient or to any general specified point. The expressions 'radiation output' and 'output' will be used for this general, non-specific quantity. In later sections the amount of radiation delivered to a specific point in specific circumstances will be precisely defined and its measurement described.

2.2.1. Telecobalt and telecaesium equipment

During a treatment session the rate of radiation output is constant and therefore the total dose delivered to the patient is satisfactorily and usually controlled by the use of a timer. The usual method of working is to set the

desired exposure time (the determination of which is described below) on the timer. When the exposure ON button or switch is operated, the source or 'shutter' moves to the ON position. As soon as it reaches this position the timer starts to run. As soon as the set time has elapsed the source or 'shutter' is returned to the safe (OFF) position and the exposure automatically terminated (see also section 2.2.3).

For telecaesium and telecobalt equipment the output rate for a specified source-surface distance (SSD) and beam size will be stated as:

$\dot{X}_s$ röntgens per minute (R/min)

or

$\dot{D}_s$ rad per minute in soft tissue (rad/min)

where $\dot{X}_s$ and $\dot{D}_s$ are the values of exposure rate and absorbed dose rate in soft tissue at the peak of the depth dose curve respectively. X_s and D_s are the corresponding values of exposure (röntgen) and dose (rad) respectively.

The numerical value of $\dot{X}_s$ or $\dot{D}_s$ for a particular telecaesium or telecobalt therapy machine operating under specified conditions (SSD, beam size, wedge, etc.) is determined by the method described in Chapter 4. The time required to deliver the desired exposure (X_s) or dose (D_s) at the peak is determined simply by use of either:

$$\text{Exposure time} = \frac{\text{Peak exposure (R)}}{\text{Peak exposure rate (R/min)}}$$

$$= X_s/\dot{X}_s \text{ min}$$

or

$$\text{Exposure time} = \frac{\text{Peak absorbed dose (rad)}}{\text{Peak absorbed dose rate (rad/min)}}$$

$$= D_s/\dot{D}_s \text{ min}$$

2.2.1.1. Radio-active decay

(a) Cobalt-60: The output of the telecobalt equipment will have been measured on a particular date. At the time of treatment for exactly the same beam size, SSD etc., the output will be less because of the radio-active decay of the cobalt. Account must be taken of this. Table II gives

TABLE II. ^{60}Co DECAY FACTORS (Half-life 5.26 yr)

Months	Years					
	0	1	2	3	4	5
0	1.000	0.877	0.768	0.674	0.590	0.517
3	0.968	0.848	0.743	0.652	0.571	0.501
6	0.936	0.821	0.719	0.631	0.553	0.484
9	0.906	0.794	0.696	0.610	0.535	0.469
12	0.877	0.768	0.674	0.590	0.517	0.454

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$$= X_s/\dot{X}_s \text{ min}$$

or

$$\text{Exposure time} = \frac{\text{Peak absorbed dose (rad)}}{\text{Peak absorbed dose rate (rad/min)}}$$

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2.2.1.1. Radio-active decay

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TABLE II. ^{60}Co DECAY FACTORS (Half-life 5.26 yr)

Months	Years					
	0	1	2	3	4	5
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9	0.906	0.794	0.696	0.610	0.535	0.469
12	0.877	0.768	0.674	0.590	0.517	0.454

examples of the values of the factor by which the measured output must be multiplied to determine the reduced output existing at a later date. A more comprehensive table intended for practical use is given in Appendix V.AV.

The following is an example of the kind of calculation required. The data refer to a particular telecobalt machine for a beam size of 10 × 10 cm at an SSD of 80 cm.

(a) Peak dose rate ($\dot{D}_s$) on 15 September 1968 was 80 rad/min in soft tissue

(b) Date of treatment 15 June 1970

(c) Required peak dose (given dose) is 400 rad in soft tissue.

The period between the date of the output measurement (15 September 1968) and that of the treatment (15 June 1970) is 1 year 9 months. From Table II the appropriate decay factor is found to be 0.79. Hence, the peak dose rate on 15 June 1970 would be

$$80 \times 0.79 = 63.2 \text{ rad/min in soft tissue}$$

The exposure duration to deliver 400 rad is therefore

$$400/63.2 = 6.34 \text{ min}$$

or 6 min 20 s.

It can be seen from Table II that the decay correction is only about 1% per month. It is therefore sufficient to correct to the nearest month. For example, the correction factor of 0.79 quoted in the above example can be applied to calculate the peak dose rate for any date within the period 1-30 June 1970. In circumstances where the treatment is fractionated over a period of several weeks, even this is unnecessarily precise. It is common practice therefore to restate the outputs of cobalt equipment at two or three-monthly intervals and to assume that the outputs are constant throughout this interval. For example, the peak dose rate for the 10 × 10 cm beam at 80 cm SSD measured to be 80 rad/min in soft tissue on 15 September 1968 would be assumed to have this value for the whole of the period 1 August – 31 October 1968. The reduced value of 63.2 rad/min in soft tissue, which strictly applies only to the 15 June 1970, would be assumed to apply for the three-monthly period 1 May - 31 July 1970.

The inconvenience of applying an individual decay correction to each patient's treatment and the consequent possibility of error can be avoided if the telecobalt equipment is provided with a new list of peak dose rates applicable to the various operating conditions (beam size, SSD, wedge etc.) at two or three-monthly intervals.

(b) Caesium-137: For radioactive caesium-137 with its much longer half-life (30 yr) it is sufficient from this point of view to restate the outputs at intervals of one or two years. However, since there is a possibility of different half-life contaminants being present, it is desirable that the output should be checked by measurement at intervals of not more than 6 months, especially during the initial years. Any restatement of output for caesium equipment should therefore be based upon measurement rather than a decay calculation.

2.2.2. X-ray equipment

The output of X-ray equipment can be maintained constant during an exposure by keeping the kV and mA constant, the appropriate controls being

adjusted whenever the associated meter readings depart from the chosen values.[1]

As in the case of the telecobalt and telecaesium equipment the output will be stated as either

$$\dot{X}_s \quad \text{R/min}$$

or

$$\dot{D}_s \quad \text{rad/min in soft tissue}$$

where the numerical value of $\dot{X}_s$ or $\dot{D}_s$ for any particular X-ray machine, SSD, beam size, filter and specified values of kV and mA meter readings is determined by the method described in Chapter 4. The constancy of output can be confirmed by using a monitor dose rate system. In this case the reading of the dose rate monitor is kept constant by adjusting the mA. This latter method of working is not very common since the use of an integrating (exposure or dose) system is preferable (see below).

The kV should be kept constant since it controls the radiation quality and therefore the penetrating power (% DD values) of the radiation. Fortunately the effect of changes in kV on percentage depth dose is not critical and it is sufficient, from this point of view, for the kV to be kept within ± 5% of its nominal value. Of course, the magnitude of the radiation output is very dependent upon the value of kV so it is only when a dose-rate or dose monitor is being used that this degree of tolerance is permissible. Even so it must be remembered that the sensitivity of the monitor ionization chamber may be energy-dependent and large deviations in kV can lead to errors in dosage for this reason.

When no monitor is fitted the kV should be maintained within 1-2% of the required value and the milliamperage within about 2%. If either of these levels is exceeded, the associated error in dose delivery will exceed 2%.

The exposure time required to deliver the desired surface exposure or dose is determined in exactly the same way as for the telecobalt unit (see above). Of course, in this case there is no radioactive decay to be taken into account. For example, consider a particular X-ray machine which is operated at 250 kV, 20 mA and an added filtration of 1.5 mm Cu. For a beam size of 10 × 10 cm and an SSD of 40 cm the surface exposure rate is measured to be 40 R/min. If the required surface exposure is 350 R, the duration of the exposure will need to be

$$350/40 = 8.75 \text{ min}$$

or 8 min 45 s.

2.2.2.1. Dose monitors

It is desirable to use a monitor dosemeter system (see below) to control the delivery of dose to the patient. Although it is usually possible to control the output rate from kilovoltage X-ray equipment by means of the tube

[1] The waveform of the voltage is also important and care must be taken that this also remains constant. Change of waveform may occur, even for a constant kV meter reading, on equipment which has more than one control for the voltage if the combination of control settings is changed.

voltage (kV) and current (mA) controls, it is more convenient and accurate to use a monitor dosemeter. With some X-ray equipment it is not possible to keep the kV and mA values steady and in these conditions the use of a monitor dosemeter is essential. For megavoltage linear accelerators and betatrons (with which this document is not concerned) such a system is likewise essential since the radiation output rate is not usually constant for the duration of the treatment exposure and is difficult or even impossible to control. It is strongly recommended that an integrating (exposure or dose) monitor system should be used on all X-ray equipment.

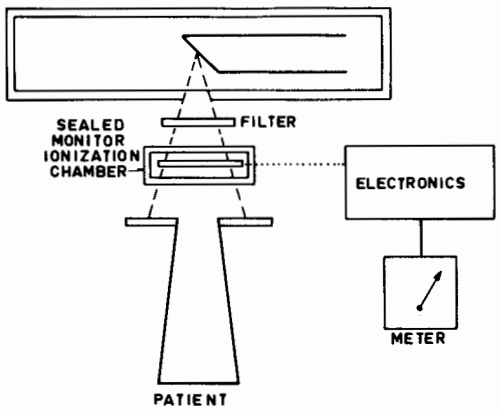

FIG. 1. Monitor ionization chamber system used for patient dose control.

(a) Sealed monitor chamber. The system consists of an ionization chamber positioned between the focal spot and the patient (Fig. 1), usually between the filter and the seating of the treatment cone. The chamber is a parallel plate chamber of uniform thickness so that it does not disturb the uniformity of the X-ray beam. It is highly desirable that the chamber be airtight (sealed) so that its response to radiation is independent of its temperature and pressure. The ionization current from the chamber is fed into a suitable electronic circuit. It is arranged that the reading of the associated meter is proportional to the electric charge fed in and therefore proportional to the amount of primary radiation which has traversed the chamber. Hence for any particular piece of X-ray equipment, monitor system, filter, treatment cone and kilovoltage the given dose is accurately proportional to the reading on the monitor meter and the output can therefore be quoted as either X'_s röntgen per monitor division or D'_s rad per monitor division, rather than röntgen or rad per minute as with a timer. The number of monitor divisions required to achieve the desired surface exposure (or dose) is calculated in the same way as is the exposure time.

For example: The following data apply to a particular 250 kV X-ray machine:

(1) Surface dose rate = 0.4 rad per monitor division
(2) Required given dose = 350 rad
Hence number of monitor divisions required
= 350/0.4 = 875 monitor divisions.

An alternative method is to use a small chamber which is positioned, within the beam, on the patient's skin or at the end of the treatment cone.

Such a system, although not recommended here, is acceptable provided it is recognized that, to be consistent with the scheme of working given in this manual, it requires to be tested and calibrated in exactly the same way as the parallel plate type described above by the methods described in Chapter 4.

(b) Unsealed monitor chamber. If the chamber cannot be made airtight, it will be necessary to apply a correction factor to its reading which will depend upon the temperature and pressure of the gas inside the chamber. Great care must be taken in doing this since the temperature and pressure conditions in the chamber may be different from those of the treatment room. This is particularly so for the temperature – many monitor chambers become quite hot during use because of their proximity to the focal spot. What is more of a problem, however, is the fact that the chamber temperature can change markedly during the course of the day. There is, therefore, great difficulty in knowing what the temperature of the chamber really is. It is therefore strongly recommended that a sealed (airtight) chamber should be fitted and every effort should be made to do this. If only an unsealed chamber is available, the necessary correction is effected by using the expression

$$m'_{p,t} = m'_{P,T} / \phi(p,t)$$

where $m'_{p,t}$ is the number of monitor divisions which will, if the pressure and temperature of the gas in the chamber are p and t respectively, result in the same surface exposure (or dose) as would $m'_{P,T}$ monitor divisions if the pressure and temperature of the gas in the chamber are P and T respectively. If T and t are measured in degC, then

$$\phi(p,t) = \frac{P}{p} \times \left(\frac{273 + t}{273 + T}\right)$$

For example: An unsealed (not airtight) monitor chamber has been calibrated (see Chapter 4 for details of how this is done) so that a particular treatment necessitates an associated monitor reading of 500 divisions if the temperature and pressure of the monitor chamber are 22°C and 760 mm Hg respectively. The treatment is required to be delivered, however, when the temperature and pressure of the gas in the monitor chamber are known to be 30°C and 750 mm Hg respectively.
The required reading in the monitor is therefore

$$500 / \left[\frac{760}{750} \times \left(\frac{273 + 30}{273 + 22}\right)\right]$$

$$= 500 \times \frac{750}{760} \times \frac{295}{303}$$

$$= 480 \text{ divisions}$$

It should be noted particularly that the correction to be applied is the reciprocal of that which would be applied to an ionization chamber reading which is being used to measure the exposure (see section 3.3.1). This fact

is useful in helping to determine whether or not the monitor chamber is airtight. In a later chapter (section 4.5.1) the routine checking of the output of the X-ray equipment is described. When making this check for equipment fitted with a monitor system, a note should be made of:

(a) The consistency check reading per monitor division
(b) The consistency check reading (without the usual correction for ambient pressure and temperature being applied) per monitor division.

Assuming that the ionization chamber used for the routine check is unsealed, as is usually the case, then from day to day as the ambient pressure and temperature vary (a) will be more constant than (b) if the monitor chamber is sealed. Conversely if the monitor chamber is unsealed, (b) will be more constant than (a). On the other hand, if the ionization chamber used for the routine check is of the uncommon, sealed kind, the reverse argument applies.

2.2.2.2. Automatic termination of irradiation

It is sometimes arranged for the required number of monitor divisions to be set on the monitor system in the same way as the exposure time is set on the timer. In this case the exposure is automatically terminated after the required (set) number of monitor units have been delivered. Clearly this is a very satisfactory system for ensuring that the required given dose is delivered in spite of variations in output rate of the X-ray equipment.

It is, of course, still necessary to ensure that the correct number of divisions are preset on the monitor and that the monitor system functions correctly.

2.2.3. Timer and monitor errors

In practice the actual exposure time will not be exactly the same as that set on, or indicated by, the timer because of the finite time of operation of the switching ON and OFF mechanism. In particular the switching ON can be slow (in some cases as much as 30 seconds!) even though the switching OFF is usually quick. The magnitude of the difference between the actual and the 'set' time can be determined by the method described in section 4.2.3.1. Fortunately this difference is typically about 1 second or less. Nevertheless, it is important to take account of this when making measurements, even though it is usually negligible compared with the exposure times used for treatment, which are usually of several minutes duration. It is essential, however, to confirm that this is so since some high-output equipment can have appreciable timer end errors which are not negligible compared with the relatively short exposure times.

The timer may be either mechanical (clockwork) or electrical. In both cases the accuracy of the timer should be checked regularly. This is particularly important for a synchronous electric timer if the mains electricity supply is subject to frequency variations. A good wrist watch (accurate to one or two minutes per day) is sufficiently accurate for checking times in the range of five to sixty minutes. It is particularly important to check the linearity of the timer. By this is meant that (starting and stopping errors allowed for) the radiation delivered in an exposure time of say M minutes is exactly M/m times as great as that delivered in m minutes where

m is the order of time used for the measurements (Chapter 4) and M is the order of the time used in a treatment.

For example: For a particular X-ray machine the starting and stopping error is such that radiation is delivered effectively for 2 seconds less than the time indicated by the timer (the timer error is said to be -2 s). If the time used for the measurement of the surface dose rate is 1 minute whilst the time for the average treatment is of the order of 10 minutes, the amounts of radiation delivered for timer settings of 1 min 2 s (i.e. a true 1 min) and 10 min 2 s (i.e. a true 10 min) must be in the ratio of 1:10, and similarly for other values of the treatment time.

The above comments apply equally to a dose monitor system. Effects associated with the ON and OFF periods are dealt with in exactly the same way. It is, of course, equally necessary to test the monitor system for linearity and for stability. Also its sensitivity must be independent of mains voltage and frequency variations, at least over the maximum likely range. Testing for mains voltage variations is easily effected by deliberately changing the input voltage to the monitor electronics by means of say, an autotransformer. Since it is not possible to change the input frequency, all that can be done is to be on guard for possible errors if the mains frequency is known to vary. This latter comment applies equally to the timer.

The monitor sensitivity should be independent of the magnitude of the exposure (or dose) rate or, at least, its known dependence on exposure (or dose) rate taken into account.

CHAPTER 3

THE RADIATION MEASURING INSTRUMENT (DOSEMETER)

3.1. ABSORBED DOSE AND EXPOSURE

Absorbed dose (D). Absorbed dose is a measure of the amount of energy deposited at the point of interest in the material situated at this point. The unit of absorbed dose is the rad (see Appendix II for the exact definitions).

Exposure (X). Exposure, as used in radiotherapy, is a special concept, the definition of which is given in Appendix II. The unit of exposure is the röntgen. Exposure can be conveniently regarded as an indirect measure of absorbed dose in air as well as being a measure of the amount of radiation incident upon the point of measurement.

The relationship between the absorbed dose (D) and the exposure (X) which gives rise to the absorbed dose is

$$D = f_{\lambda} \cdot X$$

where X is the exposure (in röntgen) at the point of interest; D is the corresponding absorbed dose (in rad) at this point; and f_{λ} is the rad per röntgen conversion factor whose value depends on the quality (HVT) of the radiation and the kind of material being irradiated. Values of f_{λ} for soft tissue, soft tissue within bone and for bone are given in Table III and repeated in Appendix V. AVII.

TABLE III. VALUES OF RAD PER RONTGEN CONVERSION FACTOR (f_{λ})

Radiation quality HVT	f_{λ} Soft tissue	f_{λ} Soft tissue cavity inside bone [a]	f_{λ} Compact bone [a]
0.5 mm Al	0.92	3.5	4.4
1.0	0.93	3.3	4.0
2.0	0.93	2.9	3.5
3.0	0.93	2.3	2.0
1.0 mm Cu	0.93	2.0	1.5
1.5	0.94	1.8	1.3
2.0	0.95	1.6	1.1
2.5	0.95	1.35	1.05
^{137}Cs γ-rays	0.96	1.15	0.92
^{60}Co γ-rays	0.96	1.05	0.92

[a] The values of f_{λ} for soft tissue within bone and for compact bone are very dependent upon the radiation spectrum and, in the case of the soft tissue within bone, also on the cavity size, especially for the softer qualities. The values given here are therefore approximate and intended only to indicate the order of magnitude and refer to cavity sizes in the range 10-50 μm which are associated with 'bone' necrosis.

Unfortunately, although the word 'exposure' has a strictly defined meaning, it is also a word in common use in radiology and it is often not clear whether the word is being used in its specially defined sense or in a general sense. In this manual the expression 'exposure (röntgen)' will be used whenever there is doubt. Usually the scale of an ionization chamber instrument is calibrated in exposure (röntgen). However, since exposure and absorbed dose are related to each other it is common practice to refer to a 'dosemeter' when strictly speaking 'exposure meter' is intended. In this manual the term dosemeter will be used in this sense. Similarly 'dose rate meter' is used when strictly 'exposure rate meter' is intended.

Although it is recommended that statements of radiotherapy treatment doses should be in terms of absorbed dose (rad), the only available radiation standards are of exposure (röntgen). It is therefore usual to determine, by measurement, the exposure and deduce the corresponding absorbed dose by means of the equation given at the beginning of the section. The value of the conversion factor f_λ, as stated above, depends on the kind of material being irradiated (e. g. soft tissue, bone) and on the radiation quality at the point of interest. The quality at a depth inside the patient is softer (i. e. lower HVT) than the quality of the primary beam, due to the presence of scattered radiation. The radiation quality at the point of interest is hence determined by the HVT of the primary radiation incident upon the patient, the size of the beam and the depth. The values of f_λ quoted in Table III are for a medium beam size and depth. Since the value of f_λ for soft tissue varies only slightly with radiation quality, the values given can be used with confidence for all beam sizes and depths. The values of f_λ given for bone and for soft tissue within bone are illustrative only since the precise value can be determined only if the radiation spectrum is well known.

3. 2. THE MEASURING INSTRUMENT

Although many different types of measuring systems exist, there is no doubt that the one which uses an ionization chamber is usually the most satisfactory for the kind of measurements with which this manual is concerned. Other types, for example the lithium fluoride and ferrous sulphate systems, have their place in special circumstances, but are outside the scope of this manual.

It is essential for every radiotherapy centre to possess a dosemeter which has been calibrated over the whole range of radiation qualities to be used. If a wide range of qualities is involved, then probably more than one kind of ionization chamber will be required. Good practice demands that each centre should have available, for each of the radiation qualities being used, more than one calibrated dosemeter. The cost and maintenance of such instruments is very small compared with the other costs of radiotherapy. There should also be available a radiation source (of known half-life) with which to check the constancy of the instrument (see section 3. 5. 2 for more details).

3. 2(a). The standard dosemeter

One of the dosemeters, the so-called 'local standard', should be used only for comparing with the others (see below) and for making special measurements. It is not to be used routinely. Its function is to provide

the standard upon which the consistency of patient dosage is based. The local standard instrument should spend most of its time in a cupboard and should be used with extreme care so that it is not damaged in any way. It is, of course, this local standard instrument which is sent to the standardizing laboratories for its own calibration. The local standard dosemeter need not be (in fact it is preferable that it should not be) a multi-range, or dual-purpose (exposure, exposure rate) instrument. Ambiguity can be avoided if the instrument is a direct reading dosemeter without range-change facilities, and has a full-scale deflection equivalent to 50-200 R.

3.2(b). The routine dosemeter

Although by no means necessary, it is useful for the dosemeter for routine use to be a multi-range instrument so that both high and low values of dose may be measured using a convenient (i.e. 1-2 min) exposure time. It is helpful if this instrument can also be used to measure dose rate directly. This is especially so for caesium and cobalt teletherapy equipment for which direct measurement of the (steady) dose rate can be very time saving. For X-ray equipment, however, it is often not possible to make direct measurements of dose rate (exposure rate) because the rate is not sufficiently constant for rate values to be useful and an instrument which measures dose (or exposure) is required.

To select a suitable instrument and, having selected it, to use it in a satisfactory manner, a number of factors must be considered. These concern principally the chamber size, the linearity of instrument response and the variation of sensitivity with radiation quality. Most of the requirements are general and apply to instruments intended for use over almost the entire range of radiation qualities commonly used in radiotherapy. The main exception concerns those instruments to be used for the measurement of low energy X-rays, the so-called 'Grenz rays'. For these radiations there are some extra requirements which are dealt with separately later (section 3.4). The major part of this chapter is directed to instruments intended for use with radiation generated at about 50 kV and above.

3.2.1. Chamber size

It is important that the size of the ionization chamber should not be too large. Ideally the air cavity should be as small as possible but in practice an internal diameter of up to (but not larger than) 1 cm is acceptable. There are two reasons for requiring a small chamber:

(a) If the chamber is to be used inside a water or solid phantom, the presence of the chamber will change the value of the exposure (röntgen) which it is desired to measure. The magnitude of this effect, which is called the 'displacement effect', increases with the diameter of the chamber. As will be seen later, a correction can be applied but it is still important to minimize the effect. For chambers of up to 1 cm internal diameter the effect can be neglected for X-rays over the quality range 0.5 to 3.0 mm Cu HVT. For ^{60}Co and ^{137}Cs γ-rays the correction amounts to nearly 2%.

(b) If the measurement is to be meaningful, the pattern of radiation over the volume occupied by the chamber must be approximately uniform. A chamber of diameter 1 cm and length up to 2-3 cm is acceptable for the kind of measurement with which this manual is concerned.

The size of the local standard ionization chamber is not of prime importance since it can usually be arranged to irradiate it with a beam which is sufficiently uniform over the volume of the chamber. If the chamber is large, however, the intercomparison between it and the routinely used instrument will have to be done 'in air' which is less preferable than is the method described below (section 3.5.6). If possible, therefore, both the standard and the routinely used instruments should have chambers of less than 1 cm diameter.

3.2.2. Instrument linearity

The design and construction of the instrument should be such that the response is linear so that equal increments in exposure result in equal increments in the reading. In other words, doubling the exposure should give double the instrument reading. It is usual to require that the ratio

$$\frac{\text{Exposure}}{\text{Reading}}$$

is constant within ± 1% over the working range of the instrument.

3.2.3. Variation of sensitivity with photon energy or quality dependence

To measure exposure (röntgen) the ionization chamber must be constructed of suitable materials ('air equivalent') and have walls of suitable thickness. The thickness required depends upon the quality of the radiation to be measured. In principle, therefore, it is impossible to construct a thimble chamber which has the same response to equal exposures of radiations of different qualities. The manufacturer takes care, however, to provide a chamber whose variation in response is small over the range of radiation qualities for which the chamber is intended to be used. This is desirable since it is difficult in practice to know exactly the quality of radiation being measured. It is therefore important to choose a chamber appropriate to the range of X-ray quality of interest. Figure 2 shows the variation of response (i.e. of calibration factor – see section 3.3) with radiation quality for three typical chambers which are intended for use with (a) low kV X-rays, (b) medium kV X-rays and (c) high kV X-rays.

It is important that the presence of the chamber and its immediate connecting parts should not change the value of the exposure being measured. For this reason a chamber with a massive metal stem needs to be used with caution for measurements in water.

3.2.4. Electronic equilibrium

The thickness and composition of the wall of an ionization chamber are the major factors which control the variation in sensitivity with radiation quality. In particular the thickness of the chamber wall must be sufficient to provide electronic equilibrium (i.e. full build-up of secondary electrons) but not so thick as to cause excessive attenuation of the X-radiation. Figure 3 shows how the response of an ionization chamber to a given exposure (röntgen) will vary as the thickness of the chamber wall is increased. For correct measurement the wall thickness needs to be at least equal to d_{min}.

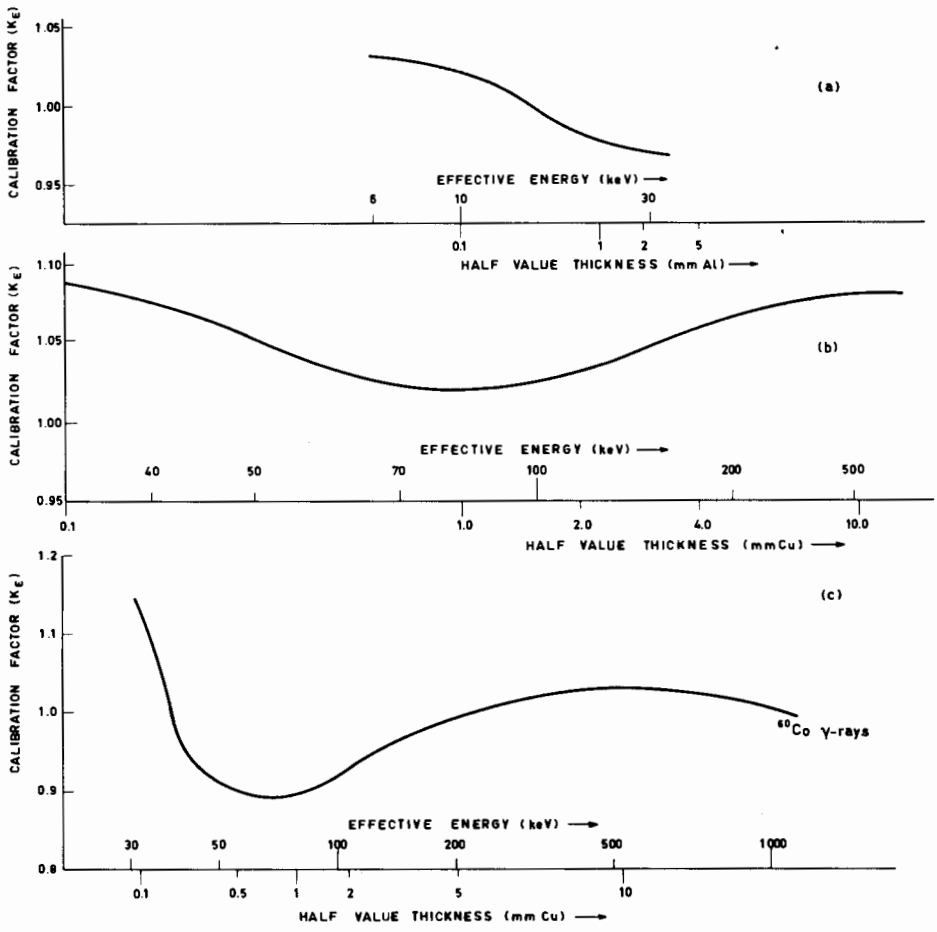

FIG. 2. Variation in response of ionization chamber with radiation quality for ionization chambers designed for use with (a) low kV X-rays; (b) medium kV X-rays; and (c) high kV X-rays.

If a chamber, designed for use over the range 1-3 mm Cu HVT is also to be used for ^{60}Co γ-rays, it is necessary to increase the wall thickness and this is usually done by fitting an equilibrium cap (often called a 'build-up cap'). This is a close fitting sleeve of Perspex[2] about 4 mm thick (Fig. 4) and is to be regarded as an integral part of the chamber for ^{60}Co γ-ray measurements.

Alternatively, some workers prefer to use a chamber that has been specially constructed for ^{60}Co γ-ray work and has an adequately thick wall. Such a chamber does not require a 'build-up' cap and is not suitable for measurement of X-rays of energy lower than 1 MV. For ^{60}Co γ-rays the thickness of the chamber wall or of the wall plus equilibrium cap should be in the range 500-800 mg cm^{-2} (i. e. 5-8 mm of unit density material).

[2] Perspex (Plexiglass, Lucite) is polymethyl methacrylate.

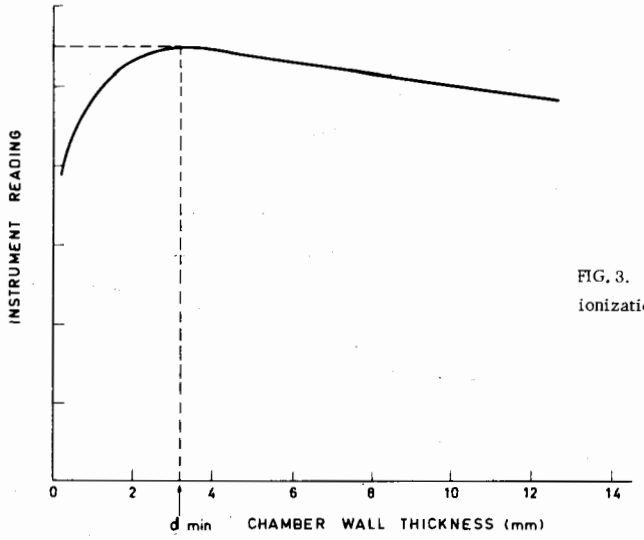

FIG. 3. Variation in response of an ionization chamber with wall thickness.

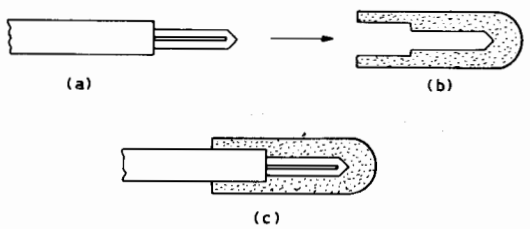

FIG. 4. Equilibrium (build-up) cap: (a) ionization chamber; (b) equilibrium cap; (c) cap fitted to chamber.

3.3. CALIBRATION OF THE DOSEMETER

In radiotherapy, statements of dosage are made in terms of the well-defined unit of exposure, the röntgen or, more recently and preferably, the unit of absorbed dose, the rad. It is only by using an agreed, well-defined unit that statements of radiation dosage made at various treatment centres throughout the world can be consistent and compatible with each other and that clinical experience therefore can be shared. The national calibration centres are justifiably proud of the accuracy of their standards. However, it must be pointed out that it is their mutual agreement and consistency over time which are the important features. Although manufacturers of dosemeters attempt to supply an instrument that, when used properly, gives directly a measurement of the exposure (röntgen), it is usually necessary to apply a small correction to the reading in order to obtain the correct value of the exposure (röntgen). The relationship between the instrument reading (I) and the corresponding value (X) of the exposure is given by

$$X = I \cdot K_E$$

where K_E is known as the calibration factor, the value of which is dependent on the radiation quality. The resulting value of exposure (X) is that at a

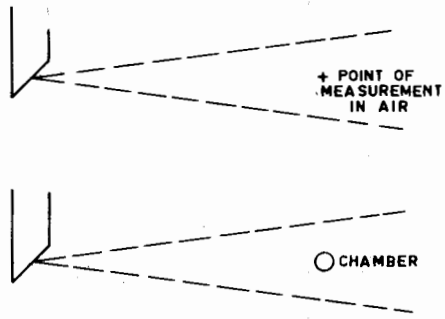

FIG. 5. The meaning of an exposure calibration.

point in air coincident with the geometric centre of the chamber, but in the absence of the chamber (Fig. 5).

Each dosemeter (i. e. ionization chamber + associated electronics and meter) has its own individual values of K_E. Table IV lists typical values of K_E for one particular ionization chamber. Other instruments will have, in general, different values.

The values of K_E quoted for ^{60}Co γ-rays and for 2-MV X-rays apply to the chamber when fitted with its build-up cap, whereas those quoted for the X-ray qualities over the range up to 3 mm Cu HVT apply to the chamber without the build-up cap.

The radiation qualities for which the calibration factors are quoted will not be exactly identical with those for which the factor is required in practice. It is usual to obtain the value of K_E by interpolation. For this

TABLE IV. TYPICAL VALUES OF CALIBRATION FACTOR K_E FOR A TYPICAL MEASURING INSTRUMENT

The values of K_E applicable to any instrument are obtained by calibration. The values given here are for purposes of illustration only.

Radiation quality Half-value thickness	Effective Energy (MeV)	K_E at 760 mm Hg, 22°C
0.5 mm Al	17.4	1.13
0.8 mm Al	20.8	1.09
1.2 mm Al	24.0	1.06
1.6 mm Al	26.6	1.03
0.1 mm Cu	33.1	1.01
0.2 mm Cu	41.4	1.00
0.5 mm Cu	60.6	1.00
0.9 mm Cu	78.0	1.01
2.0 mm Cu	113	1.02
3.0 mm Cu	138	1.03
2-MV X-rays	800	1.06 [a]
^{60}Co γ-rays	1200	1.06 [a]

[a] When chamber is fitted with 4.6 mm Perspex build-up cap.

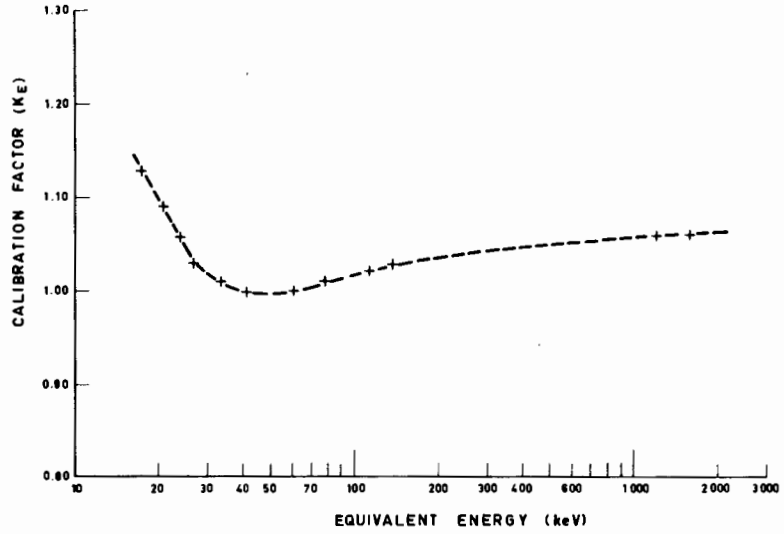

FIG. 6. Variation of calibration factor K_E with equivalent voltage.

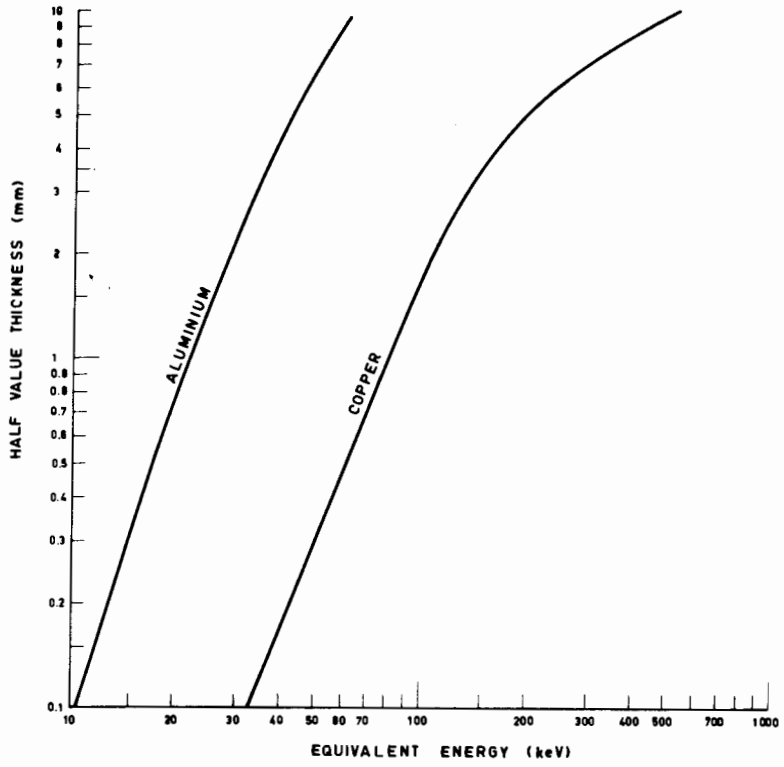

FIG. 7. Relationship between equivalent kilovoltage and half-value thickness.

it is useful to plot the value of the calibration factor against the equivalent effective energy and Fig. 6 shows the data of Table IV plotted in this way. Figure 7 shows the relationship between the equivalent kilovoltage and half-value thickness (in mm of copper or aluminium).

For example, for the particular chamber to which the data of Table IV and Fig. 6 apply it is required to know the calibration factor appropriate to 1.4 mm Al HVT. Figure 7 shows that the equivalent kilovoltage is 25 keV and hence, from Fig. 6, the required calibration factor is 1.045.

It should be noted that the change in calibration factor with X-ray quality is comparatively large in the low photon energy region. When making measurements of low-energy X-rays, particular attention must be given to the estimation of the HVT so that the appropriate values of the calibration factor, K_E, can be employed.

3.3.1. Temperature and pressure effects

Since, normally, the ionization chamber is not airtight, the instrument reading associated with any exposure (röntgen) will depend upon the atmospheric pressure and temperature of the air in the chamber because these control the mass of gas in the chamber. The calibration factor K_E is, therefore, directly applicable only at a stated pressure and temperature. If measurements are made at pressures and temperatures different from the stated standard values, it is necessary to apply a correction of the form:

$$I_{P,T} = I_{p,t} \cdot \phi(p, t)$$

where

$$\phi(p, t) = \frac{\text{Standard pressure}}{\text{Ambient pressure}} \cdot \frac{\text{Ambient temperature} + 273}{\text{Standard temperature} + 273}$$

$$= \frac{P}{p} \cdot \frac{(t + 273)}{(T + 273)}$$

and $I_{P,T}$ is the corrected instrument reading, i.e. that which would have been obtained if an identical exposure had been given at the standard pressure and temperature; $I_{p,t}$ is the instrument reading obtained; P and T are the standard pressure and temperature values; and p and t are the ambient pressure and temperature values existing in the chamber at the time of measurement. P and p, can be measured in any convenient units (mm Hg or millibar); T and t are measured in degrees centigrade and $\phi(p, t)$ is known as the pressure and temperature correction factor.

It is modern practice to adopt a pressure of 760 mm Hg (1013 millibar) and a temperature of 20°C,[3] as standards. In these circumstances the formula given above takes the form:

$$I_{P,T} = I_{p,t} \cdot \frac{760}{p} \cdot \frac{t + 273}{293}$$

If the value of the atmospheric pressure is obtained from a local weather bureau, care must be taken to use the value applicable to the

[3] A standard temperature of 22°C is sometimes used.

location of the chamber and not the value corrected to mean sea level which is the quantity usually quoted by weather stations.

When the instrument is moved to a different environment, sufficient time must be allowed for the air in the chamber to equilibrate to its surroundings. The time constant for this can be quite large since although not deliberately sealed the chamber may be almost airtight.

In later chapters of this manual it is recommended that measurements should be made inside a water or plastic phantom. It is the temperature of the phantom to which the chamber will equilibrate and which is therefore used in the above correction formula. It is good practice to keep the ionization chamber in or with the phantom in which the measurement is to be made for half to one hour (or longer if convenient) before measurements are commenced.

There are three kinds of instruments which do not require the use of a correction factor for temperature and pressure and to which, therefore, no correction should be applied. These are:

(a) Ionization chambers which are sealed (airtight) so that a fixed mass of gas is contained within the chamber

(b) Instruments which are fitted with a built-in standardizing ionization chamber and radioactive reference source. With this type of instrument the sensitivity is adjusted until the signal from the built-in standardizing chamber produces a specified instrument reading (scale deflection). The effect of temperature and pressure is therefore compensated for and no further correction factor is necessary. It must be noted that this system relies on the temperature and pressure at the measuring chamber being the same as at the standardizing chamber. This is not always necessarily so since the standardizing chamber is usually inside the instrument case which can be at a distance from and in different temperature conditions from the point of measurement.

(c) Instruments which are fitted with a calibrated sensitivity control on which the value of the correction factor $\phi(p, t)$ can be set. The required correction is thus effected by the instrument.

3.3.2. Humidity

The response of an ionization chamber is also affected by the pressure of water vapour in the air contained within the chamber. The instrument reading observed requires to be increased to compensate for the presence of water vapour by an amount which depends upon the relative humidity and the temperature. This is in addition to the pressure and temperature correction $\phi(p, t)$ already referred to above. Fortunately, the magnitude of the correction is usually so small as to be negligible. Even if the humidity and temperature are high out of doors, it is likely that the room in which the measurements are being made will be air-conditioned. The maximum likely correction is less than 1% for all values of atmospheric pressure.

If conditions of high humidity are common, it is worthwhile storing the ionization chambers in a desiccated container to avoid electrical leakage, a situation which is much more serious than is the need to apply a small correction for humidity. Of course, if the ionization chamber is sealed, no correction for humidity is necessary and the possibility of leakage is substantially reduced.

3.3.3. Calibration: the determination of K_E

In many countries there are national or other primary standardizing laboratories at which facilities are provided for the calibration of dosemeters, i.e. for the determination of K_E values. This is done by comparison of the user's dosemeter with what is known as a 'free air chamber' for kilovoltage X-rays, or a standard cavity chamber for ^{60}Co γ-rays or 2-MV X-rays. As a result of this intercomparison at the various radiation qualities requested, values of the factor K_E are quoted as, for example, in Table IV. The various national standardizing laboratories take great care to ensure that their standards are identical with each other. Hence it may safely be assumed that, to the accuracy needed, a dosemeter calibrated at any standardizing laboratory would be given the same set of calibration factors at another.

There exist also regional standardizing centres (and more of these are likely to be established in the near future) at which dosemeters can be calibrated by comparison with a secondary standard instrument which has itself been calibrated at one of the primary standardizing laboratories. It should be noted that a few manufacturers also have a free air standard ionization chamber or secondary standard instruments and offer a calibration service.

It cannot be overemphasized that one or other of these calibration facilities should be used and each radiotherapy centre must make every effort to do so. It may be, however, that access to a suitable standardizing service cannot be obtained or that there is an unacceptable delay in obtaining the service. In such circumstances the user must obtain a calibration indirectly. This is best done by comparing the instrument with a calibrated one using the technique described in section 3.5.6. The most common use of this latter technique is for the purpose of calibrating the routinely used dosemeter by comparing it with the local standard dosemeter which has been calibrated at one of the standardizing laboratories. Another use is to obtain a calibration by comparison with a calibrated dosemeter located at another nearby or more easily accessible radiotherapy centre.

In some circumstances it may be impossible to obtain even such an indirect calibration. A solution to this problem is to make use of either a national, international or other postal dose intercomparison service.

Up to date information about all the calibration services available may be obtained from:

(a) Dosimetry Section,
Division of Life Sciences,
International Atomic Energy Agency,
Kärntner Ring 11, A-1010 Vienna, Austria.

(b) The Radiation Health Unit,
World Health Organization,
1211 Geneva 27,
Switzerland.

3.3.3.1. Accuracy

It is clear that the most accurate value of the calibration factor, K_E, for any radiation quality is obtained by using the calibration services of one of the primary standardizing laboratories. Although necessarily of a

slightly lower standard of accuracy, the calibration services of the secondary standardizing laboratories are to be regarded as those which will be used by most radiotherapy centres and are of adequate accuracy. The accuracy of a calibration obtained by the technique of section 3.5.6 or, particularly, by a postal dose intercomparison service is very dependent on the experience of the workers and the amount of effort expended. Although of great value on many occasions, these latter methods should be used only to obtain temporary calibration factors pending a calibration by a standardizing laboratory.

3.3.3.2. Frequency of calibration

It is desirable that the dosemeter should be calibrated by the standardizing laboratory at intervals of about two or three years. This is particularly necessary for radiation qualities of less than 3 mm Cu HVT where changes in the chamber which are not detected by radioactive check (section 3.5.2) can affect the calibration. For ^{60}Co calibrations the interval may be longer if the radioactive check shows no significant change.

In the intervals between calibrations tests (which are detailed in section 3.5) must be performed to confirm that no gross change in sensitivity and therefore in calibration factor (K_E) is occurring. In this connection regular radioactive standard checks (section 3.5.2) are imperative. If any of the tests raise doubts, a calibration may have to be repeated before the two-yearly period is over.

3.4. SPECIAL REQUIREMENTS FOR LOW-VOLTAGE X-RAYS (GRENZ RAYS)

It has already been commented that it is important to choose an ionization chamber that is suitable for the quality range of interest. This is particularly so for low-voltage X-rays. The normal type of thimble chamber is acceptable down to about 1 mm Al HVT but below this the wall thickness needs to be very small in order to be sure that the chamber is measuring the radiation properly. For this range of quality – the so-called Grenz-ray range – it is usual to use a parallel plate chamber with a very thin front wall or even no wall at all.

The calibration of such a chamber also calls for special care and it should be done at approximately the same exposure rate as that to be measured since it is difficult to achieve full saturation at the high dose rates encountered.

3.5. CARE AND MAINTENANCE OF THE DOSEMETER

The instructions given in the manufacturer's handbook accompanying the dosemeter should be studied and followed. The user must check that the instrument is behaving in a sensible manner. Dosemeters, although moderately rugged, are sensitive scientific instruments and should be treated as such. In the event of any misbehaviour, no attempt should be made to dismantle the instrument (especially the chamber) and to attempt a repair unless the user is sure of his competence. It is preferable to

consult the manufacturer or other known expert. When not in use the instrument should be kept in a closed cupboard and handled only by qualified persons. In humid environments it is worthwhile storing the chamber in a closed container (e. g. a glass jar or even a Polythene bag) containing a desiccant and, of course, the desiccant incorporated in the electronic system should be maintained in an active condition.

3. 5. 1. Leakage testing

The instrument should be tested regularly for electrical leakage. This is simply done by switching ON and setting up the dosemeter (instrument plus chamber) exactly as if a measurement were to be made but in fact the chamber is not subjected to any radiation. The testing time under these 'no radiation' conditions should be several times (say 10 times) that which is required for each exposure measurement. The leakage should be sufficiently small that its value is substantially less than half a per cent (0. 5%) of the reading obtained due to the radiation, e. g. an exposure of about 1 minute duration is to be measured and the interval between the instrument being set to zero reading before the exposure starts and the instrument reading being noted after the exposure is over is 2 minutes. In other words, the time interval for which there is a possibility of leakage occurring is 2 minutes, although the duration of the exposure is only one half of this, i. e. 1 minute. An instrument reading of about 50 divisions is expected. When leak-tested over 20 minutes (i. e. 10×2 minutes) the instrument shows a deflection of three divisions. This leakage represents

$$\frac{3}{20} = 0.15 \text{ division per minute}$$

or

$$0.15 \times 2 = 0.3 \text{ division per 2 minutes}$$

which is $(0.3/50) \times 100 = 0.6\%$ of the expected reading.

This is to be regarded as excessive. If the leakage had been 1 division in 20 minutes, then this would represent

$$\frac{1}{10 \times 50} \times 100 = 0.2\% \text{ of the expected reading}$$

and this amount of leakage is just acceptable.

If the dosemeter is found to have higher leakage, it is desirable to attempt to cure it. Whether the source of the leak is in the chamber or the electronic part of the dosemeter can be determined by detaching the chamber (together with its attached cable if necessary) and testing the electronic part separately in the same way. The observed magnitude of the leak may be very different when the chamber is disconnected and this value must not be used to effect compensation for leakage. It is worthwhile observing and recording the leakage for the instrument with and without the chamber connected, when the instrument is in good working order. If and when any significant leakage arises subsequently, it is then more easy to recognize in which part of the instrument the trouble lies. Fortunately, when the

instrument and chamber are properly looked after, trouble due to leakage does not often occur.

An apparent leakage in the reverse direction can also sometimes occur, i. e. the instrument meter reading falls progressively with the passage of time. This effect is usually due to incorrect voltages being applied to the various stages of the associated electronic circuit or to a faulty valve (tube). Replacement of the batteries in a battery-operated instrument or of the valve will usually eliminate the problem.

If it is not possible to cure a leak, it is sometimes possible to make an allowance for it. However, this must not be done if the rate of leak is substantially variable with time, i. e. the allowance can be made only when the observed leakage is proportional to the leakage time. The reverse type of leak referred to above is often variable with time. The leakage should also correspond to less than 5% of the reading due to the radiation exposure unless the instrument is being used in circumstances where only a low accuracy is necessary (e. g. in protection measurements) when rather larger leaks are tolerable.

Compensation for leakage is effected by subtracting from (or adding to, for a reverse leak) the observed instrument reading (produced by radiation plus leakage) an amount equal to the reading which would have been obtained due to leakage alone. For example, an average instrument reading of 52. 3 divisions is obtained as a result of a series of exposures for which the time interval between the instrument reading being set to zero and the reading being noted is 2 minutes. When tested for leakage under 'no radiation' conditions a deflection of 4 divisions in 20 minutes immediately before and of 3. 8 divisions in 20 minutes after the series of exposures is observed. This leakage corresponds to an average of 3. 9 divisions in 20 minutes or 0. 39 divisions in 2 minutes. The corrected instrument reading is hence

$$52.3 - 0.39 = 51.9$$

when stated to one decimal place of accuracy.

3. 5. 2. Standard radioactive check

It is essential to confirm that the numerical values of K_E continue to apply to the instrument, i. e. to check that the overall sensitivity of the instrument to radiation remains constant. This is most easily done by subjecting the chamber to an accurately reproducible (but perhaps unknown) amount of radiation both on the occasion when the K_E values are being determined and at regular intervals subsequently. The reading so obtained on the instrument should be constant. If it is not constant, then something is amiss and new values of K_E must be obtained by either direct or indirect calibration. The most suitable way of delivering an accurately reproducible exposure is to use a fixed arrangement of radioactive sources, of known half-life, relative to which the dosemeter chamber can be placed in an accurately reproducible position. Figure 8 shows a diagram of a suitable radioactive checking device which in this instance uses strontium-90. This radioisotope is particularly suitable because of its long (28 yr) half-life. The data of Table V indicate how, after correcting for radioactive decay, the instrument reading per 100 seconds is consistent to better than

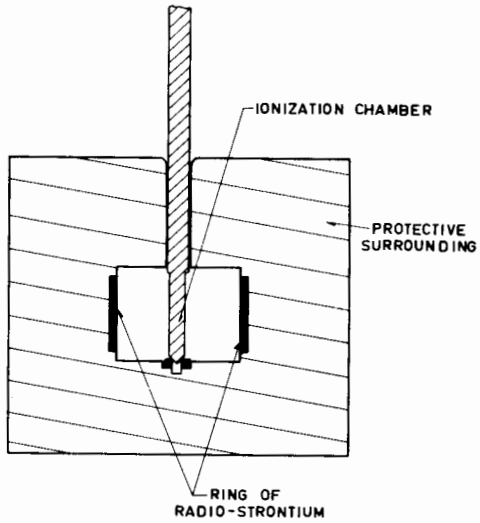

FIG. 8. Diagram of radioactive strontium check.

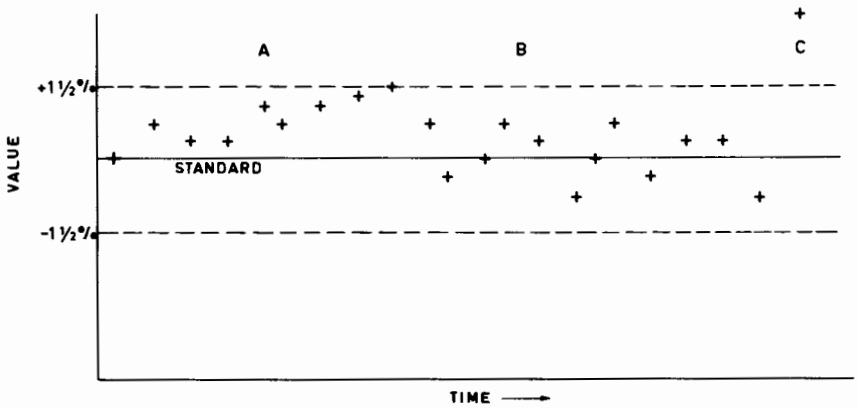

FIG. 9. Chronological plot of results of standard radioactive check.

± 1%. This is an entirely satisfactory situation and is the one usually encountered, provided the instrument is carefully handled and is not accidentally damaged.

An alternative method is to use a telecobalt machine if one is available. The chamber is held at an accurately reproducible distance from the source and the collimator set to a fixed value. The exposure may be made either in air, in which case the chamber must be fitted with its build-up cap, or inside a solid phantom such as described in section 3. 5. 6. Readings of the instrument resulting from a series of exposures of reproducible duration are noted, corrected for radioactive decay and tabulated exactly as for the strontium-90 device. Of course, care must be taken that any changes observed are not due to changes in the telecobalt machine.

It is useful to plot the values of the radioactive check chronologically as they are obtained (Fig. 9). From this kind of plot any tendency of the value to drift steadily up or down (as at A in Fig. 9) is obvious, as is any tendency for the values to be randomly distributed (as at B) or any sudden

TABLE V. EXCERPT FROM RECORD OF STANDARD RADIOACTIVE CHECK

Date	Mean time for 30 divisions i.e. 5 to 35 div.	Temp (°C)	Pressure (mm Hg)	Decay factor	Corrected value (div/100 s)	Comments (see notes)
1. 1.62	83.4	22	760	1.000	36.0	1. Base level
9. 3.65	90.1	24	765	1.079	35.9	
14. 6.65	93.2	26	762	1.086	35.2	2. recheck
15. 6.65	92.2	22	754	1.086	35.8	3. recheck
16. 6.65	91.9	22	748	1.086	35.9	4. confirmed
10. 9.65	91.0	16	748	1.092	35.8	
3. 1.66	92.4	22.5	757	1.101	36.0	
11. 3.68	97.2	24	763	1.165	36.1	
19. 6.68	98.5	22	755	1.173	36.1	
21. 8.68	96.9	20.5	764	1.179	36.0	
15.11.68	97.0	20.5	764	1.186	36.2	
11.12.68	97.0	19	764	1.187	36.1	

Corrected value = Reading which would be obtained at 22°C, 760 mm Hg in 100 s on 1.1.1962, due allowance having been made for radioactive decay.

$$= \frac{30}{\text{mean time}} \times 100 \times \phi(p,t) \times \text{Decay factor}$$

E.g. on 11.3.68:

$$\text{Corrected value} = \frac{30}{97.2} \times 100 \times \frac{760}{763} \times \frac{297}{295} \times 1.165$$

$$= 36.1 \text{ division}/100 \text{ s}$$

1. This is the reading obtained on the occasion of the calibration of the measuring instrument.
2. The reading is different by 2%; therefore recheck next day to determine whether spurious or real.
3. The reading is not significantly different from the base level; recheck to confirm.
4. Value confirmed.

unexpected change in value (as at C). When provided in this way with a clear history of the radioactive check the user is better able to know whether anything is seriously amiss or not. Further he can be prepared to take action needed as a result of a steady drift. The broken lines in Fig. 9 are drawn at $\pm 1\frac{1}{2}\%$ of the value obtained at the time the instrument was calibrated and are used as a guide to the need for action. If the check value remains within the $\pm 1\frac{1}{2}\%$ range, then no action need be taken and the original values of K_E will be used. If the value moves outside this range, a recheck should be done. If this confirms that a change of greater than $1\frac{1}{2}\%$ from the original value has occurred, it is necessary to investigate the situation further.

First, it is advisable to try to decide whether the change is due to a change in the ionization chamber or in the electronic part of the system. The electronic part can often be checked by means of a standard voltage source either built into the instrument or as an additional component. If it is shown that the change is due to a change in electrical parts of the system, then it is reasonable to change all the K_E values by exactly the same percentage as found by the radioactive check and electrical check.

If the change is not due to the electronic part, then it must be due to a change in the chamber and it is not safe to assume that the same magnitude of change will have occurred for all radiation qualities. There is no alternative but to recalibrate the dosemeter either at the standardizing laboratories or by comparison with one of the other dosemeters in the department. Hence the advisability of having at least two dosemeters at each treatment centre, one of which has been calibrated (i. e. is the local standard) and the other(s) compared with it.

3. 5. 3. Collecting voltage

The magnitude of the voltage across the ionization chamber (the collecting voltage) must be sufficiently high so that all the electrical charge produced in the chamber is collected. Figure 10 shows how the observed dosemeter reading resulting from repeated, identical exposures will change as the value of the collecting voltage is changed. A sufficient check is to confirm that the observed readings obtained from the identical exposures do not decrease by more than 1% when the collecting voltage is decreased to half its nominal value. However, such lack of saturation is rarely a problem unless either very high dose rates (greater than 1000 rad/min) or pulsed beams of radiation are being used.

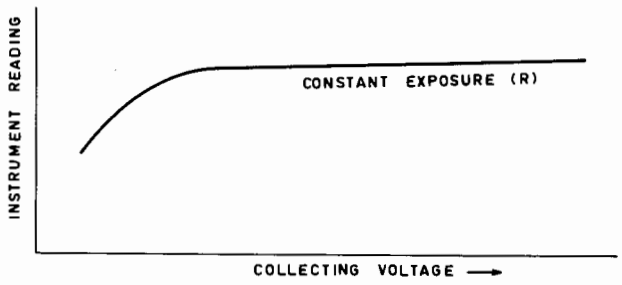

FIG. 10. Effect of variation in ionization chamber collecting voltage.

3.5.4. Radiograph of the chamber

A contact radiograph of the chamber at regular intervals is also desirable. This is to confirm that the internal electrode construction has not moved. Any such movement may not be shown up as a change in sensitivity by the low dose-rate radioactive check but can cause a loss of sensitivity (i.e. the reading is too low) when the chamber is exposed to high dose-rate radiation, particularly if this is pulsed, since there can be recombination of the ions before they have time to move to the electrodes.

3.5.5. Charge sharing type instruments

The checks described above (sections 3.5.1-4) apply equally to the type of instrument having its chamber permanently connected to the remainder of the instrument by means of a cable and to that having a chamber which is detached from the instrument during the exposure. An additional check is, however, recommended for the latter – often called the 'charge sharing or condenser' – type.

For this check the chamber is connected to the measuring system (the electrometer) and their mutual voltage adjusted to the fixed starting value (often marked 0 on the instrument scale). The chamber is then detached from the electrometer, great care being taken that there is no loss of charge from the chamber. The electrometer is now fully discharged, after which the chamber is reconnected to the electrometer and the resulting reading noted. Constancy of this reading indicates the constancy of the electrical sensitivity of the system. If the reading changes by more than 2%, a recalibration is necessary.

It should be noted that this test does not check the constancy of the ionization chamber itself and therefore does not remove the need for the radioactive check (section 3.5.2) and the dosemeter intercomparison (section 3.5.6).

3.5.6. Dosemeter intercomparison

As a regular check and also as demanded by any change in the ionization chamber brought to notice by the radioactive check the various calibrated dosemeters should be compared with each other over a wide range of radiation qualities. The constancy of the intercomparison ratios (i.e. the ratio of the instrument readings obtained for identical exposures) confirms that the sensitivity (K_E value) at that quality is remaining constant for each instrument separately. If the ratio does not remain constant, then it is usually possible, in conjunction with the radioactive check results, to determine which of the instruments is misbehaving.

It is best to effect the comparison between the dosemeters under circumstances that are similar to their intended use. As will be discussed in Chapter 4, their principal use is to determine the dose at a depth inside a tank of water, i.e. inside a scattering medium. A suitable procedure for the intercomparison of two instruments, designated A and B is therefore:

(1) Position the chamber of dosemeter A with its centre on the central axis of a 10 cm × 10 cm beam at a depth of 5 cm inside a water phantom

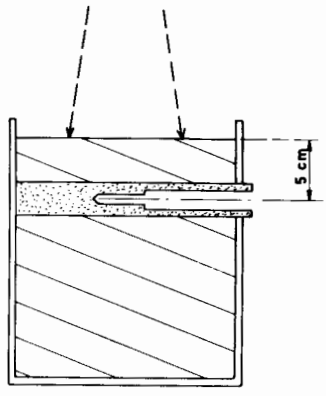

FIG. 11. Ionization chamber at a depth of 5 cm in water phantom.

(Fig. 11). For the present purposes the water phantom needs to have a minimum size of $20 \times 20 \times 20$ cm.

(2) Make an exposure at the desired quality (i. e. HVT) such that a dosemeter reading of about 2/3 full scale is obtained. Note the reading.

(3) Repeat step 2 three times, taking care that all the exposures are identical (i. e. constant kV, mA and exposure time or constant monitor dose). Take the average of these three instrument readings. Let this average by i_A. Although three readings are often sufficient it may be necessary to take more (see section 3. 6); fewer than three are never enough.

(4) Position the centre of the chamber of dosemeter B at exactly the same place and irradiate it exactly as for the other dosemeter, i. e. make three exposures which are each identical to the three previous ones. Let the average reading be i_B.

(5) Steps 1-4 should be repeated at least twice, preferably more. The means of the values i_A and i_B are calculated and let these be I_A and I_B respectively.

(6) If the calibration factors of the two instruments are $(K_E)_A$ and $(K_E)_B$ respectively at the particular radiation quality employed for this intercomparison, then, since both have received identical exposures

$$I_A \cdot (K_E)_A = I_B \cdot (K_E)_B$$

or

$$\frac{I_A}{I_B} = \frac{(K_E)_B}{(K_E)_A}$$

(7) This procedure is repeated for each quality (HVT) of interest and a check made that the ratio of the mean instrument readings (I_A/I_B) is equal to the inverse ratio of their calibration factors, i. e. $(K_E)_B/(K_E)_A$.

If the instruments being compared have very different sensitivities (i. e. K_E values), the experimental technique must be modified slightly. Suppose instrument A has a sensitivity approximately n' times that of the instrument B.

Let n be an integer near in value to n'. The procedure is now:

Steps (1) - (3) as before (see immediately above).

(4) Position the centre of the chamber of dosemeter B at exactly the same place. Make an exposure identical with that given to dosemeter A

and without taking the reading, repeat this exposure n times. Note the total instrument reading due to the n exposures. Repeat the multiple exposure three times. Let the average instrument reading for the multiple exposure by i_B. If n has been chosen correctly the readings i_A and i_B will both be approximately 2/3 full scale.

(5) Steps (1) - (4) should be repeated at least twice, preferably more. The means of the values i_A and i_B are calculated and let these be I_A and I_B respectively.

(6) If the calibration factors of the two instruments are $(K_E)_A$ and $(K_E)_B$ respectively at the particular radiation qualities employed for this intercomparison, then, since dosemeter B has received n times the exposure received by dosemeter A

$$I_A \cdot (K_E)_A \, n = I_B \cdot (K_E)_B$$

or

$$\frac{I_A}{I_B} = \frac{1}{n} \cdot \frac{(K_E)_B}{(K_E)_A}$$

(7) This procedure is repeated for each quality (HVT) of interest and a check made that the ratio of the mean instrument readings (I_A/I_B) is equal to 1/n times the inverse ratio of the calibration factors, i. e. $(1/n)[(K_E)_B/(K_E)_A]$.

It is not essential that every individual exposure made in the intercomparisons described above should be identical. What is essential is that the total exposure and the number of exposures given to the two instruments, A and B, should both be identical (or both in the ratio 1:n). If a monitor system is in use, this is easy to achieve and has the merit that, in so far as the individual exposures can be deliberately made different, the instrument readings for each separate instrument will not be identical. Reading errors will therefore average out. If a monitor systems is not in use, it is better to give identical exposures since the constancy of the separate instrument readings is indicative of the success achieved in making identical exposures.

An alternative, and preferable, procedure which is applicable if the dosemeters are similar (or identical) in size and construction is to irradiate them simultaneously. A phantom capable of holding the two ionization chambers is required. A water phantom is ideal for this type of calibration and a suitable design is shown in Appendix VIII. Alternatively a cubic block of plastic having two holes, into which the ionization chambers fit snugly at about 5 cm below the surface of the block, may be used. Figure 12 illustrates such a block, the use of which removes all uncertainty about reproducibility of position. The exact composition of the material used for this phantom is unimportant since the measurements are purely relative and the chamber positions are interchanged. Likewise, its size is not important although there should be at least 5 cm of material surrounding the radiation beam and 10 cm behind the chambers, both of which should be well inside the beam and clear of the beam edges.

A series of exposures (a) is given, the instrument readings, i_A and i_B, noted and their ratio calculated. The two chambers are then interchanged and a second series of exposures (b) is given. The instrument

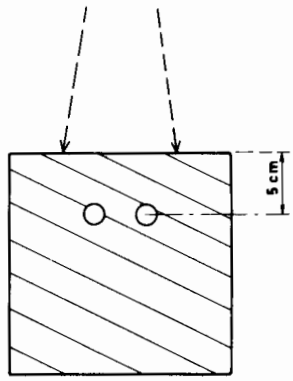

FIG. 12. Two ionization chambers inside a solid phantom for comparison of sensitivities.

readings are again noted and their ratio calculated. As before it is worthwhile interchanging the chambers at least twice and preferably more.

If $(\overline{i_A/i_B})_a$ and $(\overline{i_A/i_B})_b$ represent the mean ratios of the instrument readings in the two series (a and b) of exposures, then it can be shown that

$$\frac{(K_E)_B}{(K_E)_A} = \left[\left(\frac{\bar{i}_A}{i_B}\right)_a \cdot \left(\frac{\bar{i}_A}{i_B}\right)_b\right]^{\frac{1}{2}}$$

where $(K_E)_A$ and $(K_E)_B$ are the calibration factors for the two instruments (A and B) respectively, at the radiation quality employed for the intercomparison. The intercomparison is repeated for each quality of radiation (HVT) of interest.

When using this technique it is not necessary that all the exposures should be identical. In fact it is preferable that they should be slightly different. This ensures that different instrument readings will be obtained and therefore errors associated with the reading of the instrument scale will be averaged out, and in addition the experimenter is not 'hypnotized' by observing the same reading repeatedly.

A careful record (see Chapter 8) is kept of the various intercomparison ratios and a check made that they are in agreement with the appropriate ratios of the calibration factors (K_E). The technique of keeping a graphed chronological record described earlier (Fig. 9) is again worthwhile, and will indicate, when taken in conjunction with the radioactive check, if and when the calibration factor for any dosemeter is no longer applicable.

The prime purpose of the intercomparisons described above is to confirm that the dosemeters are functioning correctly and that the calibration factors provided by the standardizing laboratories remain valid. However, the results of the intercomparison also indicate quantitatively the extent of any departure of dosemeter sensitivity from the assumed value. If the difference between the intercomparison ratio and the ratio of the calibration factors is less than 1%, no action need be taken.

If the difference is greater than 1% but not greater than 3%, the result of the intercomparison can be used to derive a new calibration factor, $(K_E)'_B$ given by:

$$(K_E)'_B = (K_E)_A \cdot (I_A/I_B)$$

or

$$(K_E)'_B = (K_E)_A \cdot n \cdot (I_A / I_B)$$

or

$$= (K_E)_A \cdot \left[\left(\frac{\bar{i}_A}{i_B} \right)_a \cdot \left(\frac{\bar{i}_A}{i_B} \right)_b \right]^{\frac{1}{2}}$$

depending on the intercomparison technique adopted, where $(K_E)'_B$ is the new (temporary) calibration factor for the dosemeter B and $(K_E)_A$ is the calibration factor given by the standardizing laboratory for the dosemeter A, which is shown by its radioactive check to be still in calibration.

It should be noted that the calibration factor $(K_E)'_B$ determined in this way is applicable at the same pressure and temperature for which the factor $(K_E)_A$ applies. The pressure and temperature at the time of their intercomparison is not relevant, provided both chambers are unsealed or both sealed.

If the difference is greater than 3%, a recalibration is desirable. Ideally this should be done by returning the instrument to the standardizing laboratory. If this is not practicable immediately, the result of the intercomparison can be used as described above to derive a new calibration factor which can be used temporarily pending the recalibration of the instrument by the standardizing laboratory.

This technique of intercomparison is similar to that used by the secondary standard laboratories to determine instrument calibration factors. In careful experienced hands it can therefore be used to obtain calibration factors for a previously uncalibrated dosemeter. It must be noted, however, that the consequent accuracy of the calibration factor will be lower than that of the calibration factor which would be given by the secondary standardizing laboratory since the error in the calibration factor $(K_E)_A$ used in the above equation will be added to the experimental errors.

3.5.7. Frequency of checks

All the checks described above should be done whenever there is a suspicion that anything could be amiss and whenever any possible damage is suspected. They should also be done immediately before and after and, if possible, on the same occasion of any calibration (i.e. determination of K_E values). The checks should also be done at not less than 3-monthly intervals and a careful record kept of the results. This record serves to identify the date at which any changes occur and to prove the constancy of the dosemeter up to that date.

3.6. NUMBER OF EXPERIMENTAL READINGS TO BE TAKEN

The number of readings which need to be taken and the number of interchanges which need to be made depend on how reproducible are the exposures and the positioning of the chamber. The total number of readings should be such that the standard error of the mean values of the instrument readings, or their ratios, is less than 1%.

TABLE VI. CALCULATION OF THE STANDARD ERROR OF A SERIES OF INSTRUMENT READINGS

	Instrument reading	Deviation from the mean Δ	Δ^2
	51.4	+ 1.0	1.0
	49.6	- 0.8	0.64
	49.9	- 0.5	0.25
	51.7	+ 1.3	1.69
	49.4	- 1.0	1.00
Total	252.0		Total 4.58

For example: A series of 'identical' exposures are made and the associated readings obtained on the dosemeter are listed in the first column of Table VI. The readings are not identical, presumably due to variation in the actual exposure and to instrumental variations or both. It is important to notice that there is no drift of the values, i. e. they do not become progressively larger or smaller but are fairly randomly distributed (within a range less than ± 2-3%). If there is such a drift or excessive variation, it must be investigated further since either the X-ray machine or the instrument is almost certainly misbehaving.

The standard error (S. E.) is given by the formula:

$$\text{S. E.} = \left[\frac{\sum_{1}^{n} \Delta_i^2}{n(n-1)} \right]^{\frac{1}{2}}$$

where Δ_i is the difference between a particular instrument reading and the mean instrument reading and n is the number of readings taken.

In the example the number of readings is five and their mean value is calculated to be 50.4 (i. e. 252.0/5). The deviations and their squares are given in the second and third colums of Table VI. The sum of the squares of the deviation, $\sum_{1}^{n} \Delta_i^2$, is 4.58 and the standard error is, therefore:

$$\text{S. E.} = \left[\frac{4.58}{5 \times 4} \right]^{\frac{1}{2}} = 0.5$$

or, expressed as a percentage of the mean reading (50.4),

$$\% \text{ S. E.} = \frac{0.5}{50.4} \times 100 = 1\%$$

The difference between the maximum and minimum instrument readings is known as the range and in this example is:

$$\text{Range} = 51.7 - 49.4 = 2.3$$

or, expressed as a percentage of the mean reading (50.4),

$$\text{Percentage range} = \frac{2.3}{50.4} \times 100 = 5\%$$

A range of 5% and a standard error of 1% are slightly above the acceptable limits. It is desirable to have readings rather less variable than those of this example – fortunately they usually are! It may be added that the necessity of making enough readings to achieve a standard error of substantially less than 1% applies to all occasions on which radiation measurements are made.

CHAPTER 4

MEASUREMENT OF RADIATION OUTPUT

The main purpose of this manual is to give practical recommendations for the measurement of the radiation output of X and γ-ray equipment in radiotherapy. However, it is necessary first to ensure that the equipment is operating correctly, is properly adjusted and that the installation is safe. Details of the recommended method of radiation output measurement are given in this chapter but it must be emphasized that the work described in Chapters 5, 6 and 7 must precede such measurements.

4.1. DEFINITION OF OUTPUT TERMS

The quantity which needs to be measured is the exposure (röntgen) at the surface of the irradiated patient for kilovoltage X-rays or at the depth of the peak of the depth dose curve for caesium-137 and cobalt-60 γ-rays. This quantity is referred to as the 'peak or surface exposure' and designated by the symbol X_s. As described in Chapter 2, the delivery of the treatment to the patient is controlled by the use of either a timer or a monitor system. The output of therapy equipment is therefore stated in terms of 'peak exposure per min' or 'peak exposure per monitor division'. These quantities are designated by the symbol $\dot{X}_s$ and X'_s respectively.

The corresponding values of the 'peak (surface) absorbed dose', 'peak absorbed dose per minute' and 'peak absorbed dose per monitor division' are designated by the symbols D_s, $\dot{D}_s$, and D'_s respectively and are calculated by multiplying the corresponding exposure, exposure per minute or exposure per monitor division by the rad per röntgen factor, f_λ. In this manual, and in particular in this chapter, it is the absorbed dose in soft tissue which is of interest. Values of f_λ are given in Table AVII of Appendix V (see also Chapter 2). For convenience these quantities and the relation between them are listed in Tables VII and VIII respectively. These quantities represent the precisely defined output of a piece of equipment. It will be recalled that in this manual the term output is used in a general non-specific sense.

4.2. METHOD OF MEASUREMENT FOR X-RAYS

For the present (but see Chapter 9) it will be assumed that for the purposes of dosimetry the patient is equivalent to a large volume of water and that the surface being irradiated is flat and normal to the beam central axis. It is the peak (or surface) exposure under these conditions which is to be determined. The method of measurement recommended here is based on a measurement made at a depth in a water phantom and utilizes percentage depth dose data taken from the percentage depth dose or isodose charts that are to be used in clinical practice. The selection of appropriate percentage depth dose data is described in Chapter 5. This method of measurement is preferred to either a direct measurement of the peak (or surface) exposure or a measurement of the exposure in air since these, in spite of being common and popular methods, are difficult to do with sufficient accuracy. Primarily,

TABLE VII. DEFINITION OF OUTPUT TERMS

Symbol	Quantity	Unit
X_s	Peak (or surface) exposure	R
$\dot{X}_s$	Peak (or surface) exposure rate	R/min
X'_s	Peak (or surface) exposure factor	R/monitor division
D_s	Peak (or surface) absorbed dose	rad
$\dot{D}_s$	Peak (or surface) absorbed dose rate	rad/min
D'_s	Peak (or surface) absorbed dose factor	rad/monitor division

TABLE VIII. RELATIONSHIP BETWEEN OUTPUT TERMS

$$D_s = X_s \cdot f_\lambda \quad \text{rad}$$
$$\dot{D}_s = \dot{X}_s \cdot f_\lambda \quad \text{rad/min}$$
$$D'_s = X'_s \cdot f_\lambda \quad \text{rad/monitor division}$$
$$X_s = \dot{X}_s \cdot m \quad \text{R}$$
$$X_s = X'_s \cdot m' \quad \text{R}$$
$$D_s = \dot{D}_s \cdot m \quad \text{rad}$$
$$D_s = D'_s \cdot m' \quad \text{rad}$$

where m = exposure time in minutes
m' = exposure 'time' in monitor divisions

however, this method is recommended since it facilitates the use of published percentage depth dose data.

The tank of water (usually known as the water phantom) used for the measurement should be sufficiently large that the full contribution of scattered radiation reaches the measuring point. In practice this requires that there should be a minimum of 5 cm of water surrounding the geometric beam of radiation and 5 cm of water beyond the point of measurement. This is illustrated in Fig. 13.

Since this phantom can also be used for other types of measurement, it is convenient to make it fairly large, e. g. a $30 \times 30 \times 30$ cm cube. Although a water phantom is preferable, it is possible to use a solid phantom and this is sometimes more convenient to the worker. Great care must be taken, however, to use a suitable material. Sheets of 'Mix D' [5] and high impact polystyrene are two suitable materials. The requirements are that the average electron density, atomic number and bulk density should be close to those of water (or wet soft tissue). Perspex (Lucite) is not suitable.

It is sometimes inconvenient, or impossible, to make measurements with the beam directed vertically downwards on to a free water surface (Fig. 13). For this reason it is usual to employ a horizontal beam directed through the solid wall of the tank that contains the water (Fig. 14). The walls of the tank are conveniently made of Perspex and, provided the wall thickness

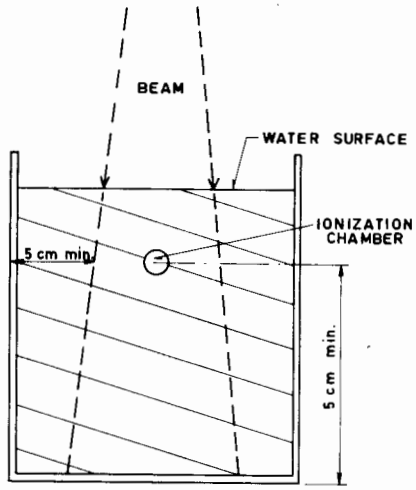

FIG.13. Size of phantom to be used for output measurements.

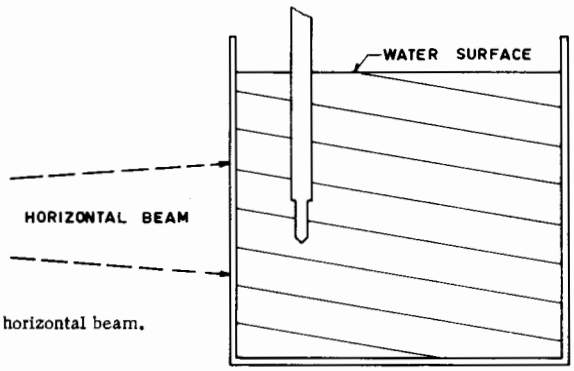

FIG.14. Water phantom used with horizontal beam.

does not exceed 3 mm, its presence has no significant effect on the value of the exposure. If walls thicker than 3 mm are used, it is advisable to have a central area of reduced thickness (a 'window') a few centimetres in diameter (Fig. 15).

A more detailed drawing of a water phantom suitable for this and other measurements described in this manual is given in Appendix VIII.

For measurements in the water phantom the ionization chamber must be fitted with a close-fitting waterproof sheath which is usually made of Perspex. Because the measurement is being made in water, the wall thickness of the Perspex sheath does not affect the observed instrument reading significantly. The wall thickness should not, however, be greater than $\frac{1}{2}$ cm and it is often convenient to make it the same thickness as the equilibrium cap used for the cobalt-60 γ-ray or 2-MV X-ray calibration (section 3.2.4).

4.2.1. Experimental procedure

The procedure to be followed is:

(1) Position the ionization chamber with its centre at a depth of 5 cm on the central axis of the beam inside the water phantom (Fig. 16). Make

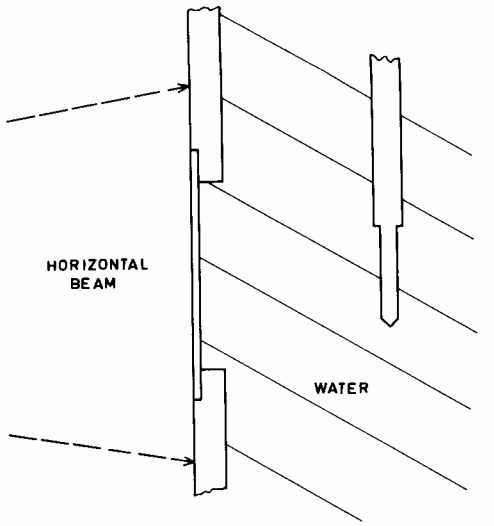

FIG. 15. 'Window' in phantom wall.

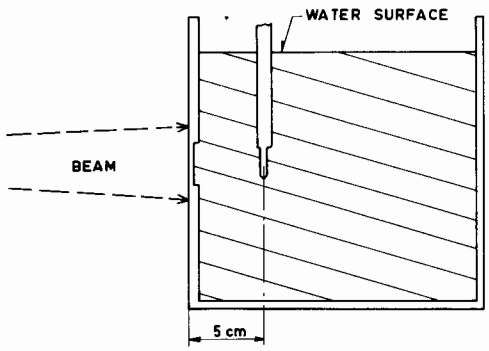

FIG. 16. Ionization chambers at 5 cm depth in water phantom.

several identical exposures (at the selected, specified operating condition), note the readings and calculate the average instrument reading. Let this average value be I divisions. Note the duration (m) of each exposure in minutes.

(2) The exposure at a depth of 5 cm (X_5) is then calculated by

$$X_5 = I \cdot K_E \cdot \phi(p,t) \qquad R$$

where K_E is the instrument calibration factor (section 3.3); and $\phi(p,t)$ is the temperature and pressure correction (section 3.3.1).

(3) The exposure (X_s) at the maximum (i.e. 100% position) of the depth dose curve is now calculated.

$$X_s = \frac{X_5 \times 100}{D*5} \qquad R$$

or

$$X_s = \frac{I \cdot K_E \cdot \phi(p,t) \times 100}{D*5} \qquad R$$

where D*5 is the percentage depth dose at 5 cm deep for the particular operating conditions employed, taken from the %DD tables to be used in clinical practice.

(4) The exposure rate ($\dot{X}_s$) at the maximum position is hence

$$\dot{X}_s = \frac{X_s}{m} = \frac{X_5 \times 100}{m \cdot D*5} \quad R/min$$

or

$$\dot{X}_s = \frac{I \cdot K_E \cdot \phi(p,t) \times 100}{m \cdot D*5} \quad R/min$$

If a monitor system is in use and m' represents the number of monitor divisions corresponding to each exposure, the exposure rate factor (X'_s) at the maximum position is given by

$$X'_s = \frac{X_s}{m'} = \frac{X_5 \times 100}{m' D*5} \quad R/monitor\ division$$

or

$$X'_s = \frac{I \cdot K_E \cdot \phi(p,t) \times 100}{m' D*5} \quad R/\ monitor\ division$$

(5) The corresponding values of the dose, dose rate and dose rate factors are given by

$$D_s = X_s \cdot f_\lambda = \frac{I \cdot K_E \cdot \phi(p,t) \times 100 \times f_\lambda}{D*5} \quad rad$$

$$\dot{D}_s = \dot{X}_s \cdot f_\lambda = \frac{I \cdot K_E \cdot \phi(p,t) \times 100 \times f_\lambda}{m \cdot D*5} \quad rad/min$$

$$D'_s = X'_s \cdot f_\lambda = \frac{I \cdot K_E \cdot \phi(p,t) \times 100 \times f_\lambda}{m' \cdot D*5} \quad rad/monitor\ division$$

where f_λ is the rad per röntgen conversion factor (Table AVII) appropriate to the radiation quality at the point of measurement in the water and to the material in which the dose is being stated.

A numerical example of output calibration using this method and associated formulae are given below in sections 4.2.3 and 4.2.5.

In principle, the presence of the ionization chamber plus waterproof sheath changes the dose at the point in the water phantom where the measurement is being made since its presence displaces the water normally there. Fortunately it is found that, for 200-400 kV X-rays, the increase in dose due to increased transmission is sufficiently balanced by the decrease in dose due to decrease in scattered radiation. The formulae given above are complete and no further corrections are needed, provided that any correction needed because of timing errors (see sections 4.2.2 and 4.2.4) has been made. For ^{137}Cs and ^{60}Co γ-ray measurements a further correction is required to take account of the displacement of water by the measuring system (see section 4.3).

4.2.2. Timing errors

4.2.2.1. During treatment

At the beginning of the exposure, the exposure rate rises from zero to its steady value over a finite period of time either because the shutter is opening and/or the kV is being raised progressively to its full value. Similarly at the end of the exposure, the exposure rate does not necessarily fall to zero instantaneously. Figure 17 shows an example of the kind of

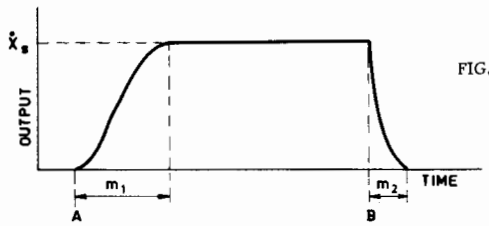

FIG. 17. Variation in exposure rate during an exposure.

pattern in exposure rate that may occur when an exposure is made. The exposure commences and the timer starts at A; the timer stops at B and operates the switching OFF mechanism. For the greater part of the time the output is constant and equal to $\dot{X}_s$. It is this value of $\dot{X}_s$ which we wish to determine since the peak exposure is related to the exposure time by the expression

$$\text{Peak exposure} = \text{Peak exposure rate} \times \text{exposure time}$$

$$X_s = \dot{X}_s \cdot m$$

However, it can be seen that at the beginning and the end of the exposure the output is less than $\dot{X}_s$ and that the real exposure time extends beyond the nominal end of the exposure. The actual exposure (röntgen) may be therefore greater or smaller than that intended. Fortunately, in practice, the difference is insignificant since the times m_1 and m_2 are usually a few seconds or less, whereas the exposure time is usually several minutes. The treatment exposure time may therefore be correctly calculated by the equation given immediately above, unless the timer end errors are found to be large relative to the treatment exposure times (see sections 4.2.3.1 and 4.2.3.2).

4.2.2.2. During measurements

When making output measurements, however, the errors associated with the starting and stopping of the exposure cannot usually be neglected. Typically the exposure times are only $\frac{1}{2}$ - 2 minutes so that their neglect may result in the introduction of errors of several per cent. The full line of Fig. 18 shows the output pattern for an exposure of duration m minutes (a) and that for two consecutive exposures each of 0.5 m minutes (b). The associated instrument readings I_1 and I_2 are proportional to the areas under these curves. It can be shown therefore that the instrument reading (I) that would have been obtained if the output had been constant during the whole time (m) of the exposure is given by

$$I = 2I_1 - I_2$$

It should be noted that I_1 may be greater, smaller or equal to I_2 depending upon the exact details of switching ON and OFF mechanism.

Hence in the expressions given above in section 4.2.1 the value of I used should be that determined by this 'double exposure' technique.

4.2.3. An example of output measurement using a timer

A particular X-ray equipment is operated at 2.0 mm Cu HVT and 50 cm SSD. The beam size is 10 × 10 cm. The depth dose data being used show that the percentage depth dose at 5 cm deep (D*5) for such a beam is 68.8%. The measurements were made at 750 mm Hg pressure and 25°C with the centre of the ionization chamber 5 cm deep on the central axis of the beam in a water phantom, and a series of identical exposures (constant kV and mA) given. The following results were obtained:

1 min exposure I_1	$2 \times \frac{1}{2}$ min exposure I_2
50.9	48.5
50.1	49.0
50.4	49.8
51.2	48.6
49.9	49.1
Mean value = 50.5	Mean value = 49.0

Hence $I = 2I_1 - I_2$

$= 2 \times 50.5 - 49.0$

$= 52.0$ divisions

Hence, if the instrument calibration factor is 1.02 at this quality (760 mm Hg and 20°C), the peak exposure rate is

$$\dot{X}_s = \frac{I \cdot K_E \cdot \phi(p,t) \times 100}{m \cdot D*5} \quad \text{R/min}$$

$$= \frac{52.0 \times 1.02}{1 \times 68.8} \times \left(\frac{760}{750} \times \frac{298}{293}\right) \times 100 \quad \text{R/min}$$

$$= 79.4 \quad \text{R/min}$$

From Table AV.III it can be seen that the rad/röntgen factor at this quality for soft tissue is 0.95. The peak dose rate is therefore given by

$$\dot{D}_s = \dot{X}_s \cdot f_\lambda = 79.4 \times 0.95$$

$= 75.4$ rad/min, in soft tissue.

4.2.3.1. Magnitude of timer error

The readings obtained using the single/double exposure technique described above can also be used to evaluate the magnitude of the combined

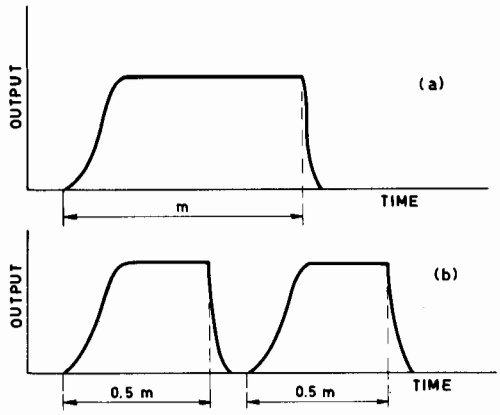

FIG. 18. Double exposure technique to determine exposure and errors:
(a) single exposure;
(b) double exposure.

starting and stopping error in terms of seconds of exposure time at the full output. The single exposure (1 min) has one starting and one stopping error associated with it, whereas the double exposure has two stopping and two starting errors associated with it. The difference between the two instrument readings obtained using a single and a double exposure corresponds, therefore, to the reading associated with one starting plus one stopping error. In the above example it is

$$50.5 - 49.0 = 1.5 \text{ divisions}$$

But the reading associated with an (error free) exposure time of 1 minute has been shown to be equal to

$$2 \times 50.5 - 49.0 = 52.0 \text{ divisions}$$

The time ϵ associated with the combined starting and stopping error is therefore, by simple proportion, given by

$$\epsilon = 1 \times \frac{1.5}{52.0} \text{ min}$$

$$= 1 \times \frac{1.5}{52.0} \times 60 \text{ s}$$

$$= \sim 1.7 \text{ s}$$

It can be seen that the effective time is less than the time set on the timer since the actual 1 min reading (50.5) is less than the true 1 min reading (52.0). The timer error is therefore,

$$\epsilon = -1.7 \text{ s}$$

4.2.3.2. Correction to treatment exposure times

If the timer error ϵ is found to be significant compared with the treatment time, it is necessary to make an allowance for it. If the error is negative then each treatment time m calculated on the basis of the equation

of section 4.2.2.1 should be increased by ϵ. On the other hand, if the error is positive, the treatment time needs to be decreased by ϵ. For example, a particular X-ray machine is found to have a timer error equal to -15 s (and this is not impossible) and it is required to give an exposure of 250 R. The measured exposured rate ($\dot{X}_s$) is found to be 62.5 R/min. If there were no timer error the exposure time would be

$$\text{exposure time} = \frac{250}{62.5}$$

$$= 4 \text{ min}$$

Making allowance for the known error (ϵ = -15 s), the corrected exposure time is

4 min 15 s

In this example neglect of the timer error would lead to an underdosage of 6%.

4.2.4. Monitor ON-OFF errors

When a monitor system is used to control the end of an exposure, an effect similar to that described above (under 'timing errors') may occur, although it is usually smaller in magnitude. As for the timer, the error can usually be neglected in the treatment exposure. When making a measurement the same technique as was used for the timer is used to eliminate the effect as the following example shows.

4.2.5. Example of output measurement using a monitor

A particular X-ray apparatus is operated at 2.0 mm Cu HVT and 50 cm SSD. The beam size is 10 × 10 cm. The depth dose being used shows that the percentage depth dose at 5 cm deep for such a beam is 68.8%. The measurements were made at 750 mm Hg pressure and 25° C, with the centre of the ionization chamber at 5 cm deep on the central axis of the beam in a water phantom and a series of exposures given. Half the exposures were given using for each a monitor reading of 100 divisions. The other exposures were each double ones consisting of 50 divisions. The results obtained were

100 monitor divisions	2 × 50 monitor divisions
I_1	I_2
73.6	73.8
74.1	74.4
73.5	74.7
73.6	73.9
73.2	74.2
Mean = 73.6	Mean = 74.2

Hence the reading which would have been received for an exposure of 100 monitor divisions without any end error was

$$I = (2 \times 73.6) - 74.2 = 73.0$$

The instrument calibration factor, K_E at 2.0 mm Cu HVT is 1.03 at 760 mm Hg pressure and 22° C. Hence the peak exposure factor is given by

$$X_s' = \frac{I \cdot K_E \cdot \phi(p, t) \times 100}{m' \cdot D^{*5}}$$

$$= \frac{73.0}{100} \times \frac{1.03}{68.8} \times \frac{760}{750} \times \frac{298}{295} \times 100$$

$$= 2.10 \text{ R/monitor divisions}$$

or, since at this quality $f_\lambda = 0.95$ rad/R in soft tissue

$$\text{peak dose factor} = D_s' = X_s' \cdot f_\lambda$$

$$= 2.10 \times 0.95$$

$$= 1.99 \text{ rad/monitor division, in soft tissue}$$

4.2.6. Measurement of output in practice

The numerical value of the output depends upon the exact operating conditions of the X-ray equipment (i.e. kV, mA, filter), the source-skin distance, the beam size, and the detailed design and construction of the X-ray equipment. In the case of X-ray equipment a treatment cone (applicator) usually controls both the beam size and SSD.

It is strongly recommended that each X-ray machine should be operated only at a small number of different kilovoltage and filter combinations. In particular the use of a low kV plus aluminium filter and a high kV plus copper filter on the same equipment is to be avoided if possible. Preferably, each machine should be used at only one setting of kV and with a permanently fixed filter since this minimizes any possible errors in treatment due to incorrect choice of operating conditions.

Likewise one (or at the most two) standard SSD should be chosen and the one specified by the manufacturers (being the one for which the machine was designed) is the obvious choice. Different beam sizes will be obtained by using either a continuously variable diaphragm system (usual on cobalt equipment) or by means of a set of treatment cones (usual on kilovoltage X-ray equipment).

Even with these limitations there are likely to be 20 or more different output values to be determined and confirmed regularly. The following recommended method of working minimizes the amount of work involved and yet achieves a high accuracy. The essence of the method is to choose for each radiation quality (HVT) in use one particular, easily reproducible, combination of beam size and SSD as a standard and to make measurements (on the central axis at 5 cm deep inside a water phantom) for all the desired combinations of beam size and SSD (i.e. with every treatment cone) for which outputs are required by reference to this standard. Such relative

measurements need not of course, be made with a calibrated dosemeter, i.e. one for which the calibration factors K_E are known. It is good practice to reserve the calibrated dosemeter for those measurements for which it is essential. This minimizes its use and therefore possibility of damage. A further attractive feature of this method of working is that an instrument which is convenient to use can be chosen.

Because of scattering, the radiation quality at a point inside a water phantom depends upon the position of the point of measurement with respect to the surface and to the central axis and upon the size of the beam as well as on the quality (HVT) of the primary radiation incident upon the water. In principle, therefore, the appropriate calibration factor for the chamber will vary with beam size and depth of measurement. Such a situation is clearly most inconvenient and it is therefore important to use a chamber that has a small variation in sensitivity over the range of qualities existing in the water. In practice this means that over the quality range 60-150 keV effective (0.5-3.0 mm Cu HVT) the value of K_E should remain within ±1% as the effective energy changes by ±30 keV. With such a chamber the same

TABLE IX. VARIATION OF RADIATION QUALITY (HVT) INSIDE A WATER PHANTOM WITH DEPTH, FIELD SIZE AND HVT OF INCIDENT RADIATION [6]
The figures given in the table are HVT in mmCu.

Depth (cm)	Area (cm^2)					
	0	25	50	100	200	400
	Primary HVT 1.25 mm Cu					
0.0	1.28	0.98	0.92	0.87	0.84	0.82
2.5	1.38	0.85	0.72	0.66	0.61	0.58
5.0	1.55	0.82	0.72	0.63	0.56	0.52
10.0	1.83	0.90	0.75	0.62	0.49	0.46
15.0	2.24	0.99	0.75	0.60	0.50	0.47
	Primary HVT 2.20 mm Cu					
0.0	2.26	1.89	1.80	1.70	1.62	1.54
2.5	2.39	1.72	1.55	1.41	1.23	1.15
5.0	2.53	1.70	1.48	1.21	1.08	1.00
10.0	2.67	1.54	1.29	1.06	0.88	0.80
15.0	2.90	1.48	1.20	0.98	0.81	0.66
	Primary HVT 3.20 mm Cu					
0.0	3.16	2.78	2.67	2.56	2.45	2.37
2.5	3.19	2.45	2.26	2.09	1.93	1.81
5.0	3.26	2.46	2.22	1.91	1.70	1.56
10.0	3.38	2.31	1.98	1.69	1.44	1.27
15.0	3.46	2.11	1.75	1.45	1.18	0.88

TABLE X. NUMERICAL DATA FOR OUTPUT DETERMINATION

Beam size square	Instrument Readings	Mean	Associated 10 cm square value	Ratio	%DD at 5 cm for SSD 50 cm	Relative output	Output
(a)	(b)	(c)	(d)	(e)	(f)	(g)	(h)
10 x 10 cm	82.5						
	83.3						
	82.9	82.9		1.000	68.7	1.453	75.4
5 x 5	61.6						
	62.0						
	61.5	61.7	82.5	0.749	56.2	1.332	69.1
7 x 7	72.1						
	71.0						
	71.4	71.5	82.5	0.868	62.3	1.391	72.2
10 x 10	82.3						
	81.6						
	82.4	82.1					
12 x 12	88.6						
	89.1						
	88.1	88.6	82.3	1.076	71.8	1.499	77.7
15 x 15	95.6						
	94.4						
	95.3	95.1	82.3	1.153	75.1	1.538	79.7
10 x 10	82.9						
	82.1						
	82.5	82.5					

value of K_E can be used for all beam sizes and depths of measurement without introducing significant error. If such a chamber is not available, it is necessary to estimate the effective energy (quality) at the point of measurement and to use the calibration factor appropriate to this quality. Information about the variation of HVT with beam size and depth for a range of primary radiation qualities is given in Table IX which is taken from Ref. [6].

To summarize: it is desirable that the measuring instrument used should have the following characteristics:

(1) A small variation in sensitivity with HVT
(2) A convenient sensitivity so that a reading of about 2/3 full scale is obtained in $\frac{1}{2}$-2 min exposure time
(3) Direct reading
(4) Cable connected type. This minimizes the number of times the X-ray room has to be entered and thus encourages taking multiple readings.

Of course the calibrated instrument may have these properties but this is not necessarily so.

After making the relative measurements, the calibrated dosemeter is used to determine the output (as described previously in sections 4.2.1, 4.2.3 and 4.2.5) for the chosen standard beam size and SSD. This output is then used to convert all the relative values into the desired output values. An example of this method, in which an uncalibrated measuring instrument having the above characteristics is used for the relative measurements, is given below. In this example it is assumed that it is desired to know the output values for a range of beam sizes expressed in rad per min in soft tissue.

4.2.7. Example of output measurement

Measuring instrument No. 2 at 5 cm deep in water phantom; Super Ragox X-ray machine in Room 2; 250 kV, 20 mA, Filter "No. 3" (1.5 mm Cu + 1 mm Al), HVT - 2.0 mm Cu - SSD = 50 cm.

The numerical data resulting from the measurements are shown in Table X, the contents of the various columns being: Column (a), the beam size. In this example, this is the only variable operating condition; in practice, of course, the SSD could also be a variable; Column (b), the readings obtained on the instrument for identical settings of the exposure timer (the exact value of the exposure time does not need to be known); Column (c), the mean of the values in column (b) for each beam size; Column (d), the interpolated value which the standard (in this case the 10 cm square) beam size would have. This is obtained by interpolation between the standard measurements done immediately before and after each group of readings. The importance of repeating the 'standard' measurements at intervals cannot be overemphasized. The choice of the time intervals between the standard measurements depends upon the reproducibility of the readings. If the X-ray equipment and dosemeter are stable, a standard measurement at the beginning and another at the end of a session lasting 1 hour or more may be sufficient. On the other hand, a standard measurement before and after each measurement may sometimes be required, although this is unusual; Column (e), the ratio of the values in columns (c) and (d):

$$\frac{\text{Mean value (c)}}{\text{Associated 10 cm square value (d)}}$$

which is the dose rate at 5 cm deep for each condition relative to that for a 10 cm square beam (in this example); Column (f), the %DD at 5 cm deep, for SSD 50 cm, for each beam size is obtained from the tables which are to be used in clinical practice; Column (g), the relative output calculated by

$$\text{Relative output} = 100 \times \frac{\text{Relative dose at 5 cm deep (e)}}{\%\text{DD (f)}}$$

Column (h), the required values of output calculated by

$$\text{Output} = \text{Relative output (g)} \times \frac{\text{Output for standard beam}}{\text{Relative output for standard beam}}$$

The output for the standard condition is determined exactly as described previously. The experimental values and calculations appropriate to the present example are repeated below.

Measurement of peak dose rate for 10 cm square beam

Instrument readings obtained:

1 min exposure	$2 \times \frac{1}{2}$ min exposure
50.9	48.5
50.1	49.0
50.4	49.8
51.2	48.6
49.9	49.1
Mean value = 50.5	49.0 = Mean value

Hence reading to be expected in 1 'true' minute

$$I = 2 \times 50.5 - 49.0 = 52.0 \text{ divisions}$$

The other relevant data are:

Instrument No. 368922, K_E = 1.02 at 760 mm Hg, 20° C.
%DD at 5 cm deep for 10 cm square beam = 68.8%
Ambient pressure = 750 mm Hg
Ambient temperature = 25° C
rad/R factor = 0.95

Hence the peak dose rate for the 10 × 10 cm beam is:

$$\dot{D}_s = \frac{52.0}{1} \times \frac{1.02}{68.8} \times \left(\frac{760}{750} \times \frac{298}{293}\right) \times 100 \times 0.95$$

$$= 75.4 \text{ rad/min in soft tissue}$$

The corresponding output (h) for the 7 × 7 cm beam, for example, is calculated from the data of Table X as follows:

Mean reading for 7 × 7 cm beam (c): 71.5

Associated 10 × 10 cm beam reading (d): 82.5

Ratio (e): 71.5/82.5 = 0.868

%DD at 5 cm deep for 7 × 7 cm beam (f): 62.3

Relative output for 7 × 7 cm beam (g): 86.8/62.3 = 1.391

%DD at 5 cm deep for 10 × 10 cm beam (f): 68.7

Relative output for 10 × 10 cm beam (g): 100/68.7 = 1.453

Output for 10 × 10 cm beam (h): 75.4 rad/min

Output for 7 × 7 cm beam (h): $\frac{1.391}{1.453} \times 75.4 = 72.2$ rad/min

4.3. RADIOACTIVE CAESIUM AND COBALT TELETHERAPY EQUIPMENT

The procedure for the output measurements of caesium or cobalt equipment is identical with that described above for X-ray equipment with one important difference. As specified earlier (section 3.2.4) for calibration with ^{60}Co γ-rays the ionization chamber needs either to have a thick wall or to be fitted with a Perspex build-up cap. Typically, the outside diameter of such a chamber or chamber plus build-up cap is 17-20 mm. The insertion of such a large measuring chamber system into the water will change the value of the quantity that is to be measured. It is therefore necessary to reduce the observed instrument reading by use of a 'displacement factor' δ. The output is hence given by:

$$\dot{X}_s = \frac{I}{m} \cdot \frac{100}{D*5} \cdot K_E \cdot \delta \cdot \phi\,(p, t) \quad \text{R/min}$$

or

$$\dot{D}_s = \frac{I}{m} \cdot \frac{100}{D*5} \cdot K_E \cdot \delta \cdot \phi\,(p, t) \cdot f_\lambda \text{ rad/min in soft tissue}$$

where the symbols have the same meaning as previously (section 4.2.1) and the value of the displacement factor δ for a chamber with an overall external diameter of 15-25 mm is 0.98 for both ^{137}Cs and ^{60}Co γ-rays. The value of the instrument calibration factor K_E to be used in this equation for both ^{137}Cs and ^{60}Co γ-rays is that quoted for ^{60}Co γ-rays or 2-MV X-rays (see Table IV).

For these measurements at a depth inside a water phantom the chamber needs to be fitted with a waterproof sheath and it is convenient if this is made of Perspex. Strictly, the thickness of this sheath should be identical with that of the build-up cap if it replaces the build-up cap, or should be very thin (say less than 2 mm) if the build-up cap is left on or if the chamber is itself thick-walled. In practice the thickness of the Perspex waterproof sheath is unimportant and it is convenient to have it the same as for the kilovoltage X-ray measurements (i.e. 3-5 mm). The displacement factor is 0.98 no matter what the thickness of the waterproof sheath, its value being determined by the thickness of the ionization chamber wall plus build-up cap used for the ^{60}Co γ-ray or 2-MV X-ray calibration.

4.3.1. Continuously variable collimators

Some X-ray equipment and most telecobalt equipment is fitted with continuously variable collimators so that a very large number of different square and rectangular beam sizes are possible. It is recommended that the outputs for a range of square beams should be measured by the technique described above and a graph of output versus beam size plotted. The outputs for intermediate beam sizes are then obtained by interpolation. The outputs for rectangular beams can be obtained by using the method of equivalent squares. Reference [3] contains a table which lists the value of the 'equivalent square' corresponding to a wide range of rectangular beam sizes. The output (surface backscatter factor, tissue-air ratios and central axis percentage depth doses) for a rectangular beam can be taken as being identical to that of its equivalent square. Alternatively the outputs for a range of rectangular beams may be obtained by measurement, using the technique already described.

Some equipment is also fitted with penumbra trimmers supplementary to the main collimators. It should be noted that for any beam size the output for a beam collimated by the main collimators only will be different from the output for a beam collimated by the main collimators plus trimmers.

4.4. MEASUREMENT OF OUTPUT FOR SPECIAL TECHNIQUES

4.4.1. Measurement of output when a wedge filter[4] is used

When using a wedge filter it is usual and convenient to retain the point of maximum dose on the central axis as the 100% point of the isodose chart (Fig. 19). Likewise the given dose required to achieve the desired tumour dose is stated at this point.

Clearly the insertion of a wedge filter into the beam will reduce the dose rate at this point and it is necessary to measure this reduced output. The method recommended is exactly the same as that given earlier for plain (non-wedged) beams with one additional step.

When making the measurement at 5 cm deep, readings of the dosemeter are taken first with the wedge one way round and the long axis of the ionization chamber lying at right angles to the direction of the wedging (Fig. 20). The wedge is then rotated through 180°, great care being taken not to move the phantom or dosementer, and the readings repeated. The mean of the two groups of readings so obtained is the value used in the calculation of the output.

The reason for this manoeuvre is that, as suggested by Fig. 20, it is difficult to position the ionization chamber exactly on the beam central axis. Because of the variation in dose across the beam, such accurate positioning is very important. However, as can be seen from Fig. 20 (a) and (b), the mean of the measurements made for the two orientations of the wedge (180° apart) will yield the desired value.

When using wedge filters with X-ray beams of conventional quality a separate wedge is necessary for each beam size and it is wise to have this permanently fastened to the appropriate treatment cone. It is therefore

[4] See Glossary, Appendix II.

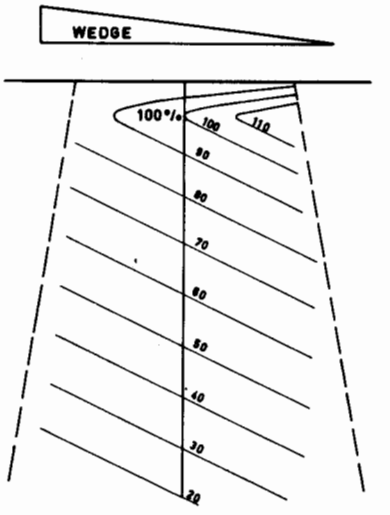

FIG. 19. Wedge isodose chart.

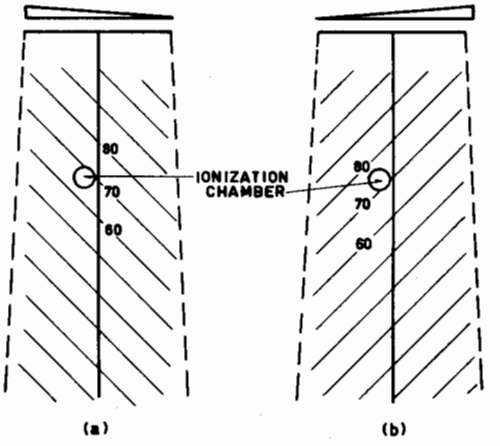

FIG. 20. Wedgr rotated through 180° to facilitate output measurement. Chamber not exactly on central axis reads too low at (a) and too high at (b).

convenient to regard the cone and its wedge as a special cone and to measure and quote its individual output value. For cobalt units, however, it is found that a single wedge filter may be used over a small range of beam sizes so that the wedge filter is not then associated with any particular beam size. Fortunately the % reduction in output resulting from the presence of the wedge filter is substantially independent of beam size. It is therefore convenient to define a 'wedge factor' that is specific to a particular wedge filter but has the same magnitude for all beam sizes to which the wedge filter applies.

The wedge factor is defined as:

$$\text{Wedge factor} = \frac{\text{Output with wedge filter}}{\text{Output without wedge filter}}$$

It is evaluated by measuring the output for the wedged beam in the way described above and by measuring the output for the beam without the wedge

filter in the way described earlier; for the same beam size and on the same occasion. The advantage of this system is that the wedge factor need be measured for only one beam size and can then be used for all beam sizes for which the wedge filter is applicable.

The peak exposure rate $\dot{X}_s$ for the wedge beam is calculated by

$$\text{Peak exposure rate for wedged beam} = \begin{bmatrix}\text{Peak exposure rate}\\ \text{for same beam size}\\ \text{without the wedge filter}\end{bmatrix} \times \text{Wedge factor}$$

For example, on a 5000 Ci telecobalt machine the peak dose rate for an 8 × 8 cm beam at 75 cm SSD is known to be 120 rad/min in soft tissue. A 45° wedge has a measured wedge factor of 0.68. The peak dose rate for an 8 × 8 cm beam at 75 cm SSD wedged to 45° is hence

$$120 \times 0.68 = 81.5 \text{ rad/min in soft tissue}$$

An alternative method used by some workers is to determine directly the output for each beam size for which the wedge filter is applicable.

4.4.2. Measurement of output for moving-field therapy

The output for moving-field therapy is stated as the exposure rate or dose rate at the axis of rotation or point of convergence in air. This value is used in conjunction with the 'tissue-air ratio' (TAR) to determine the dose rate at the axis of rotation within the patient.

The tissue-air ratio is defined[5] as the ratio of the absorbed dose at a depth inside the patient (or water phantom) to the absorbed dose in tissue at the same point in air, i.e. in the absence of the patient. The precise definition requires there to be just enough material surrounding the point 'in air' to provide build-up (electronic equilibrium). The meaning of TAR is explained by reference to Fig. 21, where ${}_M\dot{D}_P$ is the absorbed dose rate in the presence of the material and ${}_A\dot{D}_P$ the absorbed dose rate 'in air' at the same point. In the present context the point of interest is that at the axis of rotation, then TAR = ${}_M\dot{D}_P / {}_A\dot{D}_P$.

The value of TAR depends upon (1) the beam size (usually stated at the axis of rotation); (2) the radiation quality; and (3) the thickness d of overlying material (water or tissue). For practical purposes TAR is independent of the source-axis distance. Tables of TAR are available (e.g. Ref. [3]) and it is recommended that published values be used.

The absorbed dose rate at the axis is given by

$$\begin{matrix}\text{Absorbed dose rate}\\ \text{at axis in patient}\end{matrix} = \left(\begin{matrix}\text{Absorbed dose rate}\\ \text{at axis in air}\end{matrix}\right) \times \text{TAR}$$

The direct measurement of the exposure rate in air at the axis is not easy because of the difficulty of avoiding undesirable contributions from scattered radiation. It is preferable to determine the value of the absorbed

[5] The exact definition is given in the Glossary (Appendix II). Although TAR is defined as a ratio of absorbed doses (or dose rates), it can also be regarded (for 200-400 kV X-rays) as being a ratio of exposures (X) since at any given radiation quality $D = X f_\lambda$.

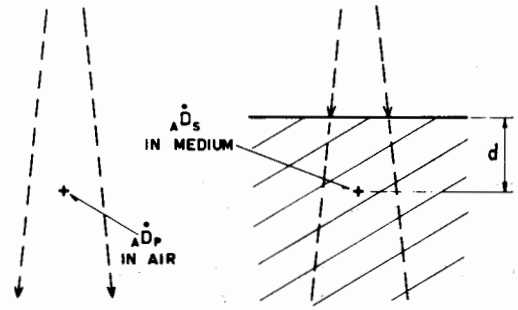

FIG. 21. The meaning of tissue-air ratio.

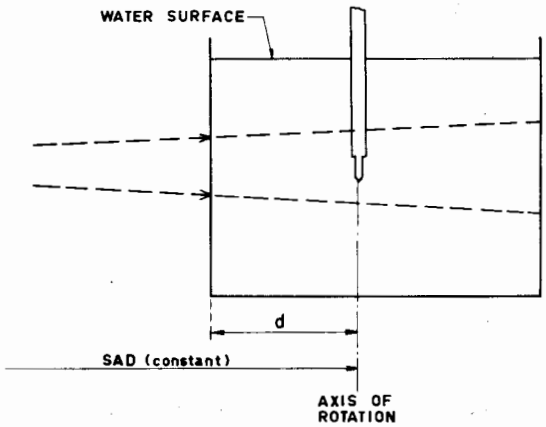

FIG. 22. Measurement of axis dose rate.

dose rate in air at the axis indirectly by measuring the dose rate at the axis at a known depth in a water phantom (Fig. 22). The absorbed dose rate in air at the axis is then calculated by

$$\text{Absorbed dose rate at the axis in air} = \frac{\text{Absorbed dose rate at the axis in water}}{\text{TAR}}$$

This indirect method of measuring is preferred to a direct measurement of the absorbed dose rate at the axis in air since the method recommended allows published TAR values to be used with confidence.

As for the percentage depth dose values used in the determination of the output for fixed-field treatments, the TAR value used for the determination of the absorbed dose rate at the axis in air is the one taken from the set of values which are to be used in clinical practice (e.g. Ref. [3], Brit. J. Radiol. Suppl. 10). The absorbed dose rate at the axis in air should be determined in this way, using a range of beam sizes and depths covering the ranges of clinical interest. For each beam size the values of absorbed dose rate in air so obtained should be identical but in practice will vary a little. The mean value is the one to be used in day-to-day practice. If the values obtained from the measurements made at the various depths are not within ±3% of their mean value, this probably indicates that the TAR values being used are not strictly applicable and a different, more applicable set should be looked for.

The mean absorbed dose rate in air at the axis will be slightly different for each beam size since on most equipment there is a contribution of scattered radiation, the magnitude of which depends on the beam size and which is specific to each individual teletherapy machine. If the air dose rate is measured directly, its variation with beam size will usually be greater than when measured indirectly. Much of this scattered radiation, however, does not contribute to the dose rate at the axis within the patient. It is for this reason that the indirect method of measurement is preferred.

TABLE XI. EXPERIMENTAL RESULTS OBTAINED IN DETERMINATION OF AXIAL ABSORBED DOSE RATE IN AIR

Beam size at axis (cm)	Overlying thickness (cm)	TAR[a]	Mean instrument reading, I	I/TAR
5 × 5	5	0.840	42.1	50.2
7 × 7	5	0.866	45.0	51.9
10 × 10	5	0.893	47.2	52.9
5 × 5	10	0.631	31.6	50.1
7 × 7	10	0.663	34.5	52.0
10 × 10	10	0.700	36.9	52.8
5 × 5	15	0.466	23.4	50.3
7 × 7	15	0.496	25.7	51.8
10 × 10	15	0.534	28.3	53.0

[a] Values taken from Brit. J. Radiol. Suppl. 10 (1961) [3].

Table XI lists some typical results obtained for a ^{60}Co teletherapy machine, whilst Fig. 22 shows the experimental arrangement.

The last column in Table XI lists the values of the quotient, instrument reading/TAR, the values of TAR are taken from the tables [3]. It can be seen that for each beam size the values are substantially the same, whereas there is a variation with beam size. The values for each beam size are plotted and the values for intermediate sizes obtained by interpolation (Fig. 23).

These values can then be converted into the corresponding values of axial absorbed dose rate in air. For example, for a beam size of 6 × 6 cm the interpolated value is 51.4. If the measurements were made at 755 mmHg, 23° C using an exposure time of 1 min and an instrument whose calibration factor K_E is 1.02 at 760 mmHg, 20° C, the axial absorbed dose rate in air is given by

$$\frac{51.4}{1} \times \frac{760}{755} \times \frac{296}{293} \times 1.02 \times 0.98 \times 0.96 = 50.2 \text{ rad/min in soft tissue}$$

where 0.96 is the rad/R factor appropriate for soft tissue for ^{60}Co γ-rays and 0.98 is the displacement factor (section 4.3).

An example of the use of the TAR method to determine the exposure time is given in Appendix IV, example 3.

4.4.2.1. The use of TAR for fixed-field treatment

The use of tissue-air ratios is not confined to moving-field treatments. Some radiotherapists find it convenient to make use of the isocentric arrange-

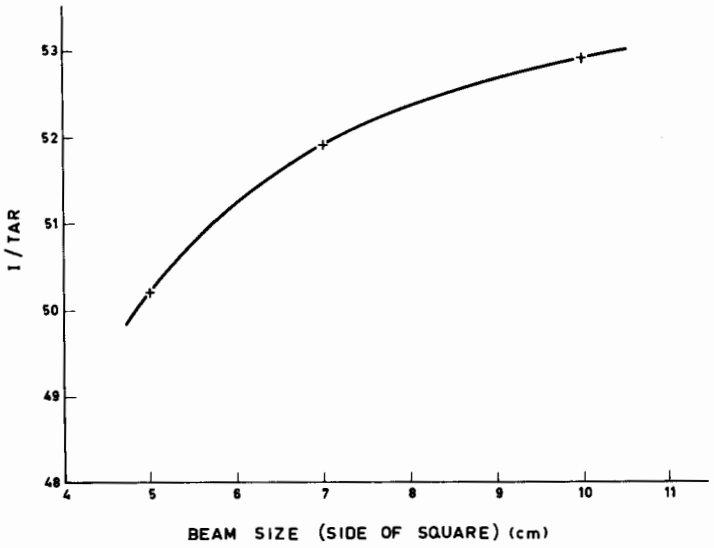

FIG. 23. Graph of axis dose rate in air plotted against beam size.

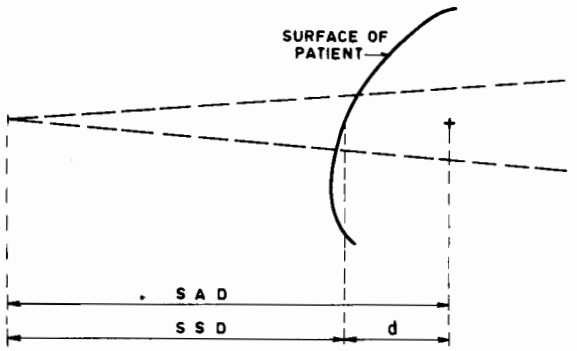

FIG. 24. Use of the TAR method for fixed-beam treatment.

ment provided by moving-beam equipment for fixed-field treatments. This is particularly so for telecobalt equipment. In such treatments the source-axis distance (SAD) is constant whilst the source-surface distance (SSD) is dependent on the thickness of tissue (d) (Fig. 24).

For this usage the isodose charts are drawn with reference to 100% at the axis of rotation. It is recommended that published isodose charts appropriate to the telecobalt equipment should be used.

The magnitude of the absorbed dose rate at the axis (i.e. 100% point of the isodose chart) is determined as for moving-field treatments.

4.4.3. Measurement of output for grid (sieve) treatments

Grid treatment involves placing across the beam a sheet of lead or lead rubber in which there are holes about 1 cm in diameter (Fig. 25). About 60% of the beam area is covered by the lead whilst the remaining 40% is in the form of many (20-200) 1 cm diameter circular holes.

For grid treatments an exposure is usually stated in terms of the exposure in air (i.e. without backscatter) at the position of the patient's skin.

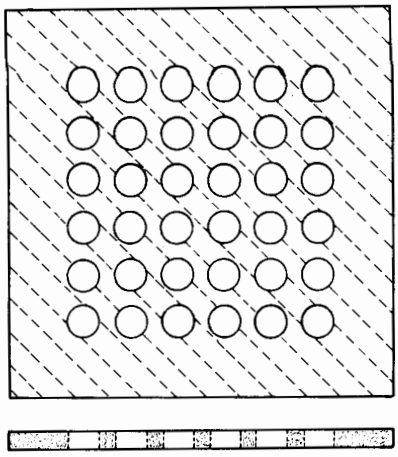

FIG. 25. Grid (sieve) therapy.

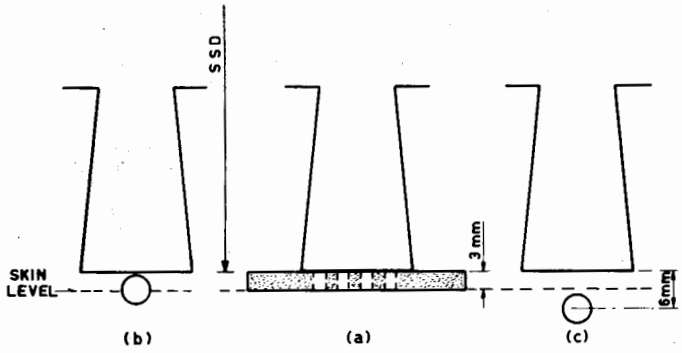

FIG. 26. Measurement of output for grid (sieve) therapy.

The exposure rate is measured directly by positioning the ionization chamber with its centre at the level where the skin will be during treatment (Fig. 26(a)). The grid is not in place whilst the measurement is being done. If the external diameter of the chamber is too large to allow it to be positioned at the desired level, it is necessary to place it as close as possible to this position and apply an inverse-square law correction to the reading. For example, in Fig. 26(c) the exposure rate at a distance of 3 mm from the end of the treatment cone (SSD = 50 cm) is to be determined. The ionization chamber is placed with its centre at 6 mm from the end of the cone and the exposure rate in air is found to be 76.2 R/min. The exposure rate for a grid treatment using this cone is hence

$$76.2 \times \left(\frac{50.6}{50.3}\right)^2 = 77.0 \text{ R/min}$$

4.4.4. Contact and superficial X-ray therapy

Treatments at voltages in the range 50-150 kV and treatment distances in the range 1-20 cm are used to irradiate the skin surface and the immediate underlying layers. The techniques described above for the measurement of

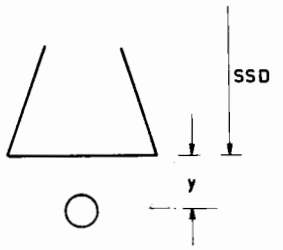

FIG. 27. Measurement of output 'in air' for low kV X-ray treatment.

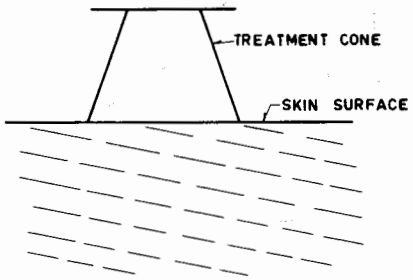

FIG. 28. Output at the skin for low kV X-ray treatment.

output are therefore inappropriate since their objective is to achieve a high accuracy for statement of dosage to points at a depth inside the patient.

The method recommended for the present situation is therefore to position the ionization chamber, in air, as close as possible to the position to be occupied by the skin during the treatment. In practice this means having the ionization chamber in contact with the end of the treatment cone as shown in Fig. 27.

The exposure rate at the centre of the skin surface (Fig. 28) in contact with the end of the treatment cone is given by

$$\dot{X}_s = \frac{I_y}{m} \cdot \phi(p, t) \cdot K_E \cdot \left[\frac{SSD + y}{SSD}\right]^2 \cdot B \quad R/min$$

where B is the surface backscatter factor appropriate to the quality and area of the beam being used and whose value is obtained from published tables [3]; SSD is the source-skin distance; y is the distance from the end of the treatment cone to the centre of the ionization chamber (Fig. 27); I_y is the average instrument reading, corrected for switching on and off errors; m is the exposure time; $\phi(p, t)$ is the pressure and temperature correction factor; and K_E is the ionization chamber calibration factor.

In theory it is necessary to make the measurement for one beam size only and then to calculate the surface exposure rate for other sizes by using the appropriate values of the backscatter factor B. A separate measurement needs to be made for each cone, however, if their lengths (SSD) are not identical and this is advisable in any case because of the different amounts of scattered radiation contributed to the surface from the applicator walls. It is, therefore, worth making the measurements separately for each treatment cone to be used. It is also worth making measurements with the ionization chamber placed at a range of distances (i.e. values of y) from the end of the cone and plotting a graph of the values of

$$I_y\left(\frac{SSD + y}{SSD}\right)^2 \text{ versus } y$$

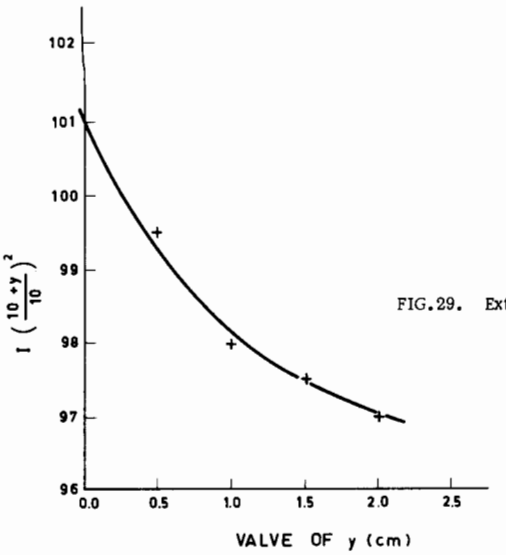

FIG. 29. Extrapolation to find output at end of treatment cone.

The curve so obtained is drawn and extrapolated to y = 0 (Fig. 29). This extrapolated value ($I_y = 0$) is used in the final calculation of output, viz:

$$\dot{X}_s = \frac{I_{y=0}}{m} \cdot \phi\,(p, t) \cdot K_E \cdot B \quad R/min$$

For example, the following results (Table XII) were obtained using a cone whose SSD was 10 cm. The beam size was a 5 cm circle and the HVT 2 mm Al. The tables give a surface backscatter (B) value of 1.15 for these conditions.

When plotted (Fig. 29) the extrapolated value of I at y = 0 is seen to be 101. This value is then used to calculate the output.

In the above example the measurements were made at 758 mmHg and 19° C using an ionization chamber whose calibration factor K_E at 760 mmHg, 20° C was 1.18. The true exposure time was 40 s. Hence the output is

$$\frac{101}{40/60} \times \frac{760}{758} \times \frac{292}{293} \times 1.18 \times 1.15 = 205 \text{ R/min}$$

An alternative method is to make the measurement with the ionization chamber half sunk into the surface of a solid phantom (Fig. 30). In this case the surface exposure rate is given by

$$\dot{X}_s = \frac{I_y}{m} \cdot \phi\,(p, t) \cdot K_E \cdot \left(\frac{SSD + y}{SSD}\right)^2 \quad R/min$$

i.e. the same as before but without the backscatter factor B. When using this latter technique it is, of course, necessary to make a separate measurement for each treatment cone. The nature of the solid phantom material is not critical. Polystyrene or Mix D is ideal. Perspex is acceptable.

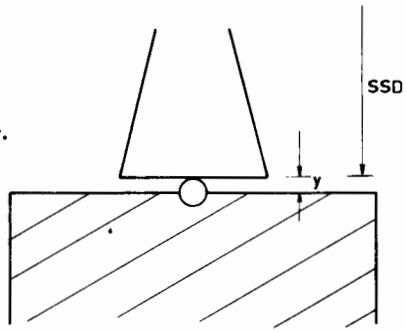

FIG. 30. Measurement of surface output with full backscatter.

TABLE XII. OUTPUT MEASUREMENT RESULTS IN CONTACT THERAPY

Average instrument readings I_y	y (cm)	$I_y\left(\frac{10+y}{10}\right)^2$
90.1	0.5	99.5
81.0	1.0	98
73.6	1.5	97.5
67.4	2.0	97

4.4.5. Grenz ray (10-30 kV) measurements

For these measurements it is necessary to have a dosemeter which is specifically applicable to these qualities and to the high output rates which this equipment often has. Furthermore, since this very soft radiation is substantially attenuated even by air, it is important for the chamber to be capable of being positioned at the level at which is required to know the dose rate. For these reasons the ionization chamber is usually in the form of a thin parallel plate chamber which has a very thin (about 1-2 mg per cm^2) front wall.

At these qualities the amount of scattered radiation is not large but it is usual to have the chamber incorporated into a phantom so that the total exposure (including back-scatter) is measured. As before the exposure rate is given by:

$$\dot{X}_s = \frac{I_y}{m} \cdot \phi(p, t) \cdot K_E \left(\frac{SSD + y}{SSD}\right)^2 \text{ R/min}$$

In view of the air attenuation referred to above, it is important to keep the value of y very small and if possible equal to zero.

Care must also be exercised in selecting the correct value of the calibration factor K_E. In particular the HVT must be measured close to the level at which the skin surface will be during treatment. It is obviously highly desirable to use a chamber whose values of K_E are as independent of quality as possible. Such independence of quality is possible only if the chamber has a very thin (ideally zero) wall thickness.

The requirements for Grenz ray measurements are very stringent and the measurements difficult to do with high accuracy. It is fortunate that accuracies of ±10% seem to be clinically acceptable for this kind of radiation therapy.

4.5. ROUTINE CHECK OF OUTPUT AND QUALITY

Due to various causes the output of an X-ray set may not remain constant in spite of the apparent consistency of the operating conditions. It is therefore necessary to check the output from time to time and either: (a) Quote a new set of values of output for all the different operating conditions; or (b) If a monitor is used, adjust the monitor sensitivity so that the outputs are maintained at their previously measured and stated values.

It is preferred to adopt method (b) if a monitor is fitted since it avoids changing the output information given to the treating radiographers (X-ray technicians), as well as obviating the work of recalculating all the outputs.

A simple but adequate method of checking that the radiation quality is unchanged is described in section 4.5.3. Such a check is necessary since it cannot be assumed that, even if the kV meter reading or the kV setting is unchanged, the actual voltage across the X-ray tube is unchanged. Nor can it be assumed that the inherent filtration is unchanged.

Provided the radiation quality has not changed the variation of output with beam size and SSD will not have changed. It is therefore reasonable to check the output for one set of operating conditions only and to make use of the information so obtained to carry out action (a) or (b) above.

4.5.1. Method of checking output consistency

It is recommended that the radiation output of an X-ray set should be checked regularly and frequently (see section 4.5.4). Such routine checking is facilitated if a fixed arrangement of the kind illustrated in Fig. 31 is utilized. It consists of a block of plastic material (Perspex, Polythene or polystyrene) about 20 cm cube. A hole into which the ionization chamber will fit snugly is drilled as shown. It is convenient if the depth of the centre of the hole is at about 5 cm below the surface of the block. The beam of

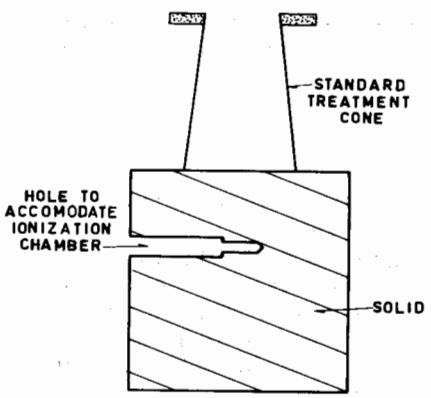

FIG. 31. Phantom used in standard output check.

radiation is directed centrally on to the block using one of the ordinary treatment cones, say a 10 × 10 cm beam at 50 cm SSD (or a selected beam size and SSD if a continuously variable diaphragm is used). The instrument used in these measurements need not be a calibrated one or even the one used for the output measurements. It should be one, however, whose consistency can be relied upon (i.e. one which is checked for consistency as described in Chapter 3) and preferably one for which the calibration factor (K_E) has been determined locally (section 3.5.6).

A measurement is made using this system on the same occasion as and with the same values of kV, mA and filtration as for the output determination. For example, the data given below were obtained on the same occasion as was the value of 75.4 rad/min in soft tissue for the peak dose rate for the 10 × 10 cm beam (section 4.2.7).

Consistency check

Dosemeter No. 3, calibration factor = 1.46 at 760 mm Hg, 20° C

Polystyrene block, 5 cm deep, 10 cm square beam, 50 cm SSD

1 min exposure 35.0, 35.6, 35.3, 35.1, 35.5 Mean 35.3

$2 \times \frac{1}{2}$ min exposure 34.8, 34.2, 34.6, 34.6, 34.7 Mean 34.6

Hence reading in 1 true min = 2 × 35.3 - 34.6 = 36.0

Temperature = 25° C Pressure 750 mm Hg

$$\text{Standard reading} = \frac{36.0}{1} \times \frac{760}{750} \times \frac{298}{293} \times 1.46$$

$$= 54.1 \text{ R/min}$$

This result is known as the consistency check value. If on a subsequent occasion it is found to have changed to say 52.0 R/min, i.e. to be 4% lower, it is safe to assume that the peak dose (and exposure) rates for all the beam sizes will also be lower by 4%. It is recommended that the results of the consistency checks be plotted as a chronological graph (as for the dosemeter radioactivity check) so that any tendency of the output to drift to a higher or lower value is obvious, as is any tendency of the output to be excessively variable from day to day. An example of such a plot is shown in Fig. 32. It

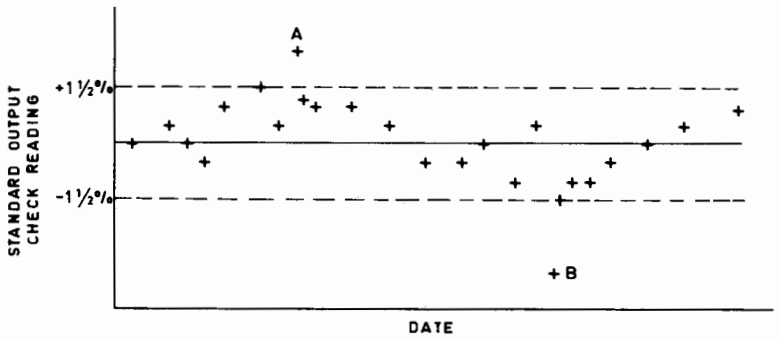

FIG. 32. Chronological plot of results of standard output check.

should be emphasized that the consistency check should be done during the course of and at various times during the working day when the equipment is fully warmed up and not at the beginning of the day.

4.5.2. Necessary adjustments

It may be difficult, on the basis of a particular consistency check, to decide when to change the quoted output values used for calculating treatment times. The chronological plot is helpful. If the check remains within $\pm 1\frac{1}{2}\%$, no action is needed. If the departure exceeds $1\frac{1}{2}\%$, the check should be repeated as soon as possible and a change made if the value remains outside $\pm 1\frac{1}{2}\%$. On any occasion when the check value is found to be outside ±3% of the one appropriate to the output value being used, the check must be repeated immediately and action taken immediately, if confirmed. These latter comments are intended only as guide lines and the user must be guided also by his experience and by the known behaviour of the equipment. For example, there can be little doubt that a change in the output values is needed at A (Fig. 32) since the consistency check value has shown a steady decrease over several weeks. On the other hand a bizarre value such as B, unless confirmed when repeated measurements are made, should not be the basis of a change. The check must be repeated and if it is not confirmed (and does not recur during subsequent weeks), it must be assumed to be due to some error of measurement.

4.5.2.1. Output table

If a monitor is not used, or even if one is used, it is common practice to continue to use the same operating conditions (i.e. kV, mA and filter) and to provide a revised list of outputs for all the various combinations of beam size and SSD available. The revised output values are calculated using:

$$\frac{\text{Revised output}}{\text{Previous output}} = \frac{\text{Present consistency check value}}{\text{Previous consistency check value}}$$

For example: the output of the 12 × 12 cm beam at 50 cm SSD was measured to be 77.4 rad/min and at that time the consistency check value was recorded as 53.8. The new consistency check value is 51.6 so that the revised output is calculated as

$$\text{Revised output} = 77.4 \times \frac{51.6}{53.8}$$

$$= 74.6 \text{ rad/min}$$

and similarly for all other beam sizes and SSDs.

4.5.2.2. Monitor sensitivity

If a monitor system is in use, the consistency check result is used to decide if and when to adjust the monitor sensitivity. This should be done so as to keep the consistency check value within $\pm 1\frac{1}{2}\%$ of the required value, using the ideas expressed above in conjunction with a chronological plot of

the results. For example, Fig. 33 shows how at A, the value having drifted steadily, the value is brought back within the permitted ±1½% whereas at B no change was made since the high reading was not confirmed.

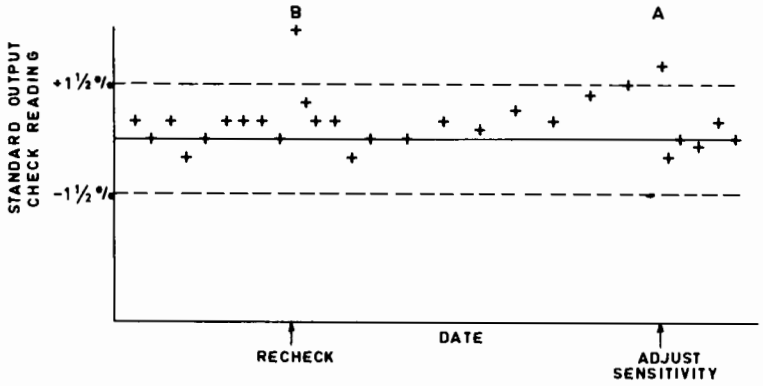

FIG. 33. Chronological plot of result of routine check of monitor sensitivity.

4.5.2.3. Operating conditions (mA)

An alternative method of working which may be adopted if a monitor system is not fitted is to change the mA at which the X-ray equipment is operated in order to keep the consistency check value within the ±1½% range. Any change is, as before, based on the chronological record. This method of working is convenient since it avoids having to prepare new lists of output values and recalculating treatment times etc. The very serious disadvantage of the method is that the X-ray tube current is changed from time to time and great care must be taken that the correct value is used. For example, suppose the peak exposure rates for a particular X-ray machine are measured and on the same occasion the consistency check value is found to be 54.1 R/min. Subsequently it is decided that the consistency check value has changed to 52.5 R/min and this is confirmed the next day. If the original tube current (mA) was 17.0, then it is necessary to change it to about

$$17.0 \times \frac{54.1}{52.5} = 17.6 \text{ mA}$$

A consistency check is now done using this new value of tube current which is adjusted further (if necessary) until a consistency check value close to the original value of 54.1 is obtained using a reasonable value of tube current. In this example it is most likely that a current of 17.5 mA will give a consistency check value within ±1½% of 54.1 R/min. Of course, it is often impossible or, at least, inconvenient to operate the X-ray machine at other than say, 10, 15, or 20 mA. In such circumstances this method is inappropriate.

4.5.3. Routine check of quality

For X-ray equipment it is also necessary to ensure that the radiation quality remains constant. By this is really meant that the percentage depth

dose values are unchanged. The quality (controlled by the actual kV across the X-ray tube plus the total filter) may be checked by remeasuring the HVT. A more directly useful and sufficient method, however, is to check that the ratio of the doses at two different depths remains constant. The plastic jig used for the routine check of output can also have a hole (into which the ionization chamber can be inserted) at say 15 cm deep. The dosemeter readings obtained with the chamber in these two holes (at 5 and 15 cm deep) alternately when given equal exposures at a selected beam size (say 10 × 10 cm) are noted and the ratio calculated. This should be done when the percentage depth dose values are checked (see Chapter 5) and output measurements made and should be repeated regularly. The consistency of this ratio indicates the consistency and therefore applicability of the percentage depth dose data. Naturally a plug should be fitted into the empty hole when a measurement is being made with the ionization chamber in the other.

4.5.4. Frequency of routine checking of quality and output

The frequency with which the output and quality of X-ray equipment needs to be checked depends very much on the particular type of apparatus being used. The frequency can be determined only on the basis of experience. When a new apparatus is installed, it is reasonable to check the output each day in the way described above. When it is apparent that the output is reasonably constant, then the intervals between checking can be lengthened. It is good practice to check the output of an X-ray set at least once each week and certainly never less frequently than once per month. As indicated above, however, the frequency must be determined on the basis of experience.

CHAPTER 5

MEASUREMENT OF RADIATION QUALITY

The quality of radiation must be known for the following reasons:
(1) To choose the appropriate percentage depth dose data
(2) To select the calibration factor (K_E) for the measuring dosemeter
(3) To select the rad per röntgen factor (f_λ)
(4) To estimate the absorption in or the shielding by bone and the extra transmission by lung.

The quality of the primary radiation incident on the patient is controlled by the kV and the filtration (inherent + added). For the present purposes the quality is satisfactorily stated in terms of the half value thickness (HVT) for X-rays generated at voltages below 400 kV. For ^{60}Co and ^{137}Cs γ-rays the quality is known.

Since the HVT used for the specification of % depth dose data etc. is that for a narrow beam of radiation, only one determination of HVT for each radiation quality (i. e. kV plus filtration) used is needed.

5. 1. MEASUREMENT OF HALF VALUE THICKNESS

For the purposes listed above the HVT needs to be known only to an accuracy of about 0. 2 mm and measurement to this accuracy poses no serious experimental problems. The half value thickness is the thickness (in mm) of the stated material (aluminium up to about 150 kV, copper above about 150 kV) which reduces the exposure rate of the radiation beam to one half of its unattenuated value. The determination is conveniently made by measuring and plotting on a graph the reduction in exposure rate as successive thicknesses of the attenuating metal are inserted into the beam (Fig. 34). The HVT is then determined by graphic interpolation.

5. 1. 1. Practical aspects

Figure 35 indicates the general experimental arrangement. The X-ray tube (1) is directed towards the ionization chamber (2) held in air at a constant distance from the X-ray tube.

There should be a space of at least 50 cm between the ionization chamber and any structure behind it that is coincidentally irradiated during the measurements. The size of the beam at the chamber is limited by means of the lead diaphragm (3). It is convenient and good practice to collimate the beam also by means of a treatment cone or an adjustable collimator (4) so that the beam size at the lead diaphragm is only 1-2 cm larger than the hole in the diaphragm. Sheets of the attenuating material are positioned as shown (5) immediately adjacent to the diaphragm which is approximately midway between the X-ray tube target and the ionization chamber. For Grenz-ray (i. e. less than 20-30 kV) measurements the chamber should be at a distance from the target as nearly equal as possible to the treating distance used. This is because for these soft qualities the intervening air can act as a filter. For other X-ray qualities the

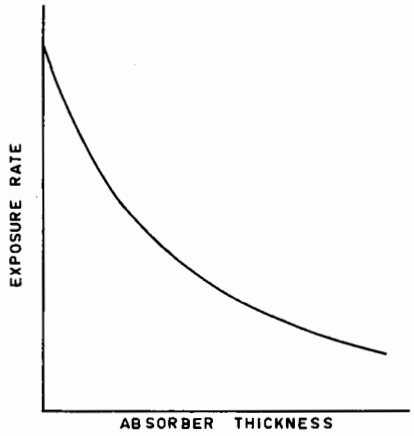

FIG. 34. Attenuation curve.

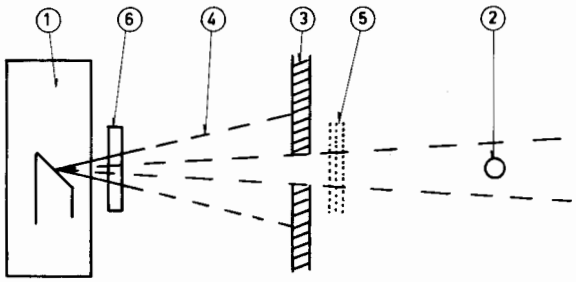

FIG. 35. Experimental arrangement for the measurement of HVT.

1. X-ray tube
2. Ionization chamber
3. Lead diaphragm
4. Treatment cone
5. Attenuators
6. Monitor chamber

distance between the target and the chamber should be about 10 times the diameter of the beam at the position of the chamber.

The following important points of technique must be noted:

5.1.1.1. Constant exposure

The values of measured transmitted radiation that are to be plotted as the ordinate of the graph (Figs. 34 and 38) refer to the situation when the radiation incident on the attenuator at the lead diaphragm is constant. This is best achieved if the X-ray tube is fitted with a monitor chamber (6) system that can be used to ensure that all the exposures made are identical (or that can be used to normalize the measurements should the exposures not all be the same). In the absence of a monitor it is important to keep the kV, mA and exposure time the same for each separate exposure.

5.1.1.2. Lead diaphragm

The lead (or other material, e.g. steel) should be sufficiently thick to reduce the radiation intensity to less than 1%. For voltages up to 300 kV 3 mm of lead is adequate.

The size of the hole in the diaphragm should be such that the beam size at the position of the chamber is only a little larger than the size of the chamber, i. e. the beam size is a minimum consistent with fully irradiating the chamber. It is suggested that the size of the beam at the chamber be such that it extends one centimetre around the chamber (Fig. 36).

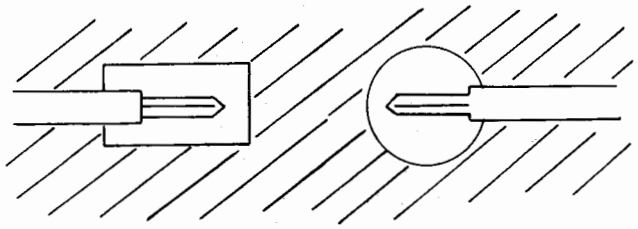

FIG. 36. Beam size just larger than ionization chamber.

It is worth confirming that the chamber is fully irradiated by placing an X-ray film behind the chamber and making an exposure before the measurements are started; Fig. 37 is such a radiograph.

5. 1. 1. 3. Ionization chamber

Ideally the measurements should be made with a chamber whose sensitivity is independent of radiation quality since the quality at the chamber may become progressively harder as the absorber thickness is increased, due to filtration by the absorber. Fortunately, for most reasonable circumstances and ionization chambers the change in sensitivity with the (small) change in quality that occurs is unlikely to be important and can be ignored. The values plotted on the ordinate of the graph (Fig. 38) are therefore 'instrument readings'.

5. 1. 1. 4. Absorber

The absorber is conveniently in the form of sheet material of a suitable thickness. It is found that thicknesses of one quarter, one half and one millimetre are suitable. It is important that the amount of any impurity present that has an atomic number much higher or lower than that of the copper or the aluminium should be small. The specifications for appropriate copper and aluminium are:

(a) Copper: less than 1% total impurities; density 8. 93 $g \cdot cm^{-3}$

(b) Aluminium: less than 0. 2% total impurities; density 2. 7 $g \cdot cm^{-3}$

For Grenz-ray measurements the purity of the aluminium used must be very high (< 0. 01% total impurity) and for this reason a specified plastic is sometimes preferred.

5. 1. 1. 5. Experimental technique

Readings of the measuring instrument should be taken for a range of attenuator thicknesses such that the maximum thickness used results in a reading which is less than 20% of that for the unattenuated beam.

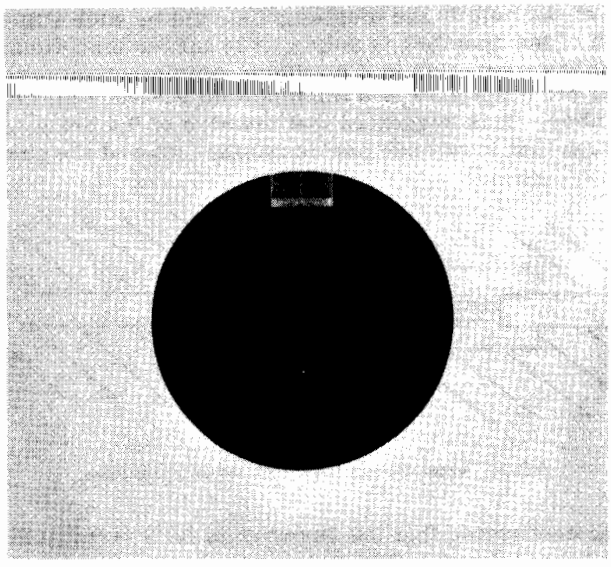

FIG. 37. Radiograph of ionization chamber in beam.

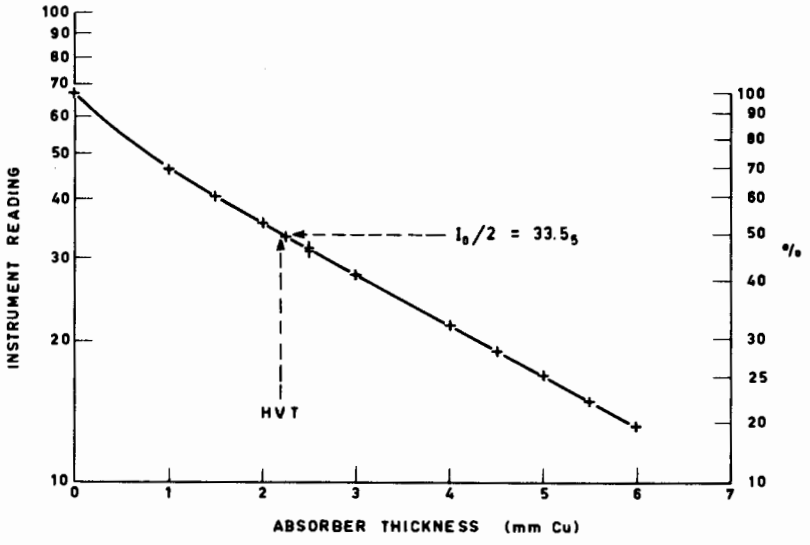

FIG. 38. Attenuation curve: log plot.

Several readings should be taken for the normal beam (i. e. no added attenuators) and it is helpful if readings can also be made with thicknesses of attenuator that are close to the half value thickness. Table XIII shows a series of experimental values recorded in chronological order. The values should be plotted on a graph (Fig. 38) as they are obtained. It can be seen from the data how the experimenter has used the earlier values to lead him to make measurements with attenuators very close to the half-value thickness. Finally measurements are made with an attenuator of

TABLE XIII. EXPERIMENTAL VALUES OBTAINED IN HVT MEASUREMENT

Attenuator thickness (mm Cu)	Instrument readings	
0	66.4	
1		46.4
2		35.5
0	67.6	
3		27.8
4		21.6
0	67.8	
5		16.8
6		13.1
0	67.0	
1.5		40.3
2.5		31.6
0	66.8	
4.5		19.0
5.5		14.8
0	66.8	
2		35.4
2.5		31.2
2.25		33.1
0	67.3	

Mean value for unattenuated beam = 67.1

half value = 33.55

Hence interpolated HVT from graph (Fig. 38) = 2.2 mm Cu

thickness close to the half value thickness and again for the unattenuated beam. In Fig. 38 the instrument readings are plotted on a logarithmic scale and this is the usual practice since the part of the curve in which greatest interest is centred is then almost linear. The interpolated value for the half value thickness is, from Fig. 38, 2.2 mm Cu.

5.1.2. Second half value thickness

The 'second half value thickness' is the extra thickness required to reduce the instrument reading from one half to one quarter of the value for the unattenuated beam. In the example of Table XIII and Fig. 38 the thicknesses required to reduce the value to 50% and 25% are 2.2 and 5.0 mm respectively. The second half value layer is therefore equal to

5.0 mm - 2.2 mm = 2.8 mm Cu

TABLE XIV. CONFIRMATION OF % DD VALUES

Depth (cm)	Beam size 4 × 5 cm (a)	(b)	(c)	8 × 10 cm (a)	(b)	(c)	15 × 20 cm (a)	(b)	(c)
5	62.6	51.8	51.8	81.3	61.9	61.9	91.7	71.1	71.1
10	29.9	22.9	22.9	41.6	31.3	31.7	52.0	41.3	40.4
15	13.0	10.0	9.9	20.8	15.2	15.8	28.0	22.7	21.7

Column (a) experimental dosemeter reading; (b) % DD from tables; (c) values from column (a) normalized to agree with those of column (b) at a depth of 5 cm for each beam size separately.

The ratio first HVT/second HVT is called the 'homogeneity coefficient' and in this example the value is:

$$\text{Homogeneity coefficient} = 2.2/2.8 = 0.79$$

The first and second half value thicknesses and the homogeneity coefficient are used to give additional information about the quality of the radiation. If it is found that the measured % depth doses (see section 5.2) do not agree with data published for the same HVT, this may be accounted for by the radiations having different spectra in spite of their first HVT's being the same. The value of the homogeneity coefficient will indicate whether this is so or not.

5.2. CONFIRMATION OF SELECTED % DEPTH DOSE DATA

Many sets of % depth dose tables are available in the literature and, although most must be regarded as acceptably accurate, the most comprehensive, consistent and widely used ones are those contained in Supplement 10 of the British Journal of Radiology [3]. Isodose charts compatible with and based on these data have also been published (Ref. [7]). However, the following comments apply to any data.

In the % DD tables values will be found for a range of square and rectangular beam sizes. If data are required for sizes of rectangular beams that are not included in the tables, these can be obtained by using the method of equivalent squares. Reference [3] includes a table which gives for a wide range of rectangular beam sizes and shapes the size of the square beam that has the same central axis % DD and TAR values as the rectangular beam.

The decision to use published data must be based on, at least, a few measurements made on the user's equipment. The adoption of the 5 cm deep scheme minimizes any errors which result from the data used being not exactly correct.

The two parameters which need to be known before the % depth dose tables can be selected are the source-skin distance (SSD) and either the identity of the radioactive isotope (i.e. ^{60}Co or ^{137}Cs) or the HVT (for X-rays). The measurement of HVT is described above. The dependence of % DD on SSD is not great so that the value of SSD does not need to be known with high accuracy. It is therefore satisfactory to accept the manufacturer's stated value, although it is worth checking from a drawing that it is correct to within 1 or 2 cm.

Using these two parameters an appropriate set of tables is selected and their validity examined as follows:

(1) An ionization chamber is placed, with its centre on the central axis of the beam at depths of 5, 10 and 15 cm successively, inside the water phantom and measurements made at these depths for a small, medium and large beam size (e.g. 4×5, 8×10 and 15×20 cm rectangular beams). For these measurements the phantom must be sufficiently large that an increase in its size will not result in an increase in the instrument readings. In practice it is satisfactory if the phantom extends 5 cm beyond the beam edges and 5 cm beyond the maximum depth at which measurements are made.

(2) Examples of the values of instrument readings so obtained are tabulated in column (a) of Table XIV, as are the values of % DD (column (b)) for these depths and beam sizes taken from the tables of data which have been chosen for clinical use.

(3) The instrument readings (column (a)) are then scaled to agree with the % DD values at 5 cm deep and presented in column (c). For example, for the 4 × 5 cm beam (Table XIV) the values of column (a) are multiplied by 51.8/62.6 to obtain the values in column (c).

(4) If the disagreements between the values in columns (b) and (c) are less than 2-3% (of the peak value = 100%), the selected data can be regarded as acceptable for clinical use. If there is significant disagreement, then alternative data must be selected. The data presented in Table XIV are actual results obtained using a 250 kV constant potential X-ray unit fitted with treatment cones. The nine measurements were made in less than 1 hour. The % DD data were taken from page 22 of Ref. [3] which did not include measurements on this particular equipment in its survey. The figures in columns (b) and (c) agree to about 1% (of the peak value) and this demonstrates that the chosen % DD data are applicable also to this X-ray equipment.

5.3. SELECTION OF ISODOSE CHARTS

The selection of appropriate isodose charts is a much more difficult problem since the pattern depends not only on the beam size, SSD and radiation quality but also on the source size and on the construction of the collimation system. This is particularly so for telecobalt and telecaesium equipment for which many different designs of collimators and source sizes are available. An attempt should be made to obtain isodose charts which have been prepared from measurements made on the same kind of equipment as that being used. Even so it is worth making a few measurements across the beam for one or two beam sizes, especially in the region of the penumbra, in order to check that the actual pattern agrees with that on the chart to be used. The IAEA Atlases (Refs. [4, 7]) previously referred to contain a large selection of charts.

CHAPTER 6

THE TELETHERAPY EQUIPMENT IN GOOD ORDER AND ADJUSTMENT ACCEPTANCE TESTS

Before any X-ray, telecaesium or telecobalt equipment is used for detailed measurements and certainly before any clinical use, it is necessary to confirm that it is in good working order and that it is safe from the radiation, mechanical and electrical points of view. It is necessary to ensure that the beam of radiation produced by the machine is of the correct size, shape, position, orientation and direction. Furthermore, suitable mechanical, electrical and radiation tests must be repeated with sufficient frequency to ensure that the equipment remains in good order.

The details of what needs to be done are very dependent upon the exact design of the X-ray or telecobalt equipment. It is therefore, unfortunately, not possible to make completely specific and unambiguous recommendations. In this chapter care has been taken to state what needs to be checked and to give some guidance as to how this can be achieved. The exact details of what to do must be left open to the reader's common sense, at least, to some extent.

Although sufficient instructions are given here to enable the user to do the tests himself, the subsequent adjustments may be beyond his capability. Every effort should be made to ensure that at the time of installation of the equipment the manufacturer leaves it in complete adjustment and good working order.

6.1. EQUIPMENT IN A SAFE AND GENERALLY SATISFACTORY CONDITION

6.1.1. Mechanical

The head movements and rotations must be smooth and free, the locks must work satisfactorily and hold the equipment firmly. In particular, the equipment must be satisfactorily and safely counterbalanced so that there is no possibility of sudden uncontrolled movement whenever any of the control locks are released. This must be true for all positions and conditions of the equipment and is essential in view of the very real possibility of the heavy treatment head falling onto a patient due to failure of a lock or of the counterbalancing mechanism. The equipment must be stable in any position into which it is put. In particular, the rotation of the treatment cone and its mounting or that of the collimating system should be smooth and free from lateral movement (wobble).

6.1.2. Electrical

The electrical safety depends largely on the competence of the engineers installing the equipment, but a check must be made that in all possible circumstances any parts of the unit accessible to the patient or

staff are satisfactorily electrically earthed (grounded). Needless to say the equipment should be adequately fused and the power supply should be sufficient for the electrical demands of the equipment. The supply should be stable or a stabilizer fitted. Any local or national regulations concerning electrically operated equipment must be observed fully.

Particular attention should be given to any high-tension cables. These must be of the shock-proof type and should be firmly clamped so that no excessive movement takes place but they should be capable of moving freely whenever the tube head needs to be moved. Care should be taken that the cables are not rubbed which may cause the outer covering to become frayed or otherwise damaged. Furthermore, the cables should never be allowed to be pulled so that tension exists in them nor should they be bent into an arc of small radius. The cables should be inspected frequently to confirm that they are free from external, visible damage and that they move freely.

It is usual and highly desirable for there to be a clear indication of the kilovoltage at which the equipment is operating and of the added filtration. In addition some equipment is fitted with an electrical interlock system which restricts the operation to a limited range of specified combinations of kilovoltage and filtration. This feature is desirable especially for low-voltage X-ray equipment where the use of an incorrect filter can result in extreme errors of dosage to the patient and for equipment where a variety of kilovoltage and filter combinations are being used (although this latter is not recommended). The indicating and interlock systems must be tested to see if they operate satisfactorily when more than one mistake has been made by the operator: failure to protect against multiple errors is a common fault in design.

The accuracy and consistency of the timer must also be checked. This is described in section 2.2.3.

6.1.3. Radiation

The detailed properties of the beam of radiation have been partly described in Chapter 4 and will be dealt with more fully later in this chapter. The protection aspects are dealt with in Chapter 7.

6.1.4. General

The design of the equipment should be such that any failure, from either the electrical, mechanical or radiation point of view, will leave the equipment in a perfectly safe condition.

Although the detailed radiation measurements described elsewhere will demonstrate any malfunctioning of the apparatus, it is clearly not worth doing this work until one is reasonably satisfied that the machine appears to be working acceptably well. By this is meant that the various meters read satisfactory values, that they are stable and that they read zero when the equipment is switched off. The ON and OFF switching mechanism and the timer (if fitted) must operate satisfactorily. When a series of exposures is made the radiation output should be the same for each (see section 3.6).

It is at this stage that, all too frequently, the X- or γ-ray equipment is regarded as satisfactory and detailed measurements or even patient

treatments are started. This is very wrong and further testing is essential to ensure that the radiation beam is acceptable.

6.2. EQUIPMENT IN CORRECT ADJUSTMENT

It is essential to know accurately the size, shape, position and direction of the radiation beam from the machine. The line on the isodose chart (Fig. 39) about which the beam is symmetrical is known as the 'central axis' and lies in the direction in which the beam is said to point. Of course, unless the radiation source is positioned properly with respect to the treatment cone or collimating system, the beam will not be symmetrical about this axis. The unit needs to be fitted with some device to indicate to the treating technician (radiographer) the position and direction of the central axis. Usually this will be done by treatment cones for kilovoltage X-ray equipment or by an optical or mechanical front and back pointer for telecobalt equipment.

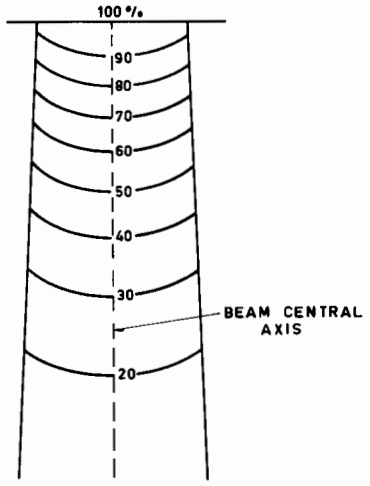

FIG. 39. Isodose chart.

It is also essential that the central axis should coincide with the axis of rotation of the treatment cone or collimating system. Unless this is so there will be a sideways movement of the beam whenever the beam orientation about the central axis is changed and this is clearly undesirable.

Although not very common with X-ray equipment, it is usual for telecobalt equipment to be fitted with a 'light-beam diaphragm' that gives visible light simulation of the gamma-ray beam. The light source and mirror (if fitted) of this system must be carefully positioned so that the light beam gives an accurate representation of the gamma-ray beam (see also section 6.2.1.6).

The situation that must exist for X-ray (a) and telecobalt (b) equipment to be in correct adjustment is shown in Fig. 40. The requirements are:

(1) There is an axis of rotation of the collimating system or treatment cone. This axis coincides with the beam 'central axis' and corresponds to the centreline on the isodose chart, normal to the surface. The

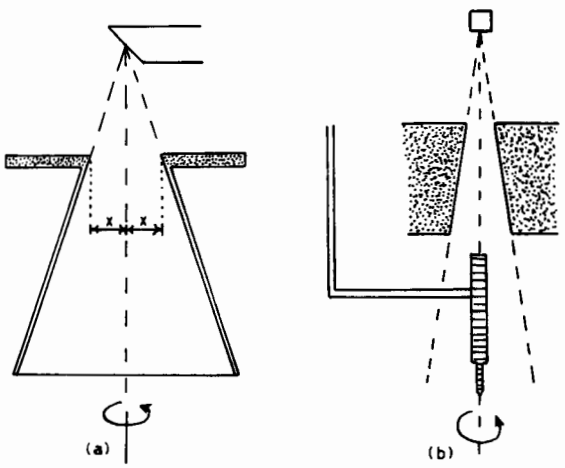

FIG. 40. Equipment in good adjustment: (a) X-ray; (b) γ-ray.

central axis is indicated either by a treatment cone or by mechanical or optical pointers. Rotation of the treatment cone or collimator system therefore takes place about the central axis.

(2) The treatment cone diaphragm or collimator jaws are symmetrical with respect to this axis for all beam sizes.

(3) The focal spot of the X-ray machine or the centre of the source of the cobalt equipment must be on the central axis and, in particular if it is a 'moving source' unit, the source must always return to this position.

(4) The field illuminating light is centred on the central axis either directly or indirectly via the mirror.

(5) The numerical readings associated with various beam sizes on the light-beam diaphragm or collimator system must indicate the 'geometric beam size' or other specified beam size.

(6) All the above requirements should be fulfilled for all orientations of the equipment.

Not only must the unit be checked initially but these checks should be repeated routinely at sufficiently frequent intervals to confirm that the unit remains in correct adjustment.

As stated earlier, in view of the many different designs of equipment available it is not possible to give unambiguous instructions as to how it may be checked to ensure that it is in correct adjustment. The following however, sets out the principles of what should be done. The order of presentation is important since some of the tests rely upon a previous test having been done and the equipment being correct from that point of view. If a different order is adopted, there is no guarantee that the end result will be satisfactory. For this reason it is worth repeating all the tests at the end without making any further adjustments to confirm that all is well. It is convenient to consider X-ray and telecobalt equipment separately.

6.2.1. X-ray equipment

6.2.1.1. Axis of rotation of the treatment cone

Check that the rotation of the treatment cone, or its seating, is smooth and without lateral play. Check also that the treatment cone fits into the

seating without lateral play. If necessary, adjustments should be made until the combined lateral movement at the end of the treatment cone is less than ± 1 mm and preferably less than ± $\frac{1}{2}$ mm. The axis of rotation of the diaphragm system or seating of the treatment cone is usually the only definite and observable location and the tests described below serve to confirm that it coincides with the central axis.

6.2.1.2. Central axis indication

All points on the beam central axis remain stationary as the treatment cone or its seating or the light-beam diaphragm system is rotated about the central axis. The marks on the treatment cone or the tip of the pointer which indicates the position of the central axis must therefore remain stationary as the collimating system is rotated. This is easily tested by mounting a stationary pointer with its tip very close to the

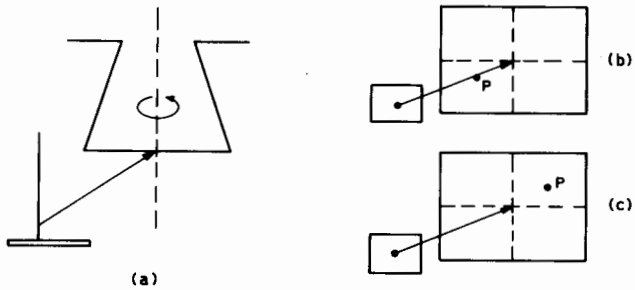

FIG. 41. Identification of central axis: (a) pointer fixed in space; (b), (c) treatment cone rotated through 180°.

central axis (Fig. 41). There should be no relative movement between the tip of the stationary pointer and the central axis indicator as the collimating system is rotated. Adjustments should be made until this is so. Figure 41 (b) and (c) demonstrates a treatment cone in correct adjustment. The tip of the pointer remains coincident with the intersection of the beam axis (which indicates the centre of the beam) for two positions of the cone 180° apart, whilst the off-axis point, P, moves with respect to the pointer. Check that the indication of any front or back pointer, whether optical or mechanical, is coincident with this axis at all SSD's.

6.2.1.3. Focal spot

The focal spot must be on the central axis and this is checked most easily by taking a pin-hole picture. A sheet of lead about 2 mm thick with a pin hole (about $\frac{1}{4}$ mm diameter) at its centre is attached to the treatment cone seating so that it rotates with it. The sheet of lead is positioned so that the pin hole is as near as possible to the axis of rotation. This can be tested by placing a fixed pointer near to the pin hole and observing that the pin hole does not move laterally as the treatment cone seating is rotated. The treatment cone is then put in position and a cross made of lead wire (electrical fuse wire will serve for this purpose) placed centrally on the end of the cone (this can be tested by rotating the cone and observing that the lead cross does not move laterally – as for the pin hole).

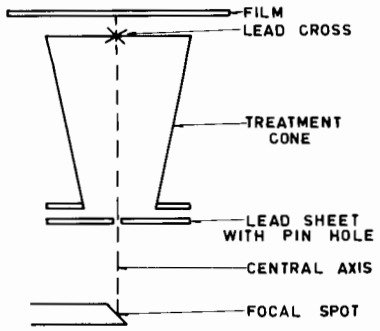

FIG. 42. Apparatus and arrangement to make pin-hole picture of focal spot.

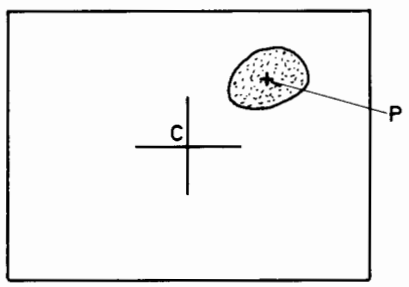

FIG. 43. Pin-hole picture of focal spot.

A film (a non-screen film in an envelope) is then held in a clamp close to the end of the cone (Fig. 42) and an exposure made.

The film will have the general appearance illustrated in Fig. 43. The real distance between the central axis and the centre of the focal spot is calculated by multiplying the distance on the film between the image of the centre of the cross wires (C) and the centre of the pin-hole image (P) of the target by the ratio H/A where A is the distance of the film from the pin hole and H is the distance of the pin hole from the target (Fig. 44).

In practice it is difficult to position the pin hole and cross wires exactly on the central axis. This problem can be solved by using the following procedure. The apparatus is set up as before and an exposure made. The treatment cone, cross-wires and lead sheet with the pin hole are then carefully rotated through 180°, the film remaining stationary. Another exposure identical with the first is now made and the film processed

The film will have the appearance illustrated in Fig. 45. There are two images of the cross wires and two images of the focal spot. The centre of the line joining the centres of the two images of the cross wires is marked on the film as is the centre of the line joining the centres of the two images of the focal spot. The distance between the mean centres of the focal spot images and the mean centres of the cross wire images is multiplied by the ratio H/A (Fig. 44), the result giving, as before, the real distance between the central axis and the centre of the focal spot.

It is desirable that the focal spot should be within 2 mm of the central axis. If it is more than 2 mm off axis, the possibility of improving the situation should be examined. On some equipment it is possible to move the axis of rotation of the collimating system (treatment cone) and in such circumstances the adjustment is relatively easy. The film test must be

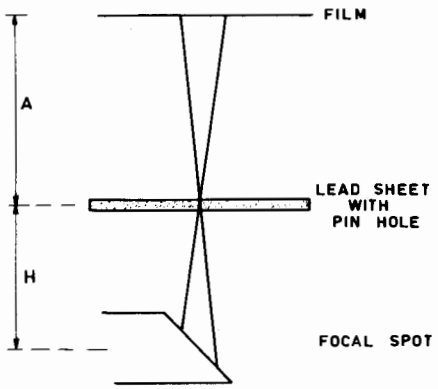

FIG. 44. Geometry of pin-hole picture of focal spot.

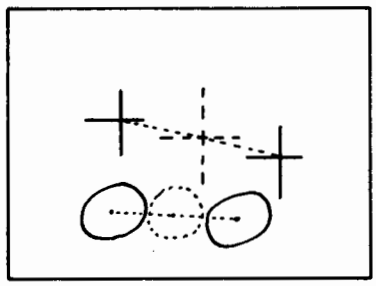

FIG. 45. Doubly exposed pin-hole picture used to test centring of focal spot.

repeated after the adjustment. If the axis cannot be moved, the only solution is to move the focal spot. This is usually a task for the manufacturer or installation engineer.

If it proves impossible to adjust the equipment so that the focal spot is within 2 mm of the axis, particular attention should be given to the size and uniformity of the beam (see section 6.2.1.5). It must also be realized that the beam direction will be incorrect in these circumstances to an extent depending on the SSD being used and on the amount by which the spot is off axis (Fig. 46).

The blackened area on the film also gives useful information about the character of the focal spot, of which it is a magnified (magnification = A/H) representation. Ideally the focal spot area on the film should be uniformly black but this is rarely possible and usually some pattern is present. A check should be made that the dimensions of the spot are consistent with the manufacturer's specification. If it is more than about 20% larger, there is a danger that the beam size will be smaller than intended (see below). If the spot is smaller than that specified by the manufacturer, the heating of the target may be excessive which can lead to premature roughening of the target surface and consequent curtailment of tube life. In this connection it is even more important to ensure that the pattern of blackening is reasonably uniform over the area of the focal spot. Small areas of substantially greater blackening than that of the remainder of the focal spot area are indicative of local hot spots which can have a disastrous effect on tube life. It is not usual these days to obtain from a new X-ray tube focal spot pictures which exhibit these defects. It is useful, however, to take focal spot pictures with the new X-ray tube and retain them for

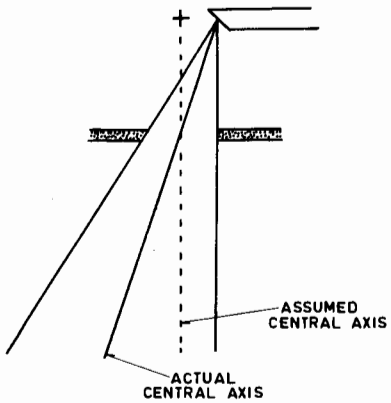

FIG. 46. Effect of non-central focal spot.

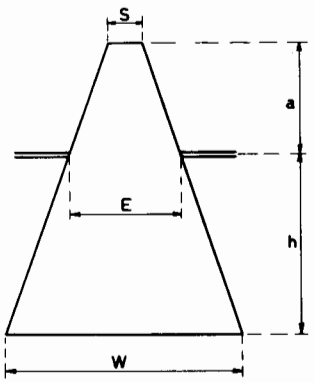

FIG. 47. Geometry of aperture size.

comparison with ones taken later in the tube's life. During the lifetime of the tube the appearance of the spot will change and these changes can give useful information about the condition of the tube filament and target surface. In particular the likely failure of a tube in the not too distant future can be predicted from changes in the uniformity of the blackening. Such advance warning can be most useful if there is likely to be delay between ordering and receiving a replacement.

6.2.1.4. Treatment cone diaphragm

The size of the defining aperture in the cone should be checked to confirm that it is of the correct dimension. Figure 47 shows that

$$E = \frac{Wa + Sh}{a + h}$$

where E is the aperture size; W is the beam size at the patient (i.e. at applicator end); a is the target-aperture distance; h is the aperture-applicator end distance; and S is the apparent focal spot size.

In practice the aperture is usually and correctly made a little larger than this, say by 1 or 2 mm, the exact size of the beam at the patient's

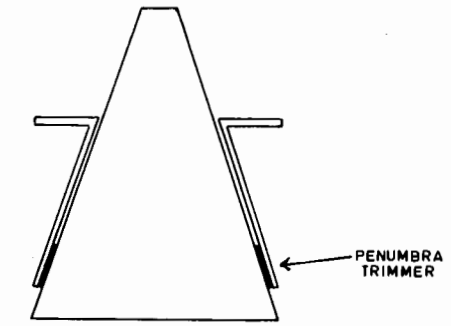

FIG. 48. Penumbra trimmers.

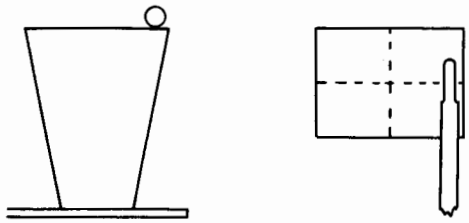

FIG. 49. Ionization chamber mounted on one end of treatment cone.

skin being controlled by the sides of the cone or by the trimmers (Fig. 48) placed close to the end of the cone.

Having checked that the aperture is approximately of the correct size, an X-ray film is then placed in contact with the end of the treatment cone and an exposure made. This should be such that the film is not very black (density about 1.5) so that it can be clearly seen either directly or by using a densitometer whether or not the whole of the field is being uniformly illuminated with X-rays. If there is any portion of the field that is not fully illuminated, a check should be made (by exposing a film in the same way but without the cone in place or at least with one of a much larger size) to ensure that no other structure (say the hole in the shutter or the master collimator) is responsible for the observed 'cut off'. If there is no such structure, the size of the aperture in the cone must be adjusted until a fully illuminated field is achieved.

6.2.1.5. Beam uniformity

An ionization chamber should be fastened to the base of one of the larger treatment cones so that it is about 1-1½ cm inside the edge of the beam (Fig. 49); for example, for an SSD of 50 cm the largest cone is likely to give a rectangular beam of 15 × 20 cm. The chamber is therefore fastened as shown in Fig. 49 at a distance of 8½ cm from the centre. No backscatter material is added so that the exposure is in air.

An exposure is made and the corresponding reading of the dosemeter observed and noted. The cone is then rotated through 45° and the exposure etc. repeated. This is continued at 45° intervals until a full rotation of 360° has been covered, the exposure rates at the points indicated by Fig. 50 thus being obtained. The chamber is now re-positioned in the centre of the beam and another identical exposure made. The results should show that:

(a) All the readings at 45° spacings are identical with ± 3%

(b) Their mean value is not less than 90% of the central value.

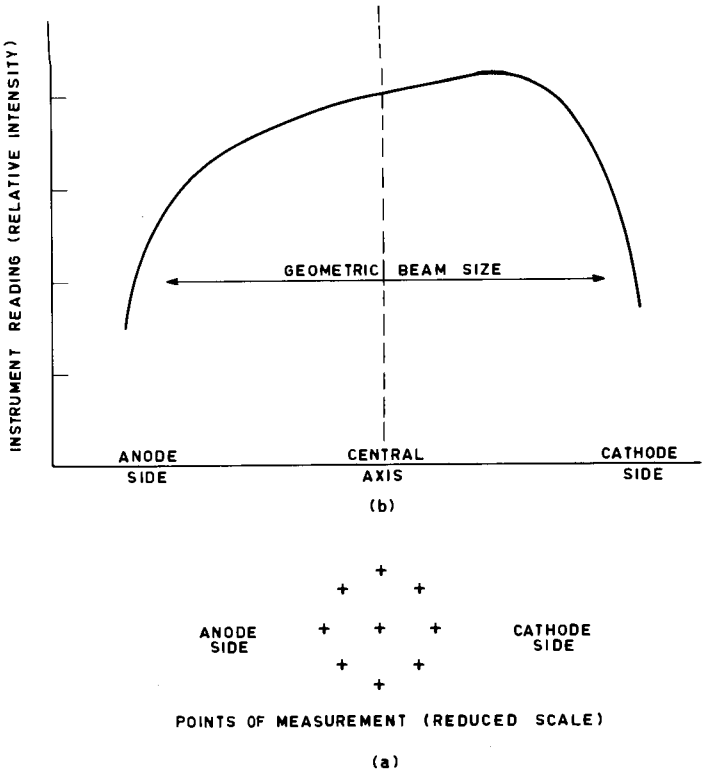

FIG. 50. Distribution of radiation in plane normal to central axis: (a) points of measurement; (b) intensity variation in anode-cathode direction.

If these conditions are not met, then either the focal spot is not correctly centred (see section 6.2.1.3) or a wedge or conical flattening filter is required. The design of such a filter although not difficult is outside the scope of this manual.

The making of such measurements, at a single distance from the centre, is the minimum which should be done. It is preferable to do this kind of measurement at several distances from the central axis (say at about $1\frac{1}{2}$ - 2 cm intervals) and hence be able to plot graphs that represent the pattern of dose along lines in various directions normal to the beam central axis. Figure 50 shows such a plot which in this case is in a direction parallel to the anode-cathode axis of the X-ray tube. It can be seen that the beam is not quite symmetrical but can be regarded as acceptable since the dose $1\frac{1}{2}$ cm inside the geometric beam edge is greater than 90% of the central value and the maximum dose is only 3% higher than that at the centre.

6.2.1.6. Light-beam diaphragm

If a light-beam diaphragm is fitted, the tests described above still need to be done. The aperture between the jaws of the diaphragm unit takes the place of the aperture in the top of the treatment cone. The

measurement of the beam flatness is slightly more difficult to do since a separate support for the ionization chamber must be provided in order to measure the dose at the points required.

A further test to confirm that the position and size of the visible light area corresponds to the position and size of the subsequent X-ray area needs to be performed at each diaphragm setting. This is easily tested by marking with lead wire a square (say 10 × 10 cm) on the surface of an envelope-wrapped film and positioning it at the required SSD. The diaphragm scale is then set to 10 × 10 cm and the illuminated light area made to coincide with the area marked by lead on the film. An exposure is made and the processed film examined to check that the X-ray patch is correctly positioned with respect to the shadow cast by the lead wire, which has been previously shown to agree with the visible light indication. The test is repeated for smaller and larger beam sizes.

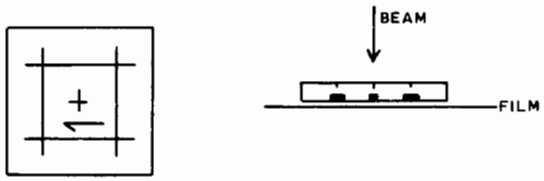

FIG. 51. Device used to check coincidence of optical and X- or γ-ray beams, and their symmetry with respect to the central axis.

A test jig of the form shown in Fig. 51 is useful. It consists of a sheet of opaque Perspex on the underside of which are milled slots into which lead wire is pressed. Lines exactly overlying the slots are drawn on the upper side of the Perspex sheet. The film is positioned below this test device. The lines on the upper surface are used to test the visible light area on the X-ray exposed film. Figure 52 is an example of such a film. In this case only the corners of the rectangles are marked in lead wire since this helps to see exactly where the edge of the X-ray area is, and the asymmetric arrow is used to indicate the orientation of the film with respect to the X-ray tube.

6.2.2. Telecaesium and telecobalt equipment

The tests to be done on this equipment are almost the same as for the X-ray equipment but require modification because of the, usually, larger source size and to the practical impossibility of taking a pin-hole picture to check that the source is centred.

6.2.2.1. Axis of rotation of collimators and central axis indication

The tests required are essentially the same as for the X-ray equipment and, if a treatment cone is available, they can be carried out exactly as for the X-ray equipment. It is not usual, however, for a treatment cone to be available for telecobalt equipment so it is more convenient to test for the existence of an axis of rotation and to check the accuracy of the indication of the central axis at the same time. For equipment on which the front pointer (optical or mechanical) rotates with the diaphragm system,

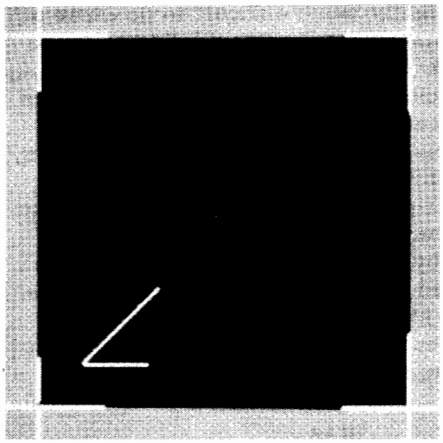

FIG. 52. Example of a film as described in section 6.2.1.6.

the pointer can be used to observe that there is no lateral movement (wobble) of the diaphragm system. The pointer is adjusted so that at all distances from the source (say over a range of distances of SSD to SSD + 30 cm) the tip (or light spot) of the pointer remains stationary on rotation. A similar test and adjustment, if necessary, are done to the back pointer. If the front pointer does not rotate with the diaphragm system, a pointer must be clamped to the diaphragm and this used to test for wobble, its indicating tip being adjusted so that it is stationary on rotation of the diaphragms. The actual front pointer (mechanical or optical) is then adjusted to agree with this pointer. The tests must be repeated over the distance range, SSD to SSD + 30 cm.

6.2.2.2. Source centring

This may be confirmed in either of two ways:

(a) By film: The diaphragm system is rotated so that the major and minor axes of the beam (i.e. the beam length and width) each lie in a convenient direction. The rotation scale reading is noted. The choice of a convenient direction depends upon the design of the telecobalt equipment under test. If facilities are provided for the position of the source to be adjusted with respect to the axis of rotation of the diaphragm system (or vice versa), the convenient direction will be that for which the beam major axis lies parallel to the available direction of movement for the source. The diaphragms are set to give a beam of 10×10 cm at the standard treating distance and great care must be taken not to disturb their setting during this test.

An X-ray film is placed at the usual SSD and a lead marker placed exactly on the central axis – as indicated by the front pointer. An exposure sufficient to moderately blacken the film (density between 1 and 1.5) is made and the distance of the centre of the blackened area from the central axis (cross) is measured in a direction parallel to the major axis of the beam.

Figure 53 shows the geometrical aspect of this test. Let this distance be x_0 mm. The diaphragm system is then rotated through 180° and the exposure (on a new film) repeated. The distance (x_{180}) between the centre

of the blackened area and the central axis in a direction parallel to the major axis of the beam is again measured.

Ideally the source position should now be adjusted with respect to the central axis of rotation until the two distances (x_0 and x_{180}) are equal in magnitude and opposite in direction. The double exposure is necessary since there is no guarantee at this stage that the collimator jaws are symmetrical about the central axis. If the jaws are symmetrical, the blackened area will be in exactly the same position for the two exposures. The source position is adjusted until the centre of the blackened area coincides with the central axis. Figure 53 shows the situation existing for the two exposures when the jaws are not symmetrical and the source is not centred. Figure 54 shows the corresponding situation when the jaws are symmetrical. The whole test is now repeated in the direction of the minor axis.

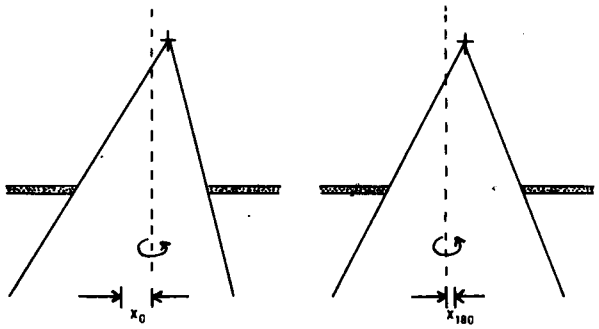

FIG. 53. Geometry of source-centring check by the film method: asymmetrical collimators; collimators rotated through 180°.

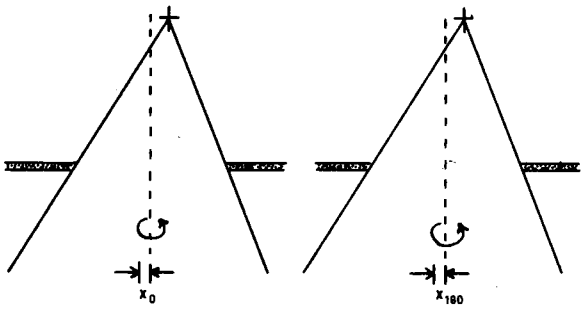

FIG. 54. Geometry of source-centring check by the film method: symmetrical collimators; collimators rotated through 180°.

Although in an ideal situation the source centre should be exactly on the central axis, it can be considered satisfactorily positioned if the algebraic sum of the two distances (x_0 and x_{180}) measured on the film placed at the usual treating distance is less than 3 mm. For example, if x_0 = 2 mm (the centre of the blackened area lying to the right of the central axis) and x_{180} = -4 mm (the centre of the blackened area lying to the left of the central axis), the algebraic sum is

$$2 + (-4) = -2 \text{ mm}$$

and this is acceptable. The consequence of a substantially non-centred source are an asymmetrical and non-uniform beam which does not point in the desired direction. If the source is known to be off centre, these aspects should be examined in the same way as described for the non-centred focal spot of the X-ray machine (section 6.2.1.5).

Unfortunately, it is sometimes difficult to decide on the exact location of the edge of the blackened area on the film, especially if the penumbra is large, so that this method is not very critical. A better, more critical method is to use an ionization chamber.

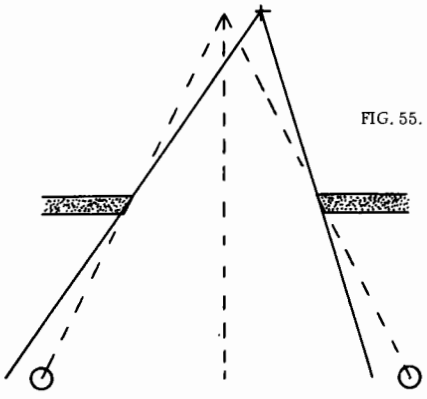

FIG. 55. Source-centring check by the ionization chamber method.

(b) By ionization chamber: An ionization chamber fitted with a build-up cap (section 3.2.4) is fixed rigidly to the diaphragm system so that it is inside the penumbra (Fig. 55). This position is found by taking the reading on the ionization chamber when it is fully inside the beam using a large beam (i.e. the diaphragm jaws are opened wide) and then closing the jaws until the reading on the dosemeter, for the same exposure time as before, is about one half of its former value. Several identical exposures are made and the average reading noted. The diaphragm system, with ionization chamber attached, is then very carefully rotated through 180°, care being taken that the jaw setting is not disturbed. Further series of readings are taken and the average noted. If the source is symmetrically positioned on the central axis, these two averages will be identical. If not, the source position must be adjusted until the readings are within 2-3% of each other. The test is then repeated in a direction at right angles to the first.

6.2.2.3. Collimator aperture: symmetry of jaws

A fixed pointer is set up so that it almost touches the face of one of the jaws. The collimator system is then rotated through 180° exactly, great care being taken not to disturb either the pointer or the collimator setting. If the jaws are symmetrical, the pointer will almost touch the face of the other jaw to which it is now close.

Alternatively, the symmetry of the collimator jaws is tested by film and in fact the films taken for the test of the source-centring may be used. If the jaws are correctly symmetrical, the degree of asymmetry will be the same on both films before the source is centred correctly and

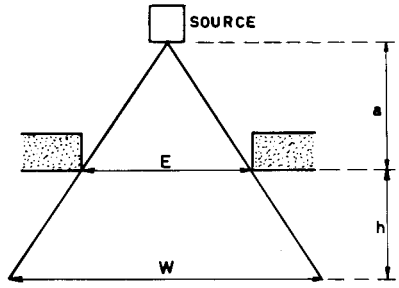

FIG. 56. Meaning of geometric beam size.

zero on both films after the source is centred. If it is not, the jaws must be adjusted until it is.

6.2.2.4. Correct spacing of jaws

It is usual to refer to telecobalt beam sizes using the convention of geometric beam size. The relationship between the aperture size (E) and the geometric beam size (W) (Fig. 56) is:

$$E = \frac{Wa}{a + h}$$

where a is the distance between the source and the aperture; and h is the distance between the aperture and the level at which the geometric beam size is specified (i. e. at the normal treating distance plus build-up depth).

The numerical indication on the collimator system should be checked by direct measurement of the jaw separation.

6.2.2.5. Beam uniformity

The beam uniformity may be tested in exactly the same way as for X-ray equipment. It is important to remember to fit a build-up cap (see section 3.2.4) to the ionization chamber.

6.2.2.6. Light-beam diaphragm

The correct positioning of the light source on the central axis, whether directly or via a mirror, can be tested by observing the symmetry of the light patch with respect to the central axis (front pointer). Incidentally, if the light source does not rotate with the collimator system, the symmetry of the jaws can be tested by observing that the light patch does not move laterally when the collimator system is rotated through 180°. This is so even if the light is not yet correctly centred and this test can be used instead of the one described above.

The size of the light patch at the standard SSD must agree with the numerical indication (of geometric beam size) on the collimator system. If it does not, then the distance of the light source must be adjusted until it does. Alternatively if the light area is too large, opaque strips (transparent to gamma rays however!) can be fitted to the jaws.

The use of the test device (section 6.2.1.6) described for X-ray equipment is also recommended to confirm the coincidence, proper position and sizes of the γ-ray beam and of the visible light-beam indication.

6.2.2.7. Interchangeable collimators

Some telecobalt and most telecaesium equipment is fitted with non-adjustable (i.e. fixed-beam size) collimators. These should be tested in the same way as for the treatment cones fitted to X-ray equipment (sections 6.2.1.1 and 6.2.1.4).

6.2.3. Source-surface distance

The precise value of the SSD is not usually important (see also section 5.2) but it is essential that it should be constant. A convenient check is to measure and record the distance from some fixed point on the X-ray tube housing, ^{137}Cs or ^{60}Co source head to the position of the treatment surface (as indicated by the front or optical pointer).

6.2.4. Isocentric mounting

Many telecobalt machines are mounted in what is known as an isocentric manner. This ensures that:

(a) the head of the unit can be rotated so that the central axis always passes through a fixed point in space; this point is known as the isocentre (Fig. 57);

(b) the treatment couch rotates about a vertical axis which passes through the isocentre and the up-and-down movement of the couch is parallel to this axis.

Optical light-beams are fitted to some equipment and these indicate the position of the isocentre.

It is essential to test that the unit is correctly installed so that all these features exist.

6.2.4.1. Identification of the isocentre

The position of the tip of the mechanical front pointer (or the tip of a pointer attached to the collimating system with its tip on the central axis) is observed with respect to a fixed pointer as the unit is rotated through 360°. The distance of the tip of the front pointer from the source is adjusted (taking care to keep the tip on the central axis) or the distance of the head of the unit from the isocentre is adjusted until the pointer's tip remains within a circle of diameter 2 mm as the unit is rotated through 360°. The centre of this circle is the isocentre. If this situation cannot be obtained by adjustment of the distance of the tip of the front pointer (or of the head), a lateral movement or angulation of the head of the unit is necessary.

Having identified the position of the isocentre in space, the tip of the front pointer is set so that it coincides with the isocentre.

6.2.4.2. Couch rotation and vertical movement

The movement of a pointer fastened to the couch with respect to the isocentre is observed as the couch is rotated through 360° about the isocentre. Any movement should be within a circle 2 mm in diameter. This test should be repeated with the couch at various heights from the floor to

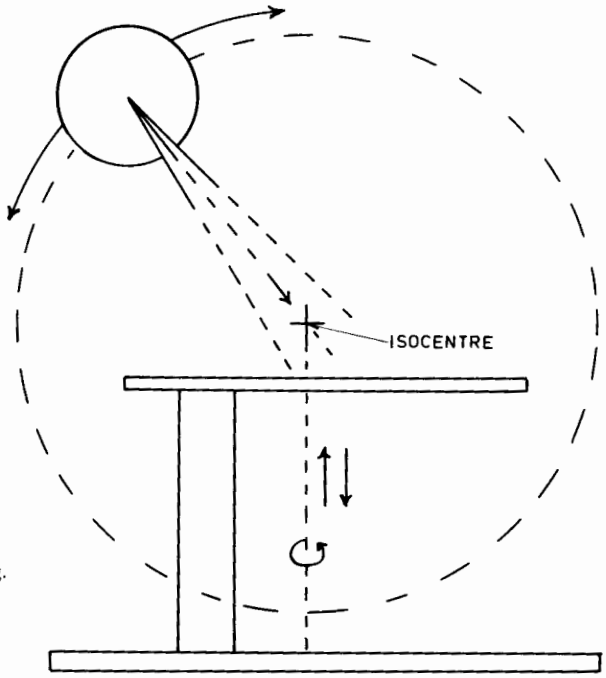

FIG. 57. Isocentric mounting.

confirm that the vertical movement of the couch is parallel to the axis of rotation of the couch.

6.2.4.3. Indicating lights

The correct indication of the isocentre by the side and ceiling lights should be checked if these are fitted.

6.2.4.4. Plane of central axis

A further requirement for the isocentric mounting and especially for rotation therapy is that the central axis of the beam and the axis of rotation of the treatment couch should be in the same plane and it is preferable, but not essential, that this plane is vertical. If the plane of coincidence is vertical, a plumb line can be used to identify it and confirm that the central axis (as indicated by the front pointer set at a range of distances, or by the optical beam) remains in this plane. Likewise a plumb line can be used to check that the couch movement is vertical.

Alternatively, the coincidence of the two planes can be checked by directing the optical light-beam from the head onto a fixed mark on the couch and observing that the central axis (centre of the light beam) continues to pass through this point for all heights of the couch and/or orientations of the head.

6.2.5. Orientation of the X-ray tube and telecobalt or telecaesium head

It is important to emphasize that the X-ray tube or telecobalt head must remain in the correct adjustment, no matter in what direction the

beam is pointing. In other words the tests described in this chapter should be repeated with the X- or γ-ray beam directed upwards, downwards, sideways and obliquely.

6.3. FREQUENCY OF TESTS

It is imperative that all the tests listed in this chapter be carried out initially, and their results and any adjustments that have been made carefully recorded. These tests should also be repeated at intervals. No firm recommendations can be made for the frequency of these tests. Items likely to change, such as the focal spot or source position, position of the light, the front and back pointers and the light and X-ray beam sizes should be tested frequently, say once per month, until confidence in their stability is established. The other items which are less likely to change should be tested at six-monthly intervals.

The appropriate items should, of course, be checked if the equipment is accidentally damaged and whenever any suspicion of a lack of adjustment is felt.

6.4. ACCURACY OF TESTS

In this chapter tests have been described and adjustments recommended which are intended to achieve a high standard of accuracy of beam position, beam size, beam direction and isocentricity. Regretfully it must be admitted that much of the available X- and γ-ray equipment can be made to meet these specifications only with the greatest difficulty and indeed some equipment may never achieve them. If circumstances leave no alternative to using such equipment, the knowledge obtained by conducting the recommended tests will, at least, enable the radiotherapist to know the circumstances in which his treatment beam is likely to be misplaced etc. and so avoid undesirable consequences.

It is clear that it is most important to examine very carefully the specifications of any equipment to be purchased and to enlist the co-operation of the manufacturer and his installation engineers in making sure that the equipment is left by them in good working order and adjustment.

CHAPTER 7

RADIATION PROTECTION SURVEY

This manual is not the place for an extensive discussion and description of the problems encountered in ensuring the radiation safety of persons liable to be exposed to ionizing radiation. The documents listed in Refs. [8-10], among others, give more detailed accounts and the reader is referred to them. It is, however, considered worthwhile to present an outline of the kind of testing which needs to precede any use of the equipment for radiotherapy. To be specific, various levels of permitted exposure and exposure rate are quoted in this chapter and these conform to the present recommendations of the International Commission on Radiological Protection. The user must, however, take great care to comply with any existing national, local or other regulations which apply to his equipment and which may contain values different from those quoted here for illustration.

There are three general classes of persons to consider:

(a) The patients, who need to be considered both during their treatment and whilst waiting nearby

(b) Members of the hospital staff, both those immediately and directly involved in the treatment and other auxiliary staff who work either full or part-time in nearby adjoining rooms

(c) Members of the general public and other members of the hospital staff who may be occasionally in the vicinity of the radiotherapy department.

It is convenient to consider these three groups of persons separately.

7.1. PROTECTION OF THE PATIENT

Although it has been decided that certain localized parts of the patient shall be irradiated to a high level of dose, it is important that other parts of his body should receive as little radiation dose as possible. Such unwanted doses may result from excessive amounts of radiation being transmitted through the tube housing at places other than where the intended treatment beam is localized (see Fig. 58).

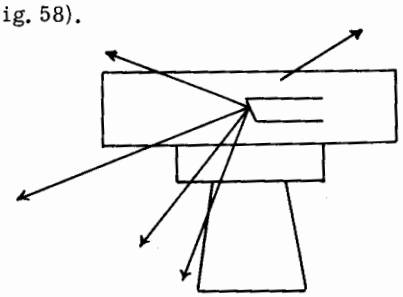

FIG. 58. X-ray tube shield leakage.

Most X- or γ-ray equipment is carefully constructed so that the leakage radiation is well below the permitted levels. It is imperative however, that this should be checked.

7.1.1. Test for leakage radiation

The existence of and position of any tiny holes or defects in the shield which allow the escape of small beams of radiation should be determined by surrounding the head of the teletherapy equipment with X-ray film and making a suitable exposure. For this exposure the main beam should be blocked off either by means of closing the shutter (if one is fitted) or by means of a thick piece of lead, 4 mm for kilovoltage X-rays, 10 cm for ^{60}Co γ-rays, inserted across the aperture. These films should be carefully marked so that afterwards it is possible to know their exact positions during the exposure.

The exposure should be made at the maximum operating kV and mA of the X-ray equipment and be of such duration that the dose in the main beam at the normal SSD is about 100 R. For ^{60}Co equipment, at which quality the film is much less sensitive, the exposure time needs to be twenty or thirty times greater.

After processing, the film should be inspected for local areas of blackening (high dose). If none are apparent or if the whole film is too black, a new set of films should be exposed using either a longer or shorter exposure time. The most appropriate exposure time must be found by trial and error. Care should be taken not to use too great an exposure since at very high exposures photographic reversal may occur and, therefore, very high intensity leaks missed.

The general exposure rate level around the tube and the exposure rate level associated with any defects in the shielding should now be measured using an ionization chamber of suitable size. Clearly if there are no defects and the general level of radiation is fairly uniform, a large volume chamber can be used. If, on the other hand, there are effectively small areas of high exposure rate, these should be measured using a chamber whose size is less than the size of the area of high exposure rate. If a sufficiently small chamber is not available, the reading obtained using a larger chamber may be adequately converted for the present purpose by multiplying it by the ratio:

$$\frac{\text{Cross-sectional area of chamber presented to beam}}{\text{Area of beam}}$$

The maximum level of permitted leakage radiation is:

(a) For deep X-ray therapy equipment working at its maximum operating kV and mA the leakage radiation at a distance of 1 m from the target should not exceed 1R in 1h and also should not, at a distance of 5 cm from the tube housing at any point accessible to patients, exceed 30 R in 1 h.

(b) For ^{60}Co therapy equipment the maximum rate of leakage radiation at a distance of 1 m from the source must be less than 1 R/h or 0.1% of the exposure rate in the useful beam at a distance of 1 m from the source, whichever is the greater.

This means, for example, that if the exposure rate in the useful beam is 100 R/min at 50 cm (i.e. 25 R/min at 1 m), the leakage during the treatment must not exceed

$$25 \times \frac{0.1}{100} \times 60$$

$$= 1.5 \text{ R/h at 1 m}$$

If the cobalt equipment is the moving source type, for this test the source must be at the ON position and the main beam blocked off by a suitably thick (about 10 cm) piece of lead or equivalent absorber.

(c) For low kilovoltage equipment which is to be held by the therapist during the treatment, the maximum level of leakage radiation must not exceed 0.1 R/h at a distance of 5 cm from the tube housing.

If any defects are found such that the associated leakage exposure rate exceeds the acceptable level, it is essential to have them repaired – preferably by the manufacturer.

7.1.2. Leakage through movable diaphragms and shutter

When the beam size is defined by means of adjustable or removable diaphragms or cones, it is generally agreed that, although they should be so constructed that the degree of protection they afford approaches that of the tube housing itself, a slightly higher level of radiation transmission is allowable. If possible this should be less than 2% and under no circumstances should it exceed 5% of the useful beam exposure rate (both measured at the same SSD). When testing for this transmission, care must be taken to avoid including in the measured radiation any scatter from the useful beam, preferably by blocking off the beam completely.

The leakage radiation through the shutter, when the X-ray machine is fully energized but prior to the treatment exposure, should be less than 1 R/h at 1 m from the source if possible. However, since it is often impracticable to modify an existing piece of equipment, some relaxation is permissible and a transmission level up to but not exceeding 5 R/h at 1 m can be accepted.

An unnecessarily large penumbra due to badly designed or badly constructed treatment cones or maladjusted diaphragm systems can result in undesirable extra exposure of the patient. Information is given in sections 6.2.1 and 6.2.2 about the testing of treatment cones and diaphragm systems.

7.2. PROTECTION OF THE STAFF

It is necessary to consider the dose to which the staff may be subjected whilst inside the treatment room preparing the patient for the treatment, whilst outside the room during the treatment and whilst in the vicinity of the X-ray room generally.

7.2.1. Dose received by staff whilst inside the treatment room

For deep (i.e. above 150 kV) X-ray therapy installations the control desk at which the exposure is switched ON and OFF must be outside the treatment room and during the treatment the technician must remain outside the room. Whilst the treatment technician is inside the room the X-ray equipment should not be energized, that is to say the kilovoltage should be switched off completely. The door of the treatment room should be fitted with an electrical interlock system so that (a) the kilovoltage is automatically switched off whenever the door is opened; and (b) the voltage can be re-applied to the X-ray tube only from the control desk and after the door has been closed. There should also be an automatic, visible

indication (e. g. a red light) both inside and outside the room to show that the kilovoltage is applied to the tube.

If, in exceptional circumstances, it is not practicable to switch off the kV completely whenever the room door is opened, the kV must then be automatically reduced to as low a value as possible and, in any case, by at least 25% and the shutter closed. In this latter case great care must be taken that the correct and proper operation of the shutter is checked frequently and regularly. The door of the treatment room should be fitted with an electrical interlock system so that (a) the shutter closes automatically whenever the room door is opened; and (b) the shutter can be reopened only from the control desk after the door has been closed. There should be an automatically operated, visible indication (e. g. a red light) both inside and outside to show when the shutter is not closed.

The correct and proper functioning of these interlock and indicating systems must be checked frequently and regularly.

Clearly, switching off the kilovoltage renders the room perfectly safe and it is the recommended procedure between exposures. If, for any reason, this is not practicable, the maximum permitted levels for leakage radiation are then:

(1) Maximum exposure rate at 1 m from the target, 10 mR/h
(2) Average exposure rate at 1 m from the target to be less than 2 mR/h
(3) Maximum exposure rate at 5 cm from the tube housing, 100 mR/h
(4) Average exposure rate at 5 cm from the tube housing, 20 mR/h.

These levels of dose rate apply also to telecobalt equipment when in the OFF position (i. e. the useful beam is switched off). The method of measuring the leakage radiation is the same as that given in section 7. 1. 1 for the measurement of the leakage radiation from equipment which is switched on.

For X-ray equipment operating at less than 150 kV, and in particular for short focus-skin distance (less than about 5 cm) and Grenz-ray type treatments, it is permissible for the operator to remain in the room during the exposure and for the control desk to be inside the room. It remains essential that he should be behind a protective barrier or should wear suitable protective clothing while the equipment is in operation and that his personal exposure should be monitored.

In the case of hand-held equipment the radiologist must wear protective (lead rubber) gloves in addition to wearing a protective apron.

7. 2. 2. Dose received whilst outside the treatment room

As stated above, members of the staff should not be inside the treatment room while the beam is switched on. Care must be taken, however, that there is no excessive leakage of primary and scattered radiation from the treatment room to:

(a) The region of the control desk
(b) Other rooms adjoining the treatment room.

It is particularly important to remember that any rooms on the floors above or below that of the treatment room are at risk as well as those immediately to the side of it. The places where the patients are likely to sit whilst they are waiting for their treatment or otherwise attending the department should clearly be included in this survey.

(c) The region outside the building. This is particularly important if there are any windows or roof lights in the treatment room. A survey, using a large-volume ionization chamber or a small Geiger-Müller counter should be made in all these places for all orientations of the teletherapy equipment and with the beam switched on. For this test the beam should be directed into a phantom and the equipment operated at the maximum beam size, tube voltage and current. Particular attention should be given to the edges of any viewing windows (lead glass), doors of the treatment rooms and also to ventilation grills and electrical conduits. If any gaps in the protection construction are suspected, they can be examined by the use of films in the same way as for the tube housing (see section 7.1.1).

During this survey the teletherapy beam should be pointed in all directions which may be used. It may be that there are some areas of the walls that are intended only as barriers against scattered and leakage radiation (so-called 'secondary barriers'). Clearly, the primary beam must never be pointed at these areas of the walls. If possible, mechanical or electrical restraints should be fitted to the equipment to ensure this, otherwise clear and unambiguous instructions about the permitted directions must be given. The areas of the room wall towards which the primary beam may be directed are known as 'primary barriers' and these must be fully tested using the maximum beam size, kV and mA.

It is difficult to state what level of exposure rate may be considered acceptable for locations outside the treatment room that are regularly occupied by members of staff. The success of the room protection depends on the degree of utilization of the equipment and on the occupancy of the protected locations. Reference [9] gives details of how to design the protective barriers. As an indication of what is likely to be satisfactory, it can be stated that the exposure rate in any regularly occupied region should not exceed 0.5 mR/h for the worst possible operating conditions of the equipment. The adequacy of the room protection is judged finally by the results of the personnel monitoring of the staff (7.2.3).

Any regions which are not permanently occupied such as paths, gardens, roofs etc. to which access by the general public is possible, can be allowed a rather higher level of exposure rate. Even here every effort should be made to keep the level as low as possible and it should not normally exceed an integrated exposure of 40 mR in any period of one month. This can be tested by mounting film badges or other dosemeters in suitable fixed positions in these regions.

7.2.3. Personnel monitoring

All members of the staff working with ionizing radiations should wear a film badge or other suitable type of monitoring device (e.g. quartz-fibre electrometer chamber, lithium fluoride capsule) so that their accumulated radiation dose can be estimated and recorded.[6] The publications listed in Refs [8-10] give the recommended values of maximum permissible doses in the various critical body organs. It must be firmly emphasized that there is no reason why these maximum values need ever be even approached and one must not be satisfied until all members of staff are receiving as

[6] The system used should be independent of radiation energy, or the quality should be known and account taken of the variation in response with quality.

low a dose as practicable and certainly one substantially less than 50 mR/week.

It is clearly unrealistic to issue film badges to all members of the hospital staff but all those who are directly involved (i. e. treatment technicians, medical staff, physicists, assistants, porters, nurses etc.) working in the radiotherapy department should have film badges.

It is also useful to place film badges in adjoining rooms in which other staff who do not wear film badges normally work. Every effort should be made to ensure that the exposure so measured is less than an integrated exposure of 40 mR in any period of one month. In this way confirmation of the safety of these locations can be established and once this is done there is no further need to continue to use film badges in this way, provided the general use of the teletherapy equipment is not changed.

7. 3. TELECAESIUM AND TELECOBALT EQUIPMENT: RADIOACTIVE CONTAMINATION

Although the manufacturer makes every effort to supply source capsules that are not externally contaminated and that will subsequently contain all the radioactive material within the sealed capsule, there is always a possibility of leakage of radioactive material, especially if the capsule suffers any mechanical damage. Such leakage would create a radiological hazard. It is therefore necessary to test that no leakage has occurred or is occurring. The regions of the telecobalt or telecaesium unit immediately adjacent to the source should be wiped over with a piece of soft foam plastic moistened with a detergent. This foam plastic should then be tested on a suitable scintillation or Geiger counting system. If the level of count rate is significantly higher than background, contamination must be suspected and the situation carefully investigated further, preferably by enlisting the aid of a suitably experienced colleague.

An alternative, simple method of testing the moistened foam plastic [7] is to place it directly on a piece of X-ray film and leave it in a darkened room or enclosure for about a week. After processing, the film will show whether there is any radioactive material on the foam plastic. Since much of the blackening is caused by soft β-particles, the film must not be inside a paper envelope or a cassette.

As for all tests, this wipe test should be repeated regularly.

7. 4. EMERGENCIES

All members of staff must be instructed in the proper procedures to be followed in case of an accident or emergency. There must be clear operating instructions posted in the vicinity of each item of equipment (section 8. 4) and the telephone numbers of responsible persons should be readily available.

[7] Some detergents can act chemically on photographic emulsion and so create blackening. A test should therefore be first made that the detergent used does not lead to such spurious blackening. If a detergent which does not blacken film cannot be obtained, a very thin sheet of plastic may be laid over the film. Some β-particles will be stopped by the plastic and the blackening due to radioactive contamination thus reduced.

CHAPTER 8

RECORD KEEPING

It is essential that good, permanent records be kept of all measurements and tests made on X- or γ-ray equipment and on the dosemeters. Details of any repair, modification or calibration should also be recorded. It is strongly recommended that these records be kept in bound, not loose-leaf, books. The records should be kept in a strict chronological order and the date (day, month and year) of each test and measurement recorded unambiguously. It is convenient to separate these records into four groups.

8.1. RECORD BOOKS

8.1.1. Dosemeter checks

A separate record book should be kept for each dosemeter. This record will contain information about the calibration of the dosemeter, together with all the results of the routine checks made on the dosemeter (see Chapter 3 for details of these checks). It should also contain a clear statement of the currently applicable values of the instrument calibration factor K_E appropriate for this dosemeter for each operating condition (i.e. quality) of the therapy equipment on which it is to be used. If and when it becomes necessary to change any value of K_E, the reasons for the change and the date from which it applies should be clearly recorded and the list of currently applicable values of K_E modified accordingly.

8.1.2. Initial checks and measurements on the teletherapy equipment

In this record book full details of all the initial tests, checks and measurements made on the therapy equipment must be recorded with sufficient detail that the test checks or measurements can be repeated exactly, if so required. Any further measurements (say for different operating conditions) made subsequently should also be recorded in this book.

8.1.3. Regular checks on the teletherapy equipment

This record book contains a chronological record of the results of the routine tests and checks (other than output measurements made on the equipment). See Chapter 6 for details of the tests.

8.1.4. Regular output checks

This record book contains results of the output check measurements that are made daily, weekly, etc. It must also contain a record of the operating conditions (i.e. kV, mA, etc.) and, if any of these are changed, the new values and dates from which they are applicable. An example of the kind of record intended is shown in Table XV.

In this record the 'instrument reading' is the reading obtained in the standard test jig (see Fig. 31) and for a standard exposure time.

TABLE XV. RECORD OF OUTPUT CHECKS

	Standard reading = 72.0		No. 3 X-ray machine		2.0 mm Cu HVT	
Date	kV	mA	Temp. (°C)	Pressure (mm Hg)	Instrument reading	Corrected reading
11.7.69	250	12.5	25	762	67.3, 68.0, 67.7	72.0
14.7.69	250	12.5	26	756	71.6, 71.1, 71.5	74.5
			Reading too high : reduce mA and re-check			
14.7.69	250	12.0	26	756	68.6, 68.2, 68.4	71.4
16.7.69	250	12.0	25	764	69.8, 69.4, 69.6	71.8
18.7.69	%DD checked. See page 47 of Routine Check Book.					
21.7.69	250	12.0	25	758	69.5, 69.5, 69.2	72.0

'Corrected reading' is the 'instrument reading' corrected for temperature and pressure, instrument calibration factor (K_E = 1.025, in this example) and timer end errors (assumed zero in this example). 'Standard reading' is the value of the 'Corrected reading' obtained in the test jig on the occasion of the 'output' measurements. The aim is to keep the 'Corrected reading' within 1-2% of the 'Standard reading'.

8.2. CHECK LISTS

It is worth preparing check lists for the various tests and measurements which need to be done, both initially and regularly. From these lists it is possible to see at a glance which tests still need to be done. In this way the worker can be certain that nothing of importance has been inadvertently omitted.

Appendix IX gives examples of three kinds of list. The first is concerned with the initial mechanical testing; the second with the regular tests to be done on a dosemeter and the third with the regular check of output on one particular piece of X-ray equipment. These lists are given only as examples and illustrations of what is intended. All three lists will probably need to be extended to suit the user's particular needs. It must be the responsibility of the user to compile his own check lists and the ones given in the appendix must not be uncritically accepted as being complete.

8.3. TREATMENT RECORDS

Although it is not relevant to the details of the present manual, it is also necessary to keep accurate records of the full operating conditions used for each treatment to the patient. The kind of details required are:

X-ray equipment used
Name of the person supervising the treatment
kV, mA, filter, HVT
Field size and SSD
Wedge filter
% DD, or TAR
Value of the output
Exposure time or monitor divisions
Given exposure or dose on each beam
Date of treatment
Tumour dose, skin dose, etc.

It is useful to connect a continuously reading pen-recorder to each teletherapy machine so that a record is made whenever the X-ray beam is switched on. If the teletherapy machine is fitted with a shutter, the pen should record whenever the shutter is not closed. If the equipment does not have a shutter, the pen should record whenever kV is applied. The record will therefore show the duration and time of day for each and every exposure. This record can be examined each day in conjunction with a chronological list of the day's work to confirm that (1) all patients have been treated; (2) all fields on each patient have been treated; and (3) the exposure times were correct.

8.4. OPERATING INSTRUCTIONS

There must be a written set of operating instructions for each teletherapy machine. These instructions must contain full details of the safe and correct operation and use of the equipment. They should also refer to any specific radiological protection requirements.

Every person using the equipment must be fully conversant with these instructions, a copy of which should be kept with the apparatus and an additional reference copy kept safely elsewhere.

CHAPTER 9

THE NEXT PROBLEMS

For the purpose of this manual the patient has been assumed to be equivalent to a large volume of homogeneous material (water or soft tissue) which has a flat surface. It is also assumed that the central axis of the beam is at right angles to the surface (Fig. 59).

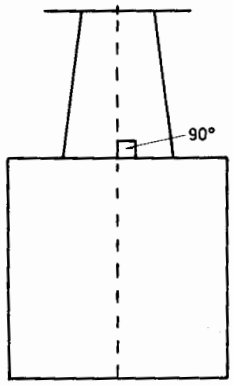

FIG. 59. Beam directed normally into homogeneous medium.

Using the techniques and methods described and recommended in this manual will enable the radiotherapist and physicist to be certain that the dose delivered to any point within this idealized patient can be stated consistently. By this is meant that a planned dose can be delivered to the point, that the true dose delivered is consistent from day to day and from year to year in any particular radiotherapy centre and also that there can be mutual consistency of dosage delivery and statement in all radiotherapy centres.

Although real patients do not conform, of course, to their idealized form, the achievement of at least this degree of consistency is essential to good radiotherapy and facilitates the interchange of experience between centres.

This is not, however, the end of the work and the contents of this manual must be regarded only as being the necessary basis for other, equally important, steps in solving the complete problem. A brief and certainly not comprehensive indication of some of these other aspects of the work is set out below. Superficially, they are more interesting and may appear to the inexperienced more important than the aspects discussed in this manual. This is not so. It cannot be overemphasized that the work described in this manual must precede that set out below, which should be attempted only by those who have at least some degree of experience.

There are, of course, additional technical aspects of radiotherapy treatments which are not discussed in this manual. For example, it may be necessary to design beam-flattening filters, wedge filters or other beam-modifying devices.

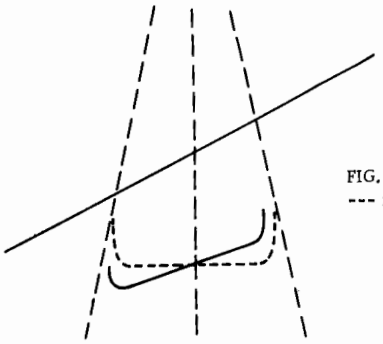

FIG. 60. Effect of 'oblique incidence'; position of isodose curve for: --- normal incidence; —— oblique incidence.

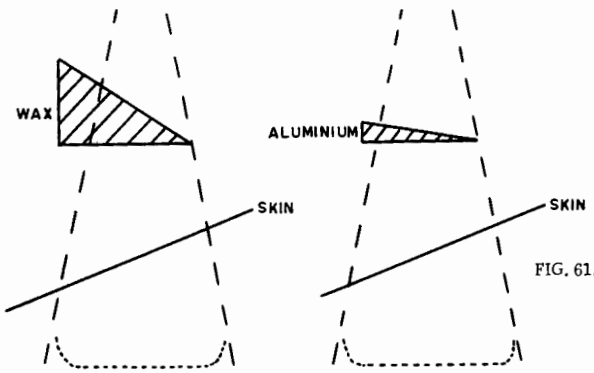

FIG. 61. Compensation for 'oblique incidence'.

The problem raised by the fact that the surface of the patient through which the radiation beam is directed is generally neither flat nor at right angles to the beam central axis can be dealt with in two ways. The effect of this 'oblique incidence' (Fig. 60) on the dose pattern can be accepted and the changes in the dose pattern calculated. Alternatively a 'compensator' can be used (Fig. 61) to counteract the effect of oblique incidence so that the dose pattern within the patient is the same as that in the idealized patient to which the standard isodose chart applies. The presence of the compensator, of course, reduces the output and this must be allowed for.

The presence of bone and, particularly, lung tissue within the irradiated beam can also change significantly both the pattern and magnitude of the dose. Again the effects can be accepted and the extent of the associated changes in dosage calculated or a compensator can be used to restore the radiation pattern to its standard forms in the tissue beyond the lung or bone. Probably the major point of this problem is that of obtaining adequate information about the size, position and character of the lung or bone, within the individual patient. Specialized radiography (e. g. tomography, especially transverse axial tomography) and transit dose measurements are helpful in this respect.

In some circumstances direct measurement of the radiation dose on or within the patient can be carried out. This requires great skill on the part of the physicist, especially since it is rarely possible to do multiple readings. Furthermore the interpretation of the results obtained needs to be done with great care, one of the usual problems being that of knowing exactly where the dosemeter is located in the patient and with respect to

the beam during the irradiation. For this kind of measurement the ionization chamber is often not convenient and thermoluminescent (lithium fluoride) or phosphate glass systems may be used.

It is the nature of the subject that it is impossible to identify all the problems and it must be left to the worker, as he develops his experience, to recognize the problems existing in his own department and to deal with them. It is comforting to realize that probably one of the more important roles which the physicist has is to recognize and identify the problems: the solutions are often, although not always, simple.

APPENDIX I

REFERENCES AND BIBLIOGRAPHY

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APPENDIX II

SPECIAL TERMS USED IN RADIOTHERAPY

ABSORBED DOSE (D): The absorbed dose of any ionizing radiation is the energy imparted to matter by the ionizing radiation per unit mass of the irradiated material at the point of interest (see also 'rad').

BUILD-UP: In a material irradiated by a beam of X- or gamma-rays the increase in absorbed dose with depth below the surface is called the build-up. This is due to (a) the forward moving nature of the secondary electrons produced in the material, as well as (b) a build-up of scattered photons due to multiple scattering in broad beams of radiation. For high-energy beams process (a) is the more important.

CENTRAL RAY: The central ray is the straight line passing through the centre of the source and the centre of the final beam-limiting diaphragm. (The latter means the centre of symmetry of the plane figure formed by that edge of the diaphragm system which defines the beam.)

DOSE DISTRIBUTION: The dose distribution is a representation of the variation of dose with position in any region of an irradiated object.

EXPOSURE (X): The exposure (X) is defined as: $X = \Delta Q / \Delta m$ where ΔQ is the sum of the electrical charge on all the ions of one sign produced in air when all the electrons liberated by photons in a volume element of air whose mass is Δm are completely stopped in air (see also 'röntgen').

GIVEN (OR APPLIED) DOSE: The given dose is the surface dose (for radiation below 400 kV) or the peak dose (for harder qualities) delivered by one beam in a complete treatment, or in a treatment session.

GEOMETRICAL EDGES OF THE BEAM: The geometrical edges of the beam are the lines joining the centre of the anterior face of the source to the diaphragm edges furthest from the source.

GEOMETRICAL FIELD (BEAM) SIZE: The geometrical field size is the geometrical projection, on a plane perpendicular to the central ray, of the distal end of the limiting diaphragm as seen from the centre of the front surface of the source. The field has thus the same shape as the aperture of the collimator.

The geometrical field size can be defined at any distance from the source. Positions of special interest are (a) at the skin surface; (b) at a distance corresponding to the centre of the target volume; or (c) at moving-field therapy.

The geometrical field as defined here will be similar in size and shape but not identical with the 'physical field' which some workers define as being outlined by the 50% isodose curve. Both the geometrical and 50% physical fields will be larger than fields defined in terms of the 80 or 90% isodose curves, and those who have been accustomed to these conventions should take special note of this, as should those who have previously been accustomed to X-rays below 400 kV where the surface dose at the geometric field edge is seldom below 85% of the central dose.

HALF VALUE THICKNESS (HVT): The half value thickness is the thickness of a sheet of the specified material which attenuates the beam to such an extent that the exposure rate is reduced to one half, under narrow beam conditions. For X-rays generated at voltages between 50 and 150 kV the HVT is usually stated in mm of aluminium, between 150 and 400 kV

copper is used, while for more penetrating radiations the HVT is usually stated in mm of lead.

ISODOSE CHART: An isodose chart is a set of isodose curves, usually drawn for regular intervals of absorbed dose or of percentage depth dose, which represents the distribution of dose over a plane surface within the irradiated body.

ISODOSE CURVE (OR CONTOUR): An isodose curve is a line along which the absorbed dose is constant. (N.B. For X-rays up to 400 kV the isodose surfaces or curves may alternatively be drawn as surfaces or curves of constant exposure.)

ISODOSE SURFACE: An isodose surface is a surface on which the absorbed dose is constant.

OUTPUT: The term 'output' has been defined by the ICRU (Handbook 87) as either the exposure rate (R/min) or dose rate (rad/min) at a specified point under specified, standard conditions. In this manual the term 'output' is used only in an undefined sense and refers to the exposure rate or dose rate of a piece of equipment in general. Other terms will be used to describe the 'output' for various specified conditions.

PEAK ABSORBED DOSE: The peak absorbed dose is the maximum value of the absorbed dose which occurs along the central ray.

The peak absorbed dose in water is situated at a depth of about 0.2 cm for ^{137}Cs, 0.5 cm for ^{60}Co, 1 cm for 4 MV X-radiation and 4 cm for 25 MV X-radiation. The exact position of the peak depends on the type of collimator and on whether a secondary electron filter is used. It also depends slightly on the SSD and the field size. (N.B. In an asymmetrical beam such as that produced by a wedge filter, the peak absorbed dose may be less than the maximum absorbed dose, which occurs off the central ray.)

PERCENTAGE DEPTH DOSE: The percentage depth dose in an irradiated body is the ratio (expressed as a percentage) of the absorbed dose, D_x, at any depth x to the absorbed dose, D_0, at a fixed reference point on the central ray.

$$\text{Percentage depth dose} = 100 \times \frac{D_x}{D_0}$$

For X-radiation produced at up to 400 kV the reference point is at the surface. For X-radiation above 400 kV and gamma teletherapy the reference point is at the position of the peak absorbed dose. For moving-field therapy it is often convenient to take the centre of rotation as the reference point.

PHANTOM: A phantom is a volume of tissue-equivalent material either large enough to provide adequate scatter or constructed to resemble some special object, such as part of the human body, for the purpose of measuring a dose distribution.

RAD: This is the special unit of absorbed dose and corresponds to 100 erg per gram or 0.01 joule per kilogram of absorbing matter.

RADIATION QUALITY: The term quality as applied to ionizing radiation attempts to characterize its power of penetrating matter. This power depends on the energy distribution of the photons which make up the beam. Many methods have been used to describe quality but for general clinical use the following methods of quality specification are recommended.

(a) For X-rays up to 2 MV state the kV or MV and the half value thickness (HVT)

(b) For X-rays above 2 MV state the MV only

(c) For γ-rays specify the type of nuclide (e.g. ^{60}Co γ-rays).

RÖNTGEN (R): This is the special unit of exposure and corresponds to 1 esu of charge per 0.001293 gram of air or 2.58×10^{-4} coulomb/kilogram of dry air.

SOURCE-SURFACE DISTANCE (SSD): The source-surface distance is the distance measured along the central ray from the front surface of the source to the surface of the irradiated object. (N.B. The above definition can apply to X-ray as well as to gamma-ray sources.) The use of the older terms 'focus-surface distance' 'focus-skin distance' and 'source-skin distance' is to be discouraged.

SURFACE ABSORBED DOSE: The surface absorbed dose is the absorbed dose delivered by a radiation beam at the point where the central ray passes through the superficial layer of the phantom or patient (N.B. In an asymmetrical beam of radiation such as that produced by a wedge filter the surface absorbed dose thus defined may not be the maximum absorbed dose at the surface.)

SURFACE BACKSCATTER FACTOR: For orthovoltage X-rays the surface backscatter factor is the ratio of the total exposure at the surface (i.e. primary plus back-scattered radiation) to the exposure 'in air' (i.e. primary plus any scatter from the treatment cone) at the same position and for the same operating conditions of the equipment.

TISSUE-AIR RATIO: The tissue-air ratio is the ratio of the absorbed dose at a given point in tissue to the absorbed dose that would be measured at the same point in free air within a volume of the tissue material just large enough to provide the maximum electronic build-up at the point of measurement.

TISSUE-EQUIVALENT MATERIAL: A tissue-equivalent material is a liquid or solid whose absorbing and scattering properties for a given radiation simulate as closely as possible those of a given biological material, such as fat, bone or muscle. For muscle or soft tissue, water is usually the best tissue-equivalent material.

TREATMENT SESSION: FRACTIONATION: A session is a treatment or group of treatments delivered in one visit. Fractionation is the splitting of a dose into a number of short sessions given over a longer period than would be required if the dose were given continuously in one session at the same dose rate. A fraction is a single session in a fractionated treatment. Overall time is the total time elapsing from the beginning to the end of a session or of a series of sessions if the treatment is fractionated.

WEDGE FILTER: A filter of graduated thickness which causes a progressive decrease in intensity across the (whole or part of the) beam.

WEDGE ISODOSE ANGLE: The wedge isodose angle is the complement of the angle which the 50% isodose curve for the wedge beam makes with the central ray, in a principal plane of the beam. The 100% level is taken at the depth of the peak absorbed dose on the central axis of the beam.

APPENDIX III

SYMBOLS USED IN THIS MANUAL

%DD	Percentage depth dose
D*5	%DD at 5 cm deep
D	Absorbed dose
X	Exposure
δ	Ionization chamber displacement factor
p	Ambient pressure
P	Standard pressure
t	Ambient temperature (°C)
T	Standard temperature (°C)
$\phi(p, t)$	Pressure and temperature correction factor
m	Exposure time (minutes)
m'	Number of monitor divisions
m_1, m_2	'Shutter' operating times
ϵ	Shutter error time
I	Instrument reading
K_E	Instrument calibration factor (rôntgen per scale division)
f_λ	Rad per rôntgen conversion factor
SSD	Source-surface distance
SAD	Source-axis distance
W	Beam size at the surface
S	Focal spot size (apparent), or source size
a	Focal spot (or source)-aperture distance
h	Aperture-surface distance
A	Pin-hole-film distance
H	Focal spot (or source)-pin-hole distance
HVT	Half value thickness
TAR	Tissue-air ratio
R	Rôntgen
x or d	Depth in, or thickness of, overlying tissue or water
mA	Milliampere
kV	Kilovolt

MV	Megavolt
X_s	Peak (surface)exposure (röntgen)
$\dot{X}_s$	Peak (surface) exposure rate (R/min)
X'_s	Peak (surface) exposure factor (röntgen per monitor division)
D_s	Peak (surface) absorbed dose (rad)
$\dot{D}_s$	Peak (surface) absorbed dose rate (rad/min)
D'_s	Peak (surface) absorbed dose factor (rad per monitor division)
B	Surface backscatter factor

APPENDIX IV

EXAMPLES OF TREATMENT PLANNING

EXAMPLE 1: Figure A1 represents an arrangement of two, so-called parallel opposed, beams of 250 kV radiation (HVT 2.0 mm Cu, SSD 50 cm) which are being used to treat, for example, a neck with the intention of irradiating the whole region to a reasonably uniform dose. The two beams of radiation are not, of course, delivered simultaneously but alternately from the left and then the right. The distance, RL, between the two positions of the ends of the applicators is referred to as the 'Interfield distance' (IFD) and in this example it is taken to be 12 centimetres.

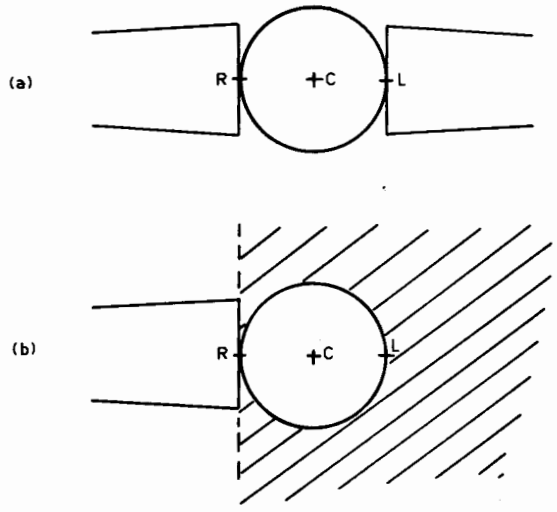


FIG. A1. Treatment by a pair of parallel opposed beams.

It is assumed that, as shown in Fig. A1(b), the spaces between the applicator base and the curved skin surface of the patient and the regions to the side and the back of the neck are filled with a suitable bolus material [11, 12]. The standard percentage depth dose data [3] are then applicable.

The depths of the three points R, C (centre) and L, measured from the entry points (R and L) of the right and left lateral beams respectively, are shown in lines 1 and 3 of Table A I and the corresponding %DD taken from the tables (p. 21) of Ref. [3] at these points and for the beam size (15 × 10 cm) used are given in lines 2 and 4, assuming that the entire region is composed of soft tissue only. The total percentage depth dose, obtained by adding the values in lines 2 and 4, is given in line 5. It can be seen therefore that if the given dose on each beam is 100 units, the dose to the irradiated tissue at points on the beam central axis lies between 116.6 and 126.6 units. The tumour dose (or mean dose) is chosen as 120 units. Hence if a tumour dose of 3750 rad is prescribed, the required given dose on each beam is calculated by simple proportion.

TABLE AI. %DD FOR A PARALLEL PAIR TREATMENT
HVT = 2.0 mm Cu; beam size, 15 × 10 cm.

Beam	Given dose	Point		
		R	C	L
1. Depth		0	6	12 cm
Right	100			
2. %DD		100	58.3	26.6
3. Depth		12	6	0 cm
Left	100			
4. %DD		26.6	58.3	100
5. Total %DD		126.6	116.6	126.6

Tumour dose = 120%
= 3750 rad in soft tissue
Hence given dose = 100%
$= 3750 \times \frac{100}{120}$
i.e. given dose = 3125 rad

The total dose at the surface (126.6%) and at the mid-point (116.6%) are also calculated by similar proportion and found to be

$$\text{Surface dose} = 3750 \times \frac{126.6}{120} = 3950 \text{ rad in soft tissue}$$

$$\text{Mid-point dose} = 3750 \times \frac{116.6}{120} = 3650 \text{ rad in soft tissue}$$

These are the maximum and minimum doses respectively. If this treatment is to be delivered over three weeks by means of 15 equal daily sessions, the given dose for each beam on each treatment day will be 3125/15 = 208 rad in soft tissue.

EXAMPLE 2: Figure A2 is a cross-sectional drawing of a patient who is to be treated by means of three beams positioned and directed as indicated. This arrangement is chosen purely for purpose of illustration but is the kind of arrangement that could be used for the treatment of, say, a mid-brain. In this example it is assumed that a 300 kV X-ray machine, working at an HVT of 2.5 mm Cu and an SSD of 50 cm, is being used. The problem is to determine the given doses which must be delivered to the 100% points of each beam (A, L and R) in order to have a uniform dose in the vicinity of the tumour, centred at T, and a dose level of, say, 5000 rad in soft tissue delivered in 5 weeks. Since the right and left posterior beams are symmetrical, it is to be expected that their given doses will be identical.

Using isodose charts appropriate to the field size and SSD selected, the percentage depth dose from each beam separately can be read off at each of the points (equidistant from the tumour centre) shown in the figure. Typical values of %DD are shown in lines 1, 2 and 3 of Table A II. The total relative doses resulting from the summation of the contribution from the two posterior fields are shown on line 4 and are obtained by adding together the values of lines 1 and 2.

TABLE A II. % DD FOR A THREE-FIELD TREATMENT

	Beam	Given dose	Point x	y	t	a	b
1.	Left posterior	100	29	42	39	52	27
2.	Right posterior	100	29	42	39	27	52
3.	Anterior	100	86	42	63	56	56
4.	Left plus right posterior	100 + 100	58	84	78	79	79
5.	Anterior	$100 \times \frac{26}{44} = 59$	51	25	37	33	33
6.	4 + 5		109	109	115	112	112

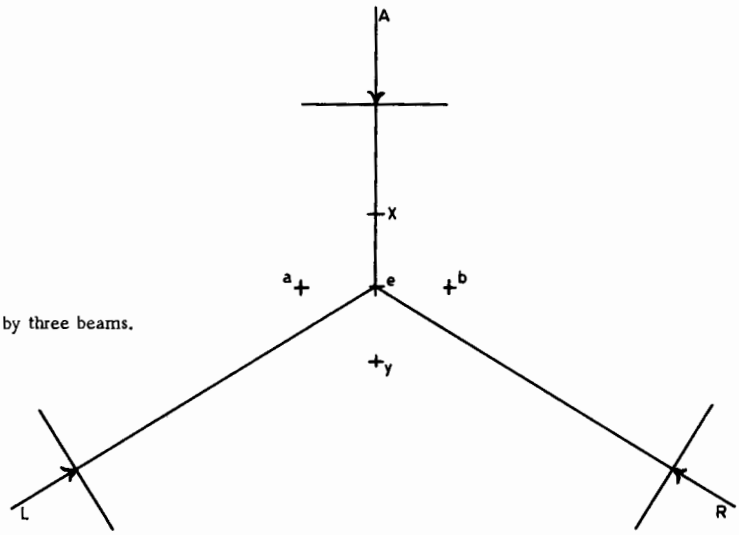


FIG. A2. Treatment by three beams.

To obtain a uniform dose over the whole region of the tumour, the total summed contributions at the points x and y must be made equal. The total contributions from the combined posterior fields are 58 and 84 units respectively (line 4), the relative dose at x being less than that at y by 84 - 58 = 26 units. The contributions to the two points x and y from the anterior field are 86 and 42 units, respectively (line 3) the relative dose at x being greater than that at y by 86 - 42 = 44 units. To ensure a uniform distribution of dose around the tumour these two differences must be made equal. This is done by reducing the given dose on the anterior field from 100 units to $100 \times 26/44 = 59$ units which will also result in a corresponding reduced relative dose at all points. The new values of dose obtained by multiplying those given in line 3 by the factor 0.59 are shown in line 5. The total relative doses at the various points which will exist if the given doses are 100 on each of the posterior fields and 59 on the anterior field are obtained by adding the values in lines 4 and 5, with the result shown in

line 6. It can be seen that these relative values of given dose produce a reasonably uniform pattern of radiation dose since all the numbers are identical within ± 3%. In deciding on the mean dose to the tumour region it is important to recognize that a much greater fraction of the volume is receiving the slightly lower dose represented by the values for the points x, y, a and b, whereas only a small fraction is receiving the maximum dose represented by the value for the point t. Due cognizance of this is taken if the mean tumour dose is chosen as being equal to the mean of the dose at these five points. In this example this is 111%. If the tumour dose is chosen to be 5000 rad then, by simple proportion

$$\text{Given dose on each posterior field} = 5000 \times \frac{100}{111}$$
$$= 4560 \text{ rad}$$
$$\text{Given dose on anterior field} = 5000 \times \frac{59}{111}$$
$$= 2660 \text{ rad.}$$

This method of, so-called, beam-balancing can be applied equally to other configurations and numbers of beams. For further examples the reader is referred to Paterson (see Bibliography).

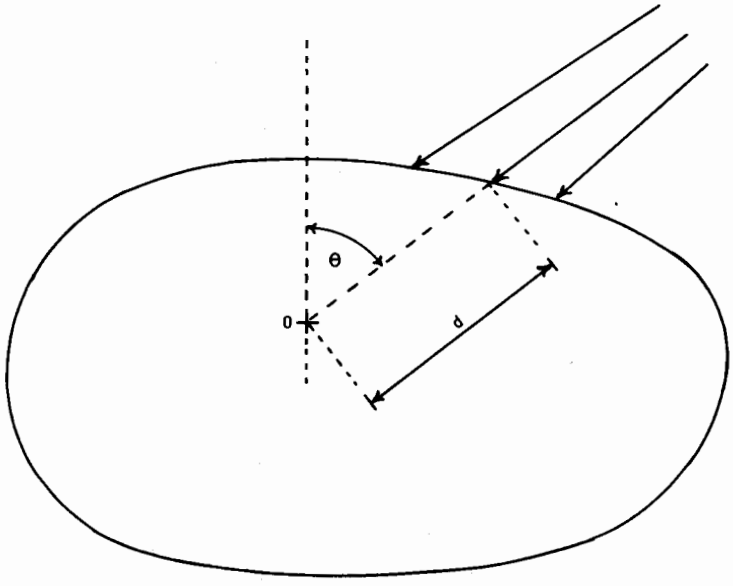


FIG. A3. Moving beam treatment.

EXAMPLE 3: Figure A 3 represents a cross-sectional drawing of a patient who is to be treated by moving-field ^{60}Co therapy. The axis of rotation (centre of convergence) is at point 0, and a full swing of 360° has been chosen. The beam size at the axis is 6X6 cm.

The data required to calculate the exposure time for each of the 15 treatment fractions chosen are given in Table A III.

The distance (d) from the axis (0) to the surface of the patient is measured at 20° intervals around the axis. The corresponding values of

TABLE A III. CALCULATION OF MOVING-FIELD TREATMENT

Angle	d (cm)	TAR	Angle	d (cm)	TAR
0	10.0	0.648	180	12.5	0.560
20	11.0	0.611	200	12.0	0.577
40	11.5	0.594	220	15.0	0.481
60	12.0	0.577	240	16.0	0.454
80	13.0	0.543	260	15.0	0.481
100	15.0	0.481	280	13.0	0.543
120	16.0	0.454	300	12.0	0.577
140	15.0	0.481	320	11.5	0.594
160	12.0	0.577	340	11.0	0.611

TAR as given in the tables [3] AI for the quality of radiation and field size being used are listed in columns 3 and 6 of the table. The mean value of the TAR is found to be 0.547. Previous measurements have shown that the 'air' dose rate at the axis is say 50.2 rad/min in soft tissue. The mean dose rate at the axis is hence given by the product:

$$\text{Mean TAR} \times \text{'air' dose rate} = 0.547 \times 50.2$$
$$= 27.5 \text{ rad/min in soft tissue}$$

If it is required to have a dose of 2000 rad in soft tissue at the axis, the exposure time is

$$\frac{2000}{27.5} = 72.7 \text{ min}$$

which represents 4 min 51 sec on each of 15 separate treatment sessions.

EXAMPLE 4: Figure A 4 represents the cross-section of a patient who is to be treated with three ^{60}Co beams, using an isocentric arrangement. The tumour centre is at point 0, as is the axis of rotation, the beam size at the axis is 6×8 cm and the source-axis distance (SAD) is 80 cm.

Table A IV lists the relevant data necessary to calculate the individual absorbed doses at the axis (0) for each beam. The total tumour dose at the axis is to be 6000 rad in soft tissue. In this example the beams are arranged symmetrically, i.e. at 120° apart, so that a satisfactory dose distribution is obtained if each beam delivers an equal contribution to the point 0. Since the total tumour dose is 6000 rad, each field must contribute 6000/3 = 2000 rad at the tumour centre. The corresponding axis absorbed dose in air is calculated using the formula:

$$\text{Axis absorbed dose in air} = \frac{\text{Axis dose}}{\text{TAR}}$$

and the resulting values are listed in the fourth column of Table A IV.

Previous measurement and the use of 'equivalent' squares (see sections 4.4.2 and 4.3.1) have shown that for a beam size of 6×8 cm the axial absorbed dose rate in air is 50.7 rad/min in soft tissue.

TABLE A IV. CALCULATION OF FIXED-FIELD ISOCENTRIC TREATMENT

Field	Surface-axis distance α (cm)	TAR	Absorbed dose at axis in air (rad)	Total exposure time (min)
Anterior	6.0	0.826	2420	47.7
Left posterior oblique	7.5	0.763	2620	51.6
Right posterior oblique	8.5	0.722	2770	54.6

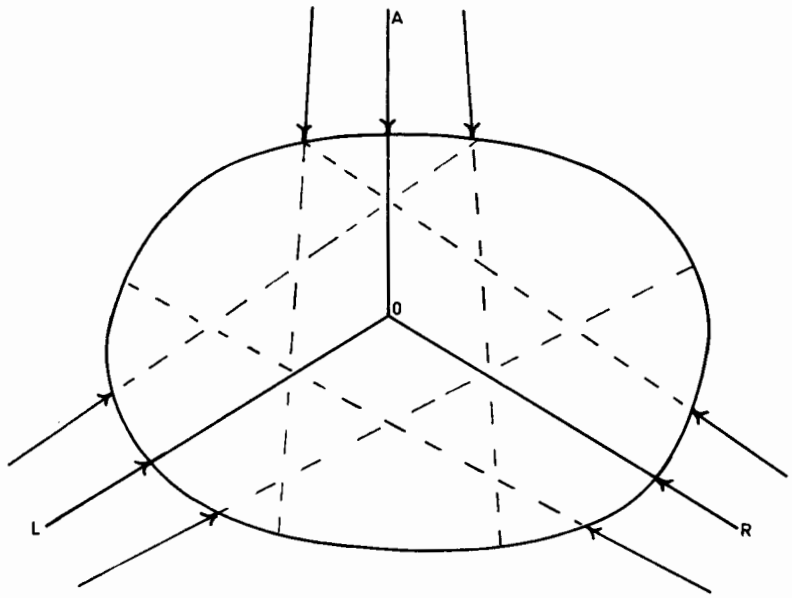


FIG. A4. Fixed beam treatment using isocentric arrangement.

The final column of the Table lists the total exposure time required on each beam, calculated using the formula:

$$\text{Total exposure time} = \frac{\text{Axis absorbed dose in air}}{\text{Axis absorbed dose rate in air}}$$

For example, the anterior beam will require a total exposure time of

$$\frac{2420}{50.7} = 47.7 \text{ min}$$

which represents an exposure of 3 min 11 sec on each of 15 separate treatment sessions.

If the beams are not symmetrical, a uniform dose distribution requires that the contribution to the axis from each beam be weighted appropriately. The details of this type of calculation follow the same principle as those described in Example 2 but are outside the scope of this manual.

APPENDIX V

USEFUL TABLES

TABLE A V. ^{60}Co DECAY FACTORS (half-life 5.26 yr)

Months	years					
	0	1	2	3	4	5
0	1.000	0.877	0.768	0.674	0.590	0.517
1	0.989	0.867	0.760	0.666	0.584	0.512
2	0.978	0.858	0.752	0.659	0.578	0.506
3	0.968	0.848	0.743	0.652	0.571	0.501
4	0.957	0.839	0.735	0.645	0.565	0.495
5	0.947	0.830	0.727	0.638	0.559	0.490
6	0.936	0.821	0.719	0.631	0.553	0.484
7	0.926	0.812	0.712	0.624	0.547	0.479
8	0.916	0.803	0.704	0.617	0.541	0.474
9	0.906	0.794	0.696	0.610	0.535	0.469
10	0.896	0.785	0.688	0.603	0.529	0.464
11	0.886	0.777	0.681	0.597	0.523	0.459
12	0.877	0.768	0.674	0.590	0.517	0.454

TABLE A VI. ^{137}Cs DECAY FACTORS (half-life 30 yr)

Months	years								
	0	1	2	3	4	5	6	7	8
0	1.000	0.977	0.955	0.933	0.912	0.891	0.871	0.851	0.831
3	0.994	0.972	0.949	0.928	0.907	0.886	0.866	0.846	0.827
6	0.989	0.966	0.944	0.922	0.901	0.881	0.861	0.841	0.822
9	0.983	0.960	0.938	0.917	0.896	0.876	0.857	0.836	0.817
12	0.977	0.955	0.933	0.912	0.891	0.871	0.851	0.831	0.812

TABLE AVII. VALUES OF RAD PER RÖNTGEN CONVERSION FACTOR f_λ

Radiation quality HVT	Soft tissue	Soft tissue cavity inside bone[a]	Compact bone[a]
0.5 mm Al	0.92	3.5	4.4
1.0	0.93	3.3	4.0
2.0	0.93	2.9	3.5
3.0	0.93	2.3	2.0
1.0 mm Cu	0.93	2.0	1.5
1.5	0.94	1.8	1.3
2.0	0.95	1.6	1.1
2.5	0.95	1.35	1.05
^{137}Cs γ-rays	0.96	1.15	0.92
^{60}Co γ-rays	0.96	1.05	0.92

[a] The values of f_λ for soft tissue within bone and for compact bone are very dependent upon the radiation spectrum and, in the case of the soft tissue within bone, also on the cavity size, especially for the softer qualities. The values given here are therefore approximate and intended only to indicate the order of magnitude, and refer to cavity sizes in the range 10 - 50 μm which are associated with 'bone' necrosis.

TABLE A VIII. VARIATION OF RADIATION QUALITY INSIDE A WATER PHANTOM WITH DEPTH, FIELD SIZE AND HVT OF INCIDENT RADIATION [6]

The figures given in the table are HVT in mm Cu.

Depth (cm)	Area (cm^2) 0	25	50	100	200	400
	Primary HVT 1.25 mm Cu					
0.0	1.28	0.98	0.92	0.87	0.84	0.82
2.5	1.38	0.85	0.72	0.66	0.61	0.58
5.0	1.55	0.82	0.72	0.63	0.56	0.52
10.0	1.83	0.90	0.75	0.62	0.49	0.46
15.0	2.24	0.99	0.75	0.60	0.50	0.47
	Primary HVT 2.20 mm Cu					
0.0	2.26	1.89	1.80	1.70	1.62	1.54
2.5	2.39	1.72	1.55	1.41	1.23	1.15
5.0	2.53	1.70	1.48	1.21	1.08	1.00
10.0	2.67	1.54	1.29	1.06	0.88	0.80
15.0	2.90	1.48	1.20	0.98	0.81	0.66
	Primary HVT 3.20 mm Cu					
0.0	3.16	2.78	2.67	2.56	2.45	2.37
2.5	3.19	2.45	2.26	2.09	1.93	1.81
5.0	3.26	2.46	2.22	1.91	1.70	1.56
10.0	3.38	2.31	1.98	1.69	1.44	1.27
15.0	3.46	2.11	1.75	1.45	1.18	0.88

APPENDIX VI

EQUIPMENT REQUIRED

		Approximate cost US $
A.	ESSENTIAL	
1.	Calibrated exposure meter suitable for range 50 kV ^{60}Co γ-rays: including 'equilibrium cap': 50-200 R full scale	1250
2.	Radioactive check source for use with 1 (and 3)	350
3.	Calibrated exposure meter suitable for range 10 - 50 kV - (? 100 kV) (this may be an extra chamber used with item 1)	1000 (200)
4.	Calibration of exposure meters over a range of qualities	250
5.	Radiation protection survey meter 0.1 - 200 mR/h	400
6.	Thermometer 0-30°C ($\frac{1}{2}$ degC divisions)	5
7.	Barograph or Barometer	80
8.	Stopwatch, 1 min per revolution	25
9.	Clamp stands, bosses, clamps	25
10.	15, 30, 100 cm steel rules 2 or 3 metre measuring tape	25
11.	Slide rule, graph paper (linear and log-linear)	20
12.	Sheets of copper, aluminium, lead, X-ray film	25
13.	Access to film badge (or other) monitoring service and to X-ray film development	
14.	Water phantom	250
B.	DESIRABLE	
15.	Exposure rate meter - multirange	1250
16.	Selection of ionization chambers, each	200
17.	Personnel (QFE) dosemeters, each	25
18.	Hygrometer	25
19.	Desk calculator	250
20.	Bolus material	50
21.	Tool kit	500
22.	Solid phantom	100
23.	Access to: (a) scintillation or geiger counting system, (b) mechanical, electrical and electronic workshops.	

Note: The cost of the items quoted is to be regarded as very approximate and is given for guidance only. Import duties, taxes and transport charges may necessitate additional expenditure.

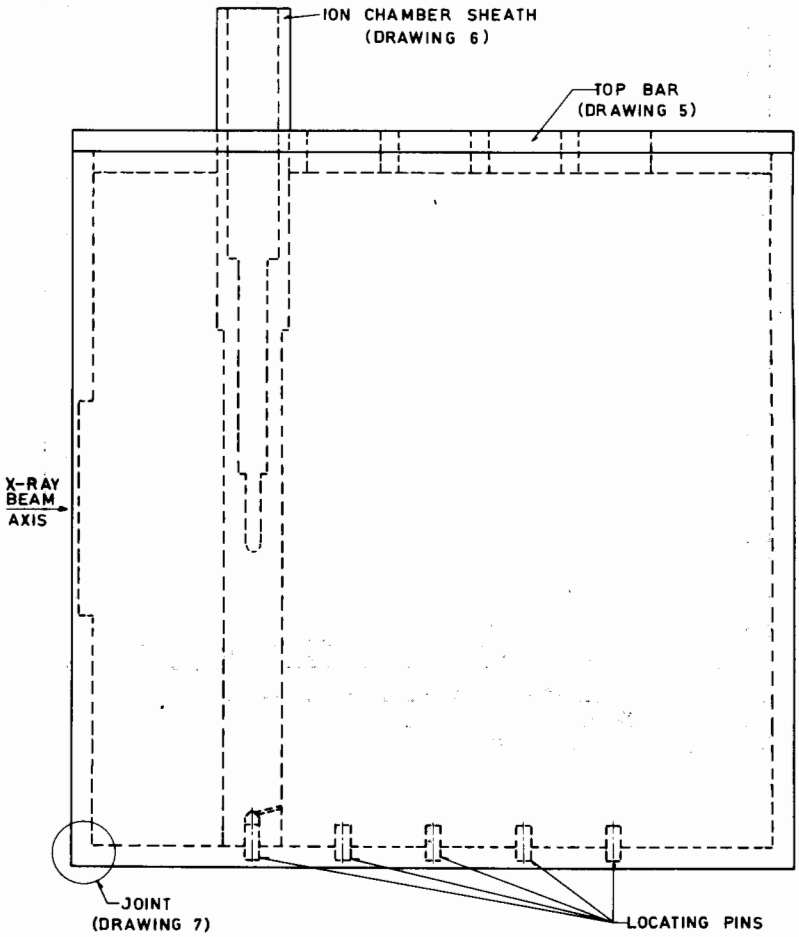
APPENDIX VII

SUGGESTED ORDER OF WORKING

		Section
A.	THE DOSEMETER	
1.	Test that the dosemeter behaves correctly.	3.5
2.	Establish the base lines for the routine tests and measurements on the dosemeter.	3.5.1, 3.5.2, 3.5.4
3.	Calibrate the dosemeter. (If this is to be done indirectly on the user's own X- or γ-ray unit it may be necessary to wait until after step 10, below.)	3.5.6
4.	Repeat the above at regular intervals.	
B.	THE X- OR γ - RAY UNIT	
5.	Check that the general electrical and mechanical behaviour of the equipment is satisfactory.	6.1
6.	Check that the equipment is in adjustment	6.2
7.	Protection survey	7.1
8.	Check that the radiation output is stable (at fixed kV, mA and/or fixed monitor reading).	2.2.2, 3.6
9.	Check the timer for accuracy and linearity, or check the monitor for linearity.	2.2.2.1, 2.2.3
10.	Measure HVTs	5.1
11.	Confirm selection of percentage depth dose values.	5.2
12.	Measure the relative outputs (plain and wedged beams).	4.2.6, 4.4.1
13.	Measure the output for the standard beam size etc.	4.2.3, 4.2.5
14.	Measure the mean 'air' dose for use in rotation therapy.	4.4.2
15.	Obtain the 'standard reading' for the standard check jig.	4.5.1
16.	Establish the routine tests; (a) mechanical (b) %depth dose; (c) radiation output, which are to be repeated regularly.	6.2, 4.5.3, 4.5.1
C.	TREATMENT	
17.	Select the beam sizes and configuration.	
18.	Calculate the required 'given' dose for each beam	A.4
19.	Calculate the exposure time (or monitor division) for each beam.	2.2.1, 2.2.2.

APPENDIX VIII

DESIGN OF A SUITABLE WATER PHANTOM



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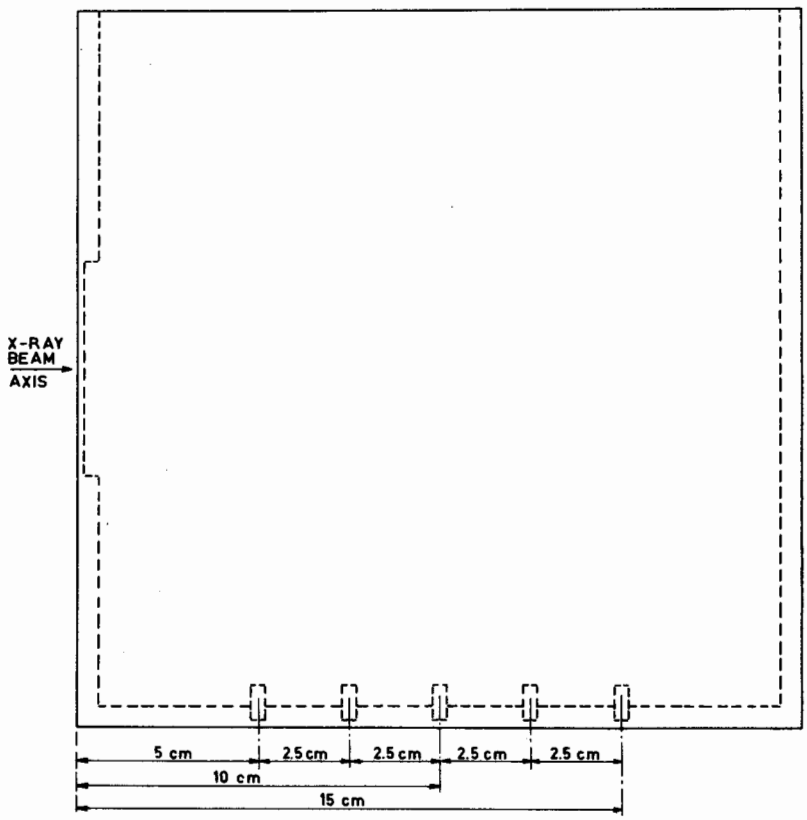

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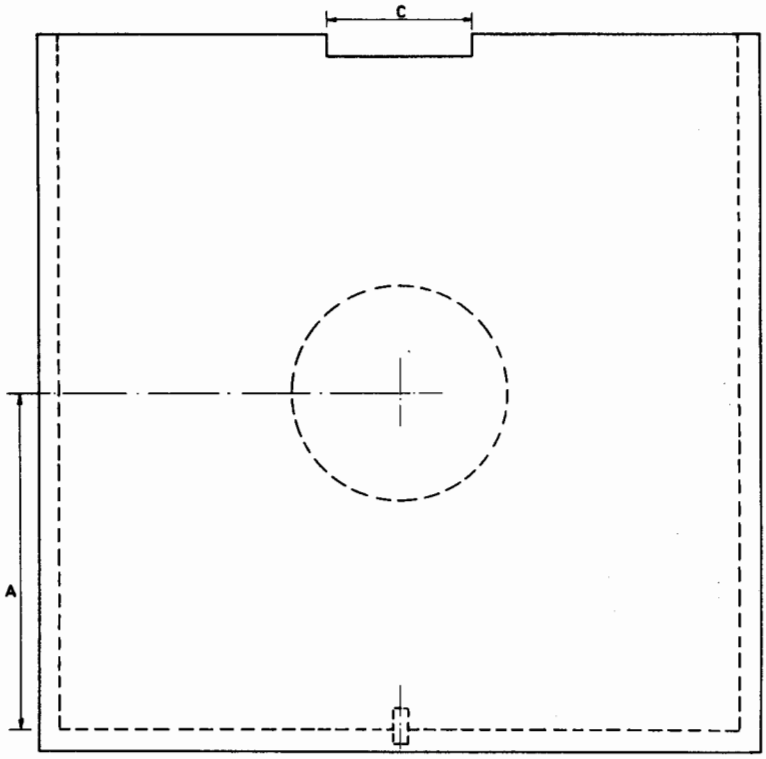

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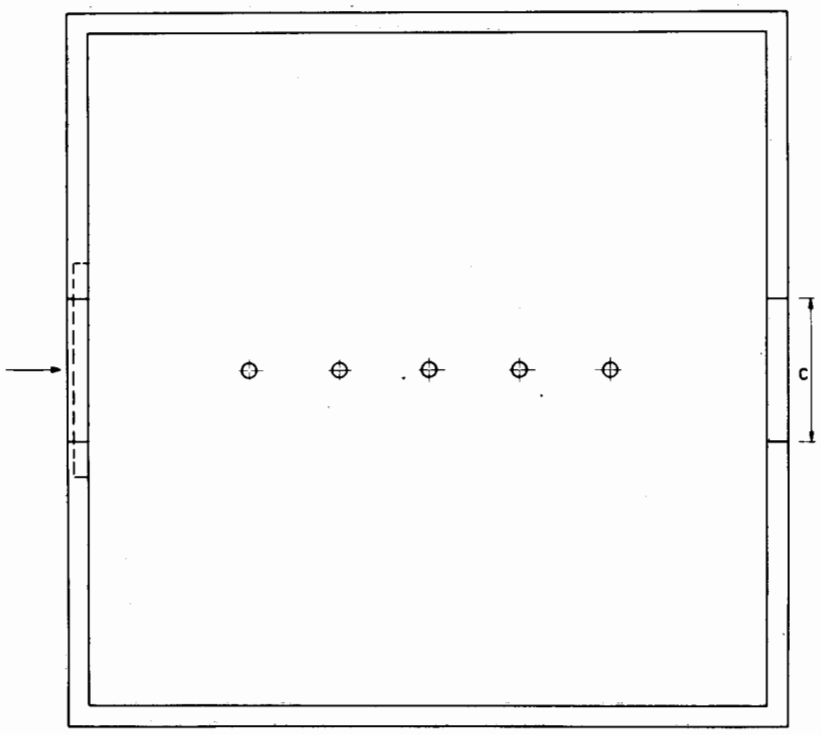

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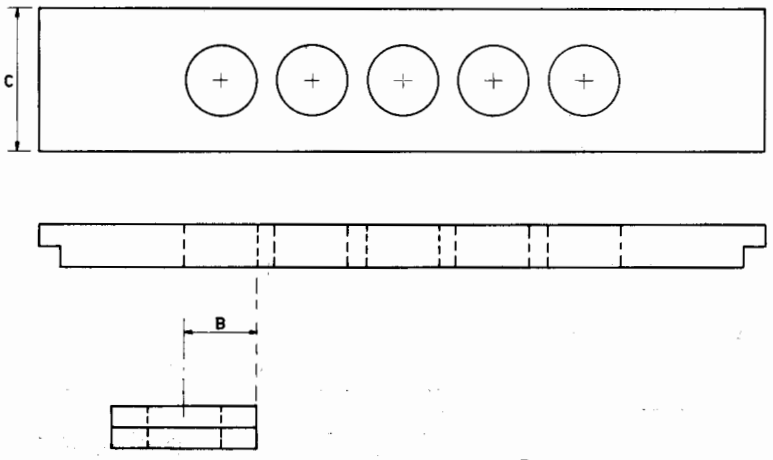

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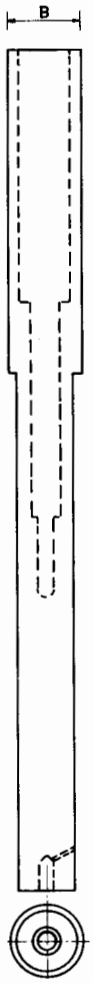

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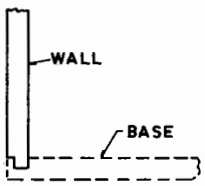

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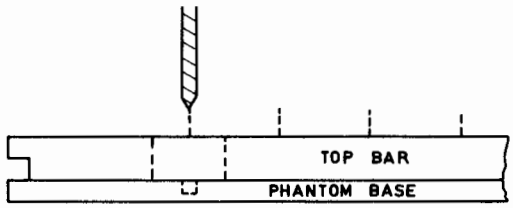

8

APPENDIX IX

EXAMPLES OF CHECK LISTS

1. INITIAL TESTING OF X-RAY EQUIPMENT (tick when test done and satisfactory)

X-ray machine number

A. General

Mechanical movements
Mechanical locks
Electrical earth
Electrical supply
Electrically stable
Cables clamped and in good condition
Electrical interlocks
Switching ON/OFF
Operating generally

B. Adjustment

Treatment cone axis
Central axis indication
Focal spot: distribution of blackening
: position
Treatment cones: aperture sizes
Beam uniformity
Light-beam diaphragm
Source-surface distance
Isocentre
Couch rotation
Couch vertical movement
Indicating lights

Date of test:
Name of tester:
Details of tests recorded in book number:
pages :

2. REGULAR DOSEMETER CHECKS (tick when test done and satisfactory)

Dosemeter number:

Battery voltages
Leak: whole system
: without chamber
Standard radioactive check
Radiograph
Intercomparison with local standard at (a) 100 kV
(b) 250 kV
(c) ^{60}Co

Date of test:
Name of tester:
Details of tests recorded in book number:
pages:

3. REGULAR OUTPUT CHECK (tick when test done and satisfactory)

X-ray machine number

Quality check (5 : 15 cm deep ratio)
Timer check/shutter ON/OFF error
Standard output check
Any adjustment made?

Date of test:
Name of tester:
Details of test recorded in book number:
pages:

APPENDIX X

List of participants at the IAEA Panel on Dosimetric Requirements of Radiotherapy Centres, held in Caracas, Venezuela, 22 - 26 April 1968.

Participants marked with an asterisk formed the subcommittee which drew up the recommendation leading to the present manual.

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