Methods of Evaluating Radiological Equipment and Materials

Recommendations of the International Commission on Radiological Units and Measurements

Handbook 89

United States Department of Commerce
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*In preparation.

UNITED STATES DEPARTMENT OF COMMERCE • Luther H. Hodges, *Secretary*

NATIONAL BUREAU OF STANDARDS • A. V. Astin, *Director*

Methods of Evaluating Radiological Equipment and Materials

Recommendations of the International Commission on Radiological Units and Measurements (ICRU) Report 10f 1962

National Bureau of Standards Handbook 89

Issued August 23, 1963

(This publication supersedes parts of H78. Handbooks 84 through 89 will completely replace H78. For an explanation, see the Foreword. Also, for a list of these titles, see page 3 of cover.)

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Foreword

The reports of The International Commission on Radiological Units and Measurements for a number of years have been published by the National Bureau of Standards in the Handbook series. In the past, each of the triennial reports of the ICRU represented a complete restatement of the recommendations of the Commission. Because of the increasing scope of its activities, however, the Commission in 1962 decided to modify the previous practice. It will issue a series of reports presenting the current recommendations of the Commission. Each report will cover a particular portion of the area of interest to the ICRU. This procedure will facilitate revision of ICRU recommendations and also spread out in time the workload of the Commission. This Handbook is one of the new series presenting the recommendations of the Commission on one aspect of the field with which the Commission is concerned. It presents recommendations agreed upon at the meeting of the Commission held in Montreux, Switzerland, in April 1962.

The National Bureau of Standards is pleased with its continuing opportunity of increasing the usefulness of these important reports by providing the publication outlet.

A. V. Astin, *Director*.

Contents

Preface

A. Scope

The International Commission on Radiological Units and Measurements (ICRU), since its inception in 1925, has had as its principal objective the development of internationally acceptable recommendations regarding:

(1) Quantities and units of radiation and radioactivity,

(2) Procedures suitable for the measurement and application of these quantities in clinical radiology and radiobiology,

(3) Physical data needed in the application of these procedures, the use of which tends to assure uniformity in reporting.

The Commission also considers and makes recommendations on radiation quantities, units and measurements in the field of radiation protection. In this connection, its work is carried out in close cooperation with the International Commission on Radiological Protection.

B. Policy

The ICRU endeavors to collect and evaluate the latest data and information pertinent to the problems of radiation measurement and dosimetry and to recommend the most acceptable values for current use.

Recognizing the confusion that exists in the evaluation of different radiological equipment and materials, the ICRU is studying standard methods of determination of characteristic data for the equipment and materials used in diagnostic and therapeutic radiology. This activity is confined to methods of measurement and does not include the standardization of radiological equipment or parts thereof.

The Commission's recommendations are kept under continual review in order to keep abreast of the rapidly expanding uses of radiation.

The ICRU feels it is the responsibility of national organizations to introduce their own detailed technical procedures for the development and maintenance of standards. However, it urges that all countries adhere as closely as possible to the internationally recommended basic concepts of radiation quantities and units.

The Commission feels its responsibility lies in developing a system of quantities and units having the widest possible range of applicability. Situations may arise from time to time when an expedient solution of a current problen may seem advisable. Generally speaking, however, the Commission feels that action based on expediency is inadvisable from a long-term viewpoint; it endeavors to base its decisions on the long-range advantages to be expected.

In 1955 the Commission entered into an official relationship with the World Health Organization (WHO). In this relationship, the ICRU will be looked to for primary guidance in matters of radiation units and measurements, and in turn WHO will undertake the worldwide dissemination of the Commission's recommendations. In 1960 the ICRU entered into consultative status with the International Atomic Energy Agency (IAEA).

The above relations with other international bodies do not affect the basic affiliation of the Commission with the International Society of Radiology.

The ICRU invites and welcomes constructive comments and suggestions regarding its recommendations and reports. These may be transmitted to the Chairman.

C. Current Program

A two week meeting of the ICRU was held in Montreux, Switzerland, April 2 to April 14th 1962. This meeting included the Main Commission and all the Committees that had reports prepared for final approval. Some 70 persons attended. An additional meeting of the Commission and Committee Officers was held in Ottawa from August 21 to August 23, 1962, for the principal purposes of the preparation of the status report for the Xth International Congress of Radiology and the outlining of program objectives for the next several years.

Several meetings of committees or committee task groups have been held during the past three years. There were meetings of various task groups of the Committee on Standards and Measurement of Radiological Exposure—Paris in January 1961 and London in April and September 1961. The Committee on Radiobiological Dosimetry also held a meeting in April 1961. The ICRU was also represented at a meeting of the Consultative Committee on ionizing radiation of the International Committee of Weights and Measures at Sèvres in October 1961.

As noted in the last report, two joint committees had been established between the ICRU and the ICRP. The Joint Committee on RBE has met twice with ICRU participation. The Committee on Methods and Instruments for Radiation Protection has not met.

Upon the request from the United Nations Scientific Committee on the Effects of Atomic Radiations, the ICRU and the ICRP agreed to undertake a second study dealing with the Medical and Physical Parameters in Clinical Dosimetry. This committee met in New York for one week in September 1959 and for a week in Stockholm in June 1960. A report of this study entitled "Exposure of Man to Ionizing Radiation Arising from Medical Procedures with Special Reference to Radiation Induced Diseases, An Inquiry into Methods of Evaluation", was published in Physics in Medicine and Biology **6,** No. 2, 199 (Taylor and Francis, Ltd., London, England, Oct. 1961).

Reports and recommendations of the ICRU originally designed for medical applications, have come into common use in other fields of science, particularly where "dosimetric" considerations are involved. For this reason the committees have included in their membership some scientists having competence outside of the medical radiology field. Material in the reports is designed to meet physical, biological, and medical requirements wherever possible.

This has introduced a small problem in terminology. The name of the Commission includes the term "radiological". In many European countries the term "radiological" is taken as inclusive of both the physical and biological sciences. In other countries, the United States, for example, "radiological" appears to carry the primary connotation of relationship to medicine. It therefore may be desirable to change the name of the Commission from "Radiological" to "Radiation". It is believed that this would be properly understood by all concerned. The question has been debated by the Commission, but final action is being delayed for future consideration.

D. The Current Series of Reports

Hitherto, the triennial reports of the ICRU have been published in single volumes. However the reports are now becoming too extensive, and in some cases too specialized, to make a single publication practicable. Beginning with this 1962 series, the ICRU reports will be issued in smaller entities, each dealing with a limited range of topics. The 1959 Report will not be reprinted. Revisions of the 1962 series will be undertaken individually as circumstances warrant. A full listing of ICRU recommendations, including the present series, is given on page iii of the cover of this report.

The current report series includes revisions of much of the material that appeared in the 1959 report in addition to a number of new topics. The following summary indicates some of the highlights of the current report series.

Radiation Quantities and Units (Report 10a)—One of the most important changes is the revision of the section on quantities and units. This revision resulted from the thorough study by the *Ad Hoc* Committee on Quantities and Units mentioned above. It includes new names for certain quantities and clarified definitions for others. It presents a system of concepts and a set of definitions which is internally consistent and yet of sufficient generality to cover present requirements and such future requirements as can be foreseen.

Physical Aspects of Irradiation (Report 10b)—This report deals broadly with the physical aspects of irradiation with a considerable amount of new material added since the 1959 report. It includes an extensive discussion of the various techniques for the measurement of absorbed dose as well as exposure. Characteristics of radiation instrumentation are covered in some detail including the more sophisticated work on standards. The section on spectra has been up-dated and a new section added on neutron measurements and standards. Available data for stopping power ratios and the average energy (W) required to produce an ion pair in a gas have been reviewed. On the basis of this review it has been necessary to modify the previous ICRU tables for these factors. This modification amounts to about 1 or 2 percent change in stopping power ratios and up to 1 percent in W.

Radioactivity (Report 10c)—The portions of the report dealing with direct and relative measurements of radioactivity and the availability and requirements for radioactivity standards, and the parts dealing with the techniques and measurements of radioactivity in hospitals and biological laboratories are revisions of the 1959 report, embracing a review of the developments that have occurred since that report and bringing up to date the material included. In addition, a new section on low level radioactivity in materials as related to the problems of radiological measurements has been added. This topic is important because of the problems arising from the contamination, or possible contamination, in the last decade of a great many of the materials used in the construction of counting equipment, shields, and in the reagent chemicals employed in radioactivity measurements.

Clinical Dosimetry (Report 10d)—Much of the Commission's work on clinical dosimetry is brought together in this report. Included is an extensive discussion of practical calibration procedures and the determination of dose along the central ray. Depth dose data relative to stationary and moving-field therapy have been extended as have the conversion data necessary to relate ionization measurements to absorbed dose.

The principal effort has been toward the definition of nomenclature and the indication of methods. While some examples are given and data are provided for these, in general the reader is referred to other published data. The report considers ways of increasing the accuracy and comparability in clinical dosimetry. The discussion includes not only the physical aspects of dose measurement but also the wider subject of planning treatment in such a way as to deliver the prescribed absorbed dose to a defined "target volume". It also includes comments upon the common sources of error in clinical dosimetry and discusses the information which should be recorded during treatment and that which should be reported about any new treatment technique. Appendices to this report include pertinent material taken from other reports in this series.

Radiobiological Dosimetry (Report 10e)—This report deals primarily with radiobiological dosimetry, and considers methods of improving the accuracy and intercomparability of absorbed dose measurements in radiobiology. It is in effect a handbook for the experimental radiobiologist. It emphasizes the great importance of planning the experimental work in a way which makes the dosimetry easier and more accurate and it illustrates how this can be done.

E. Operating Funds

Throughout most of its existence, the ICRU has operated essentially on a voluntary basis, with the travel and operating costs being borne by the parent organizations of the participants. (Only token assistance was available from the ISR.) Recognizing the impracticability of continuing this mode of operation on an indefinite basis, operating funds were sought from various sources in addition to those supplied by the International Society of Radiology.

Prior to 1959, the principal financial assistance to the ICRU had been provided by the Rockefeller Foundation which supplied some $11,000 to make possible various meetings. In 1959 the International Society of Radiology increased its contribution to the Commission to $3,000 to cover the period until the Xth Congress. In 1960 the Rockefeller Foundation supplied an additional sum of some $4,000 making possible a meeting of the Quantity and Units Committee in 1960.

In 1960 and 1961 the World Health Organization contributed the sum of $3,000 each year to the Commission for carrying forward its work. This was increased to $4,000 in 1962. It is expected that this sum will be allocated annually, at least for the next several years. In addition, the WHO has provided substantial assistance to the Commission in providing meeting space, secretarial services, etc., for the meetings held in Geneva and Montreux.

In connection with the Commission's Joint Study with the ICRP, the United Nations allocated the sum of $10,000 for the joint use of the two Commissions for the purpose of carrying out their second study. This fund has been administered by the ICRP.

The most substantial contribution to the work of the ICRU has come from the Ford Foundation through the particular efforts of Dr. Paul Pearson. Effective in December 1960, the Ford Foundation made available to the Commission the sum of $37,000 per year for a period of five years. This money is to be used for such items as travel expenses to meetings, for secretarial services and other operating expenses. To a large extent, it is because of this grant that the Commission has been able to hold the several meetings considered to be necessary to move forward actively with its program.

The International Atomic Energy Agency has allocated the sum of $6,000 per year for use by the ICRU. It is expected that this sum will be allocated annually at least for the next several years.

A valuable indirect contribution has been made by the U.S. National Bureau of Standards where the Secretariat has resided. The Bureau has provided substantial secretarial services, reproduction services and travel costs in the amount of several thousands of dollars.

The Commission wishes to express its deep appreciation to all of these and other organizations that have contributed so importantly to its work.

F. Composition of the ICRU

(a) It is of interest to note that the membership of the Commission and its committees for the period 1959–62 totals 139 persons drawn from 18 countries. This gives some indication of the extent to which the ICRU has achieved international breadth of membership within its basic selection requirement of high technical competence of individual members.

(b) The membership of the Main Commission during the preparation of this report was as follows:

Lauriston S. Taylor, Chairman	United States.
L. H. Gray, Vice-chairman	United Kingdom.
H. O. Wyckoff, Secretary	United States.
K. K. Aglintsev	U.S.S.R.
A. Allisy	France.
R. H. Chamberlain	United States.
F. Ellis	United Kingdom.
H. Fränz	Federal Republic of Germany.
H. E. Johns	Canada.
W. J. Oosterkamp	Netherlands.
B. Rajewsky	Federal Republic of Germany.
H. H. Rossi	United States.
M. Tubiana	France.

G. Composition of Committee Preparing Initial Draft of Present Report

Members of ICRU Committee IV on Methods of Evaluating Radiological Equipment and Materials

B. Combee, Chairman
E. D. Trout, Vice Chairman
E. Zieler, Technical Secretary
H. Berger
J. Massiot
B. J. O'Loughlin
S. W. Smith
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W. Hondius Boldingh, Chairman
S. Ledin
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K. H. Reiss
J. Remy
R. W. Stafford
M. Ter-Pogossian

Members of Subcommittee on Body Section Equipment

W. Watson, Chairman
P. de Vulpian
F. E. Stieve
O. Vallebona
B. G. Ziedses des Plantes

H. The Present Report

Changing concepts of what represents acceptable and necessary methods, materials and systems consistent with current medical requirements in radiology call for agreement as to methods for measurement and evaluation of radiological materials and systems. Numerous factors involved in this field contribute importantly to the results in medical procedure. Agreement is needed on the ways to describe and measure these basic characteristics. It was this need which led to the decision of the ICRU to formulate recommendations in this area.

This report is the first of a new group of ICRU reports dealing with methods of evaluating radiological equipment and materials. The aim of these reports will not be to develop standards for equipment and materials, but rather to make recommendations that may be used as a basis for standards. This report includes a revised discussion on the measurement of focal spots and new sections on grids, image intensifiers and body section equipment.

Methods of Evaluating Radiological Equipment and Materials

International Commission on Radiological Units and Measurements (ICRU) Report 10f 1962

I. Measurement of Focal Spot Sizes (Revision)

A. Methods of Measurement of Dimensions of Focal Spots of X-ray Tubes

The ICRU reaffirms, in essence, its 1959 recommendation (see reference 1 at the end of this chapter), regarding "Method of Focal Spot Image Formation and Measurement", and extends this recommendation to include therapy tubes for voltages up to 300 kv.

1. *Basic Method:*

An image of the focal spot is produced on fine-grain x-ray film by means of an x-ray pinhole camera. Focal spot, pinhole, and film are on the axis of the cone of radiation used in normal operation ("central ray"). The size of the focal spot is determined from the image.

2. *Specifications of Pinhole Camera:*

a. *Pinhole Diameter:* .030 mm for focal spot sizes below 1.0 mm; .075 mm for focal spot sizes 1.0 mm to 2.5 mm; .100 mm for focal spot sizes above 2.5 mm.

b. *Pinhole Diaphragm Construction:* The pinhole diaphragm shall be manufactured from a 90:10 gold-platinum alloy. Dimensions shall be in accordance with the table and figure I.1 below:

Nominal pinhole diameter (mm)	*D* (mm)	*L* (mm)
.030	.030±.005	.075±.010
.075	.075±.005	.350±.010
.100	.100±.005	.500±.010

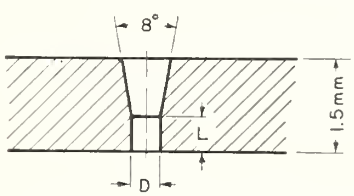

Figure I.1. *Cross-sectional view of pinhole indicating the legends for the important dimensions.*

c. *Distance from Focal Spot to Pinhole:* 10 cm minimum.

d. *Enlargement Factors:* 2.0 minimum, for focal spot sizes up to 2.5 mm; 1.0 minimum, for focal spot sizes above 2.5 mm. The "enlargement factor" is the ratio of the pinhole-film distance to the focal-spot-pinhole distance. If the distance between focal spot and pinhole cannot be measured directly, it is recommended that dual pinholes spaced approximately 7 mm apart be used in the camera, to produce dual images from which exact location of the focal spot can be determined by triangulation.

3. *Photographic Technique:*

a. *Type of Film:* Any fine-grain x-ray film (single coating).

b. *Image Density:* 0.8 to 1.2 as measured in the most dense portion of image. Standard developing technique should be employed. Density shall be regulated by means of exposure time only. Tube current and voltage shall be fixed at values specified for conditions under which focal spot measurement is stipulated. Fog density should not exceed 0.1.

4. *Measurement of Image:*

a. *Lighting:* Back-lighted at approximately 20 foot-candles.

b. *Measurement Procedure:* Use magnifying glass with built-in scale, 0.1 mm divisions, 5x to 10x magnification.

For line-focus tubes, measure width and length, respectively, including all perceptible portions of the image. For non-rectangular focal spots, measurements should be taken of all significant dimensions.

c. *Correction of Image Measurements:* All dimensions shall be divided by the enlargement factor.

5. *Statement of Dimensions:*

a. *For Line-Focus Tubes:* Width of focal spot shall be stated as the width of the image divided by the enlargement factor. Length of focal spot shall be stated as the length of the image divided by the enlargement factor, further corrected by a multiplier of 0.7 (tentative value). The need for a fractional multiplier for correction of the measured image length arises from the fact that the lengthwise distribution of energy in the focal spots of line-focus tubes tends to be peaked in the center and diminishes gradually to zero at the ends. Hence, the effective length, based on its effect on both radiographic definition and loading capacity, cannot be stated as equal to the measured image length as corrected only by the enlargement factor. The proposed correction factor of 0.7 is a tentative one pending more detailed investigation.

b. For non-rectangular focal spots, all other significant dimensions shall be stated as indicated by actual measurements divided by the enlargement factor.

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B. Detailed Specifications of Pinhole Camera

The above statement describes the essential details of the pinhole camera to be used for focal spot image formation. Exact details of its construction within the limits specified should be left to the discretion of the individual.

C. Measurement of Size of Radioactive Sources

The size of a radioactive source is merely the internal diameter of the capsule containing the radioactive material. There is no need to specify how this diameter is to be measured.

D. Investigation of the Accuracy of Focal Spot Measurement

In this connection, the ICRU has had the benefit of earlier work previously reported, especially as covered by the references listed at the end of this chapter. The current investigations of which the commission is cognizant tend to confirm the accuracy of the recommended method. However, the proposal of a fractional multiplier for correction of the measured image length in the case of line-focus diagnostic tubes, to compensate for the lengthwise distribution of energy, has not been fully evaluated. The ICRU has not been able to carry out a previous recommendation to enlist a suitable research organization to investigate this matter comprehensively, with the objective of establishing a proper formula for focal spot length. We now recommend that the proposed correction factor of 0.7 be established on a tentative basis, as a part of the ICRU Recommended Method and that additional work should be done to examine the accuracy of this factor. The program on measurement of energy distribution in focal spots, discussed in subsection E, may provide useful information in this connection.

It must be recognized that the effect of any off-focus radiation on definition is not taken into account by these measurements, and unless off-focus radiation is suppressed, the definition to be expected from the tube may be adversely influenced by it.

E. Method of Measurement of Energy Distribution on Focal Spots

It is proposed that the energy distribution be examined by a microdensitometry scan of the density pattern of the focal spot image obtained by the recommended procedure. Film densities will be converted to energy values by means of the exposure density curve of the film. There have not been sufficient results with such a method, or reports of such work, to enable an evaluation of its usefulness at this time.

F. References

1. Report of the International Commission on radiological units and measurements, 1959. NBS Handb. 78 (1961).
2. Fenner, E., Jochim, H., Determination of the focusing distance based on the geometric shape of the blur (lack of focus), (in German), Fortschr. Rontgenstr. **87,** 109 (1957).
3. Kuntke, A. H. G. On the determination of roentgen tube focal spot sizes by pin-hole camera roentgenography. Acta Radiol. **47,** 55 (1957).
4. Polansky, D., O'Connor, D. T., X-ray focal spot measurement. Nondestructive Testing **12,** 1 (1954).
5. Rogers, T. H. Definition and measurment of the focal spot size of an x-ray tube. Presented, Eighth Intern. Congress of Radiology, Mexico City, July 24, 1956.
6. Servais, J., Control of the emission source (focus) in radiogenic (X-ray) tubes and the relationship between the dimensions of the focus and the sharpness as related to radiographic images, (in French), J. Radiol. Electrol. **39,** 383 (1958).

II. Measurement of the Conversion Factor of X-Ray Image Intensifiers

A. Terminology

1. *X-ray image intensifier:* a device which converts instantaneously an x-ray pattern into a corresponding light image of higher energy density.

2. *X-ray image scanning system:* a device which uses scanning techniques to convert an x-ray pattern into an electrical signal which can be displayed on a display tube.

3. *X-ray image recording or display system:* a device using electronic intensification and by which an x-ray pattern is converted and displayed or recorded.

4. *X-ray image intensifier tube:* a vacuum tube containing an input screen which converts an x-ray pattern into an electron pattern, and in which the electrons are accelerated and focused on to an output screen which converts the electron pattern into a light image of higher density.

5. *Light amplifier tube:* a vacuum tube containing a photoemissive layer which converts a light image into an electron pattern, and in which the electrons are accelerated and focused onto an output screen which converts the electron pattern into a light image of higher luminance.

6. *Input screen:* the surface, either flat or curved, in which x radiation is absorbed and in which the x-ray image is converted into either an image of fluorescent light or of electric charge. (See fig. II.1.)

7. *The input plane:* the plane which intersects the input screen at the periphery of the area specified by the manufacturer. (See fig. II.1.)

8. *The output screen of small-ray image intensifier:* the surface in which is located the phosphor which presents the light image. (See fig. II.1.)

9. *The optical output surface:* the surface in which the output phosphor appears to lie when observed, taking into account the shortening of the optical path caused by the tube window. (See fig. II.1.)

10. *The entrance plane of an x-ray image intensifier:* the plane in front of the input screen, through which, according to the specification of the manufacturer, no part of the image intensifier will protrude. (See fig. II.1.)

11. *X-ray pattern:* the distribution of x-ray exposure rate on a surface.

12. *Conversion factor of an x-ray image intensifier:* the ratio of the luminance of the output screen to the input exposure rate* of the applied x radiation.

Symbol: G_x

Unit: $\mathrm{cd} \cdot \mathrm{m}^{-2} \cdot (\mathrm{mR})^{-1} \cdot \mathrm{sec}$

13. *Definitions of terms used in illumination field:* terms and definitions on illumination are derived from the International Lighting Vocabulary published by the "International Commission on Illumination" in their publication CIE–1.1 (1957), Paris.

a. *Radiant power (or radiant flux):* power emitted, transferred, or received in the form of radiation.

Symbol: Φ_e, F_e, P

Unit: watt.

b. *Relative luminous efficiency of a monochromatic radiation, of wavelength* λ: the ratio of the radiant flux at wavelength λ_m to that at wavelength λ which produces equally intense luminous sensations under specified photometric conditions, λ_m being chosen so that the maximum value of this ratio is unity.

Symbol: V_λ.

Unless otherwise indicated, the value used for the relative luminous efficiency relates to photopic vision by a normal eye having the characteristics laid down by the CIE (see fig. II.2).

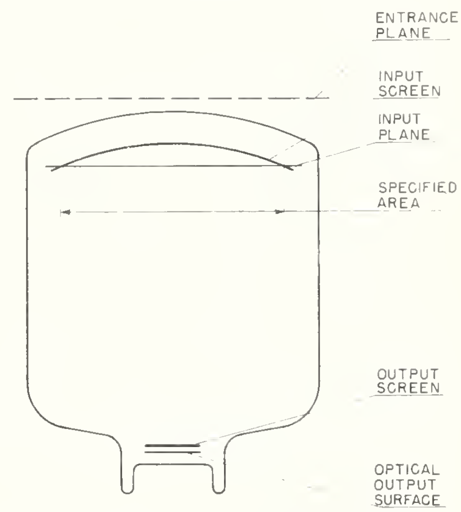

Figure II.1. *Schematic view of an image intensifier tube which illustrates some of the terminology defined in II.A.*

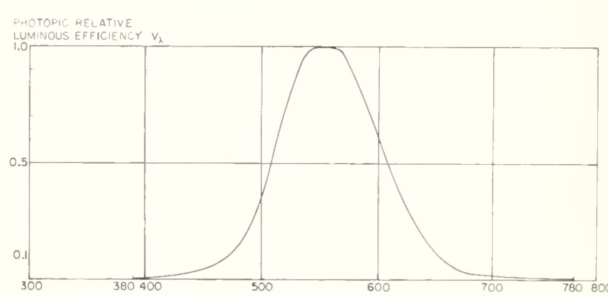

Figure II.2. *Variation of photopic relative luminous efficiency with light wavelength.*

*The text of ICRU Report 10a (reproduced as appendix I of this report) contains definitions of various radiation quantities and units.

c. *Luminous flux:* the quantity characteristic of radiant flux which expresses its capacity to produce a luminous sensation, evaluated according to the values of relative luminous efficiency.

Unless otherwise indicated, the luminous flux in question relates to photopic vision, and is connected with the radiant flux in accordance with the formulas adopted in 1948 by the CIE, i.e., by the relation

$$\Phi = K_m \int P_\lambda V_\lambda d\lambda$$

in which $P_\lambda d\lambda$ is the radiant flux corresponding to the radiation comprised between λ and $\lambda + d\lambda$ and V_λ is the relative luminous efficiency, the values of which as a function of λ are given above. Applied to the radiation of a full radiator at a temperature of solidification of platinum, the preceding formula determines the value of $K_m (lm/W)$.

Symbol: Φ, F $\quad \Phi = K_m \int P_\lambda V_\lambda d\lambda$

Unit: lumen (lm) $\quad P_\lambda \left(= \frac{dP}{d\lambda} \right) = \Phi_{e\lambda}$.

d. *Luminous intensity* (in a given direction): the quotient of the luminous flux emitted by a source, or by an element of a source, in an infinitesimal cone containing the given direction, by the solid angle of that cone.

Symbol: I $\quad I = \frac{d\Phi}{d\omega_1}$

Unit: candela (cd).

e. *Luminance* (at a point of a surface and in a given direction): the quotient of the luminous intensity in the given direction of an infinitesimal element of the surface containing the point under consideration, by the orthogonally projected area of the element on a plane perpendicular to the given direction.

Symbol: L, B $\quad L = \frac{d^2\Phi}{\cos \epsilon_1 dA_1 d\omega_1}$

Unit: $cd/m^2 = 1$ nit

$= 10^{-4}$ stilb

$= \pi$ apostilb

$= \pi \cdot 10^{-4}$ lambert

$= \frac{1}{3426}$ foot-lamberts.

f. *Relative spectral energy distribution:* description of the spectral character of a radiation (description of a light) by the way in which the relative spectral concentration of some radiant quantity (e.g., radiance) varies throughout the spectrum.

Symbol: S_λ.

g. *Photopic vision:* is vision mediated essentially or exclusively by the cones. It is generally associated with adaptation to a luminance level of at least several cd/m^2.

B. Introduction

The need for an increased luminance in fluoroscopy and for a higher sensitivity in radiography has led to the development of intensifying systems in which both electronic and optical techniques are used.

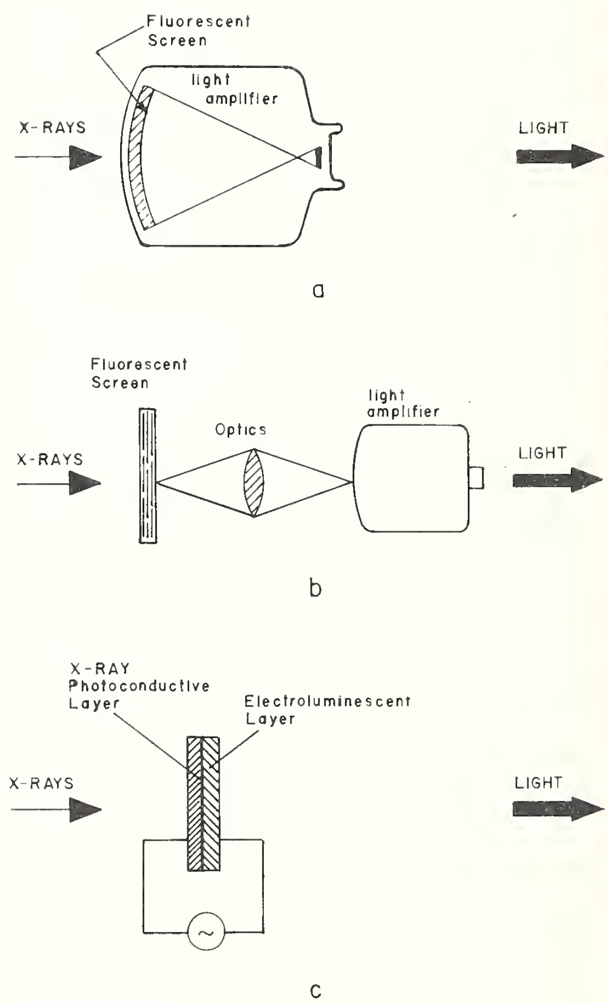

FIGURE II.3. *Image intensifiers.*

a. Image intensifier tube.
b. Fluoroscopic screen and optical amplifier system.
c. Solid state image intensifier.

The different x-ray image intensifying systems can be divided into the following three groups:

1. *X-ray image intensifiers:* electronic devices which convert instantaneously an x-ray pattern into a corresponding light image of higher energy density.
Examples are:

a. Electronic tube incorporating a fluorescent screen and light amplifier, commonly called x-ray image intensifier tube (fig. II.3a).

b. Device including fluoroscopic screen, optical system and light amplifier (fig. II.3b).

c. Solid state x-ray image intensifier, including an x-ray photoconductive layer in contact with an electro-luminescent layer (fig. II.3c).

2. *X-ray image scanning systems:* devices which use scanning techniques to convert an x-ray pattern into an electrical signal which can be displayed on a display tube.
Examples are:

a. Device including fluoroscopic screen optically coupled to a TV camera tube (fig. II.4a.).

b. Device including an x-ray image intensifier coupled to a TV camera tube by means of an optical system (fig. II.4b).

c. X-ray photoconductive TV camera tube (fig. II.4c).

3. *X-ray image recording or display systems:* devices using electronic intensification and by which an x-ray pattern is converted and displayed or recorded.

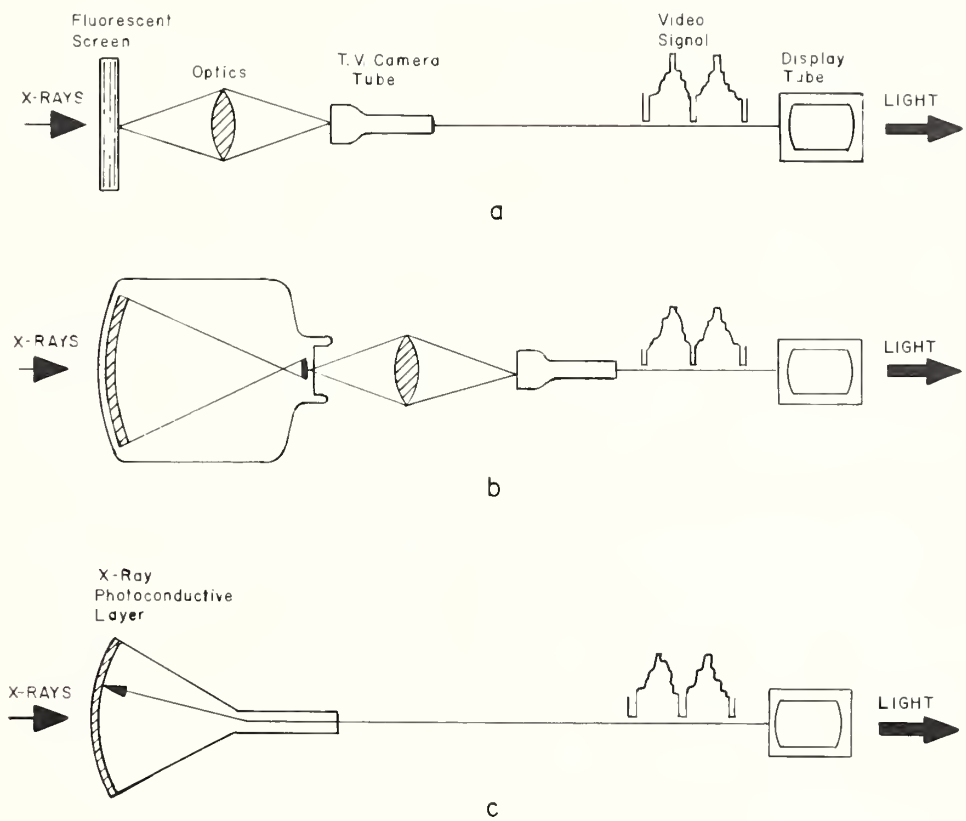

FIGURE II.4. *Image scanning systems.*

a. Fluroscopic screen optically coupled to a TV camera tube.
b. Image intensifier coupled to a TV camera tube.
c. Photoconductive TV camera tube.

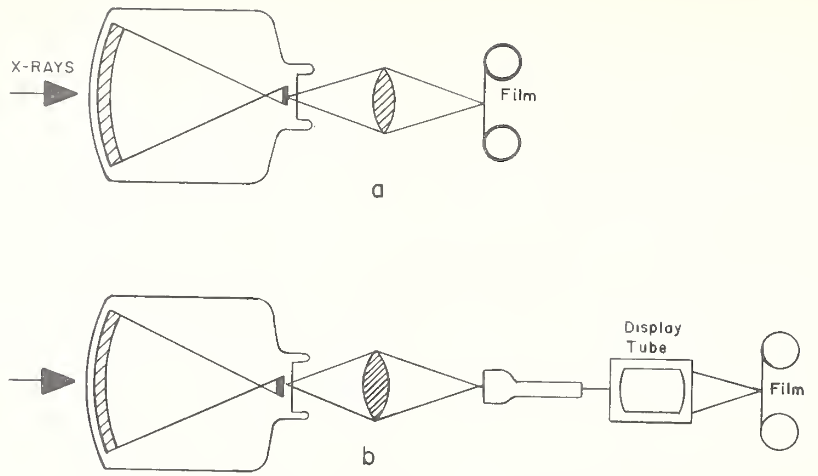

FIGURE II.5. *Image intensifier optically coupled to a cine camera.*

a. Image intensifier optically coupled to a cine camera.
b. Image intensifier optically coupled to a photographic camera.

Examples are:

a. An x-ray image intensifier or an x-ray scanning system optically coupled to a cine or photographic camera (figs. II.5a and II.5b).

b. Magnetic recording system in combination with an x-ray scanning system.

c. Memory tubes.

This report is concerned with x-ray image intensifiers (group 1 of page 5).

C. The Concept of "Conversion Factor"

Since there is no unanimity of opinion on the definition of the brightness intensification of x-ray image intensifiers, it is felt that there is a need to make recommendations to prevent misunderstanding in the practical application of the published data.

The term "intensification factor" was used in radiology before image intensification was introduced. Intensifying screens had long been in use in radiographic techniques but, even so, considerable confusion existed in the data published on the intensification factors of these screens. Usually only one figure was given with the object of indicating the ratio of the exposure to be used for a film-screen combination to the exposure for a film alone. Characteristics such as x-ray wavelength-dependence of this factor and the different behavior of a film emulsion to x radiation and visible light were neglected.

In radiology different definitions of intensification factors of x-ray image intensifiers have been used. They were all based on the ratio between the luminance of the output-screen and the luminance of a fluoroscopic screen, but they were measured at different kilovoltages, wave forms and total filtrations. Moreover, there was a variation in the speed of the fluoroscopic screens of a given type. At first glance such definitions appeared to be quite reasonable and were usually used without taking these variables into account.

Three classes of techniques with which x-ray image intensification must be compared are fluoroscopy, radiography with intensifying screens, and cinefluorography. It has been difficult for the radiologist to evaluate the intensification factor of an x-ray image intensifier against any of these techniques.

To avoid a misunderstanding in the statement of the intensification factor, it is recommended that the intensifying systems be evaluated by the quotient of the luminance of the output screen to the input exposure rate. This quotient is called the "conversion factor". (In French, "facteur de conversion"; in German, "Conversionsfaktor".) This term can be used for the conversion of the input exposure rate or exposure into any kind of signal, whether it be a luminance, a film density or a signal output current as in case of scanning systems. Although all of these types of conversion factors must eventually be defined, the first attempt is restricted to the electronic x-ray image intensifier.

It is known that the x-ray quality is not defined exactly by a statement of kilovoltage and filtration alone. The type of apparatus, voltage waveform, variation in radiation output of the x-ray tube, etc., may give rise to differences in quality that cannot be neglected. As the conversion factor of the image intensifier may vary with the radiation quality, it is advisable to recommend a specified radiation quality, at which the conversion factor should be measured. Therefore, it is arbitrarily recommended that an x-ray beam having a HVL of 7 mm Al be used for the measurement of the conversion factor. This HVL is produced with a total filtration of 22 mm Al and a peak kilovoltage of approximately 85 kv full-wave rectified or constant potential.

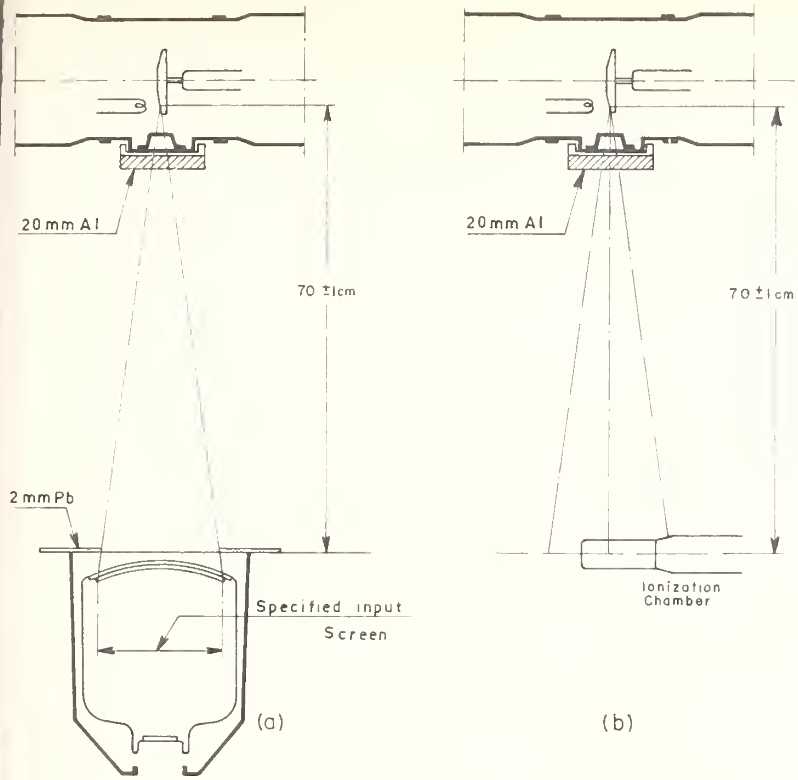

FIGURE II.6. *Illustration of the arrangement for measurement of a conversion factor.*

a. With image intensifier in the beam.
b. With an exposure meter in the beam.

D. Conversion Factor of X-Ray Image Intensifiers

The "conversion factor", symbol G_x, of an x-ray image intensifier is the ratio of the luminance of the output screen to the input exposure rate of the applied x radiation. The luminance is to be expressed in candela per square meter (cd/m^2) and the input exposure rate in milliroentgens per second (mR/s). The conversion factor is to be expressed in $\mathrm{cd \cdot m^{-2} \cdot (mR)^{-1} \cdot s}$. The data on luminance of the output screen should be accompanied by the relative spectral energy distribution curve of the fluorescent light emitted by the output screen.

E. Recommendations for Measurement of Conversion Factor

1. *X-ray source:* Diagnostic x-ray beam after a filtration of 2 mm Al.
2. *Added filtration:* 20 ± 0.5 mm Al. The added filter should be placed in the x-ray beam and close to the x-ray housing. (See fig. II.6.)
3. *Kilovoltage:* The kilovoltage should be such that the HVL of the radiation transmitted by this filter is 7 ± 0.2 mm Al. Fullwave rectified or constant potential may be used.
4. *Focus-image intensifier distance:* The distance between the focal spot of the x-ray tube and the entrance plane of the x-ray image intensifier should be 70 ± 1 cm.
5. *Irradiated field:* Only the input screen are specified by the manufacturer should be irradiated. A field limiting diaphragm as indicated in figure II.6a should be used.
6. *Exposure rate:* To be adjusted by the tube current to 1 ± 0.1 mR/s, measured free-in-air. The focus-chamber distance should be 70 ± 1 cm. (See fig. II.6(b).) The exposure meter should be calibrated for the specified radiation quality.
7. *Luminance of the output screen:* The luminance of the output screen should be measured at the center of the output screen with the image intensifier electron optically adjusted. The measurement of the luminance of the output-screen of an x-ray image intensifier can be carried out with any instrument [1] as long as the relative spectral energy sensitivity curve of the receptor matches the photometric standard for photopic vision.

[1] The luminance of the output screen of an x-ray image intensifier can be measured in any of the following ways:

(a) By means of a direct-reading instrument for which the receptor has a relative spectral energy sensitivity curve which matches the photometric standard for photopic vision. The accuracy of the measurement is given by the accuracy of the instrument itself.

(b) By comparison with the luminance in a photometer. The photometer need not have exactly the same energy distribution as the output screen. The measuring accuracy is very dependent on the deviation of both distribution curves, since it is difficult for the observer to compare two luminances of different relative spectral distribution.

(c) By means of a direct reading instrument with a receptor of an arbitrary relative spectral sensitivity. For this purpose the receptor has to be calibrated by using a reference output screen for which the luminance is known and which has a relative spectral distribution equal to that of the output screen of the x-ray image intensifier. Values obtained from such a measurement should only be used to compare the conversion efficiencies of output screens having the same relative spectral distribution.

The use of either method (a) or (b) is recommended.

III. Measurement of the Characteristics of Grids

A. Preferred Terms and Symbols [2]

Alphabetical order is not considered to be necessary nor opportune.

In *calculations* and *formulas* use of the *metric* system is recommended.

1. *Scattered radiation:* Not: secondary radiation.

2. *Grid, roentgengrid, x-ray, and scattered-ray grid:* The term "Grid" alone may be used when it is obvious that the well-known radiographic device is meant. Not: diaphragm (see item 11). Not: Bucky grid or Potter-Bucky grid, in order to avoid confusion with item 11. Not: wafer grid, in order to avoid confusion with crossed grids and thin grids.

3. *Stationary grid and moving grid:* Stationary (not fixed) grids are used in front of, and touching the cassette or may comprise the front face of a cassette.

Moving grids are mounted in a casing with a driving mechanism (see item 11).

4. *Linear grid:* X-ray grid composed of plane strips which are parallel in the direction of their largest dimension.

It is *focused* when the planes of the strips converge to a line parallel to the grid surface. It is called parallel when these planes are parallel.

5. *Cross grid:* Two linear grids built together in such a way that the directions of their strips form an angle. When the angle is 90°, the cross grid is called orthogonal.

6a. *Central line:* The line of a focused linear grid, where the strip is perpendicular to the grid surface. In a parallel grid the center line may mark the geometrical center.

6b. *Central point:* The point of a focused cross grid where the strips are perpendicular to the grid surface. In a parallel grid, the central point may mark the geometrical center.

7a. *Convergence line:* Virtual line of convergence of the planes of all strips of a focused linear grid.

7b. *Convergence point:* Virtual point of convergence of the planes of all strips of a focused cross grid.

8. *Focusing distance:* Distance between the convergence line or point and the grid surface. Not: focal or focus distance, in order to avoid confusion with focus-grid distance (see also item 9).

Symbol: f_0.

A parallel grid is indicated by $f_0=\infty$

Unit: cm (or in.).

9. *Focus-grid distance:* Not: focal or focus distance.

Perpendicular distance of the tube focus to the grid surface.

Symbol: f

Unit: cm (or in.).

10. *Focus-grid distance limits:* Limits between which the focus-grid distance can be varied without losses of primary radiation in excess of those specified in section C.

Symbols: f_1 and f_2 (lower and upper limit).

11. *Bucky and Potter-Bucky:* Radiographic accessory unit; apparatus including a driving mechanism to move the grid. Not: Bucky diaphragm, because the word diaphragm is used for primary diaphragms etc.

12. *Strips:* Not: lines (see item 17).

It is not necessary to add the prefix "lead", but if another absorbing material is meant, this is to be added.

13. *Interspace, interspace material:* The space and the material between the strips.

14. *Height of the strips:*

Symbol: h

Unit: cm (or in.).

15. *Thickness of the strips:*

Symbol: d

Unit: cm (or in.) (mm, micrometer =micron, or mill=0.001 in.).

16. *Distance between the strips:*

Symbol: D

Unit: cm (or in.).

17. *Number of strips:* Not: number of lines, in order to avoid confusion with the indications used in optics and TV, where the "black" and "white" lines are counted together.

Symbol: $N=\frac{1}{D+d}$ per cm (or in.)

Not: l or L per cm or in.

18. *Grid ratio:* Ratio alone may be used when it is obvious that the radiographic ratio is meant.

Symbol: $r=\frac{h}{D}$, not $\frac{h}{D+d}$

Write: $r=10$, not 10:1 or 1:10.

[2] French and German translations of this subsection are available from the ICRU Secretariat, NBS, Washington, D.C., 20234.

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The ratio of a cross grid composed of two equal grids, each with $r=6$, is to be indicated by $r=2\times 6$.

19. *Lead content:* Mass, not volume, of the lead per unit surface area.

Symbol: P.

When an absorbing material other than lead is used, this is to be indicated.

Unit: g/cm^2.

20. *Transmission:* The ratio of I' to I where I' is the luminance produced by an intensifying screen after the x-rays have passed through the grid and I the luminance produced when no grid is present.

Symbol: $T=I'/I$

Unit: fraction

$T_p=I'_p/I_p$=transmission of primary radiation

$T_s=I'_s/I_s$=transmission of scattered radiation

$T_t=I'_t/I_t$=transmission of total radiation.

For the sake of simplicity and in order to avoid confusion the values $1-T_p=V$=loss factor and $1-T_s=\eta$=efficiency="clean-up", should not be used.

21. *Bucky factor:* Total incident radiation divided by total transmitted radiation.

$$\text{Symbol: } B=I_t/I'_t=\frac{1}{T_t}.$$

Equivalent to the reciprocal of the transmission of the total radiation.

22. *Selectivity:* Transmission of primary radiation, divided by transmission of scattered radiation.

Symbol: $\Sigma=T_p/T_s$.

23. *Contrast improvement factor:* This factor is the ratio of the x-ray contrast *with* grid divided by the x-ray contrast *without* grid.

Symbol: $K=T_pB=T_p/T_t$.

24. *Index, n:* To be used as a subscript for factors measured under the following conditions: 100 kv d.c., standardized x-ray source and water phantom of area 30x30 cm and 20 cm thick, as specified in section B.

B. Measuring Method

To obtain comparable measurements on grids by a procedure which can be performed in any place in the world, a photoelectric method is recommended because of its sensitivity and reproducibility. The results of such measurements may, if desired, be translated into photographic units (1).

1. *Phantom:* A water phantom having an area 30x30 cm and with a thickness of 20 cm should be used. When a range of thicknesses is desired 10, 20, and 30 cm are recommended.

The container may be made of acrylic plastic, Plexiglass, Perspex, Lucite, polyethylene or any material not containing heavier elements, such as sulphur or chlorine, e.g., polyvinylchloride. The thickness of the bottom should not exceed 1 cm.

2. *X-ray Source:* The x-ray tube should have a total (inherent + added) filtration of approximately 2 mm aluminum at 60 kv, 4 mm Al at 100 kv, and 6 mm Al at 200 kv. The high voltage applied to the tube should have a ripple of less than 10 percent. Constant potential is recommended in order to avoid discrepancies due to different voltage wave form. Since for these measurements only small tube currents are required, a constant-potential therapy apparatus, or a four valve generator with adequate voltage smoothing by condenser or H.T. cables may be used. Peak-voltage variations during measurement should not exceed ± 1 percent.

Electronic input stabilization is recommended; if this is not available a voltmeter should be used for monitoring the voltage, or, still better, a compensation recorder showing clearly the fluctuations of the primary radiation intensity falling on to the grid. A sphere gap can be used to calibrate the kv indicator in the equipment.

In cases where only one voltage value is to be used, 100 kv is recommended.

Other voltages which may be used for reference are 60, 75, 125, 150, and 200 kv.

3. *Measuring Arrangement:* As the radiation detector, a fluorescent screen is to be used in combination with a photomultiplier, the size of this screen being approximately 20 mm in diameter.

To achieve the same radiation quality dependence as in practice, a calcium tungstate screen ("medium speed", intensifying "back" screen, 70–90 mg/cm^2), backed with 0.5 mm aluminum for electron absorption, should be used. The multiplier phototube may be of any type yielding at least 20 amperes per lumen (such as RCA 931A). It is desirable to choose a phototube the spectral sensitivity of which has a maximum in the region of greatest light output from the calcium tungstate screen. Voltage stabilization of the multiplier phototube within 0.1 percent is necessary.

The tube focus should be placed 100 cm perpendicularly above the center of the grid. This distance is independent of the focusing distance, f_0, as only the radiation coming through the central part of the grid is measured. The diameter of the x-ray beam at the grid surface should be 36 cm except when measuring primary radiation.

The grid should be placed close to the bottom of the phantom. The screen to grid distance is 2 cm. The distance between multiplier phototube and screen is not critical provided that the tube is shielded from unwanted radiation, e.g., by

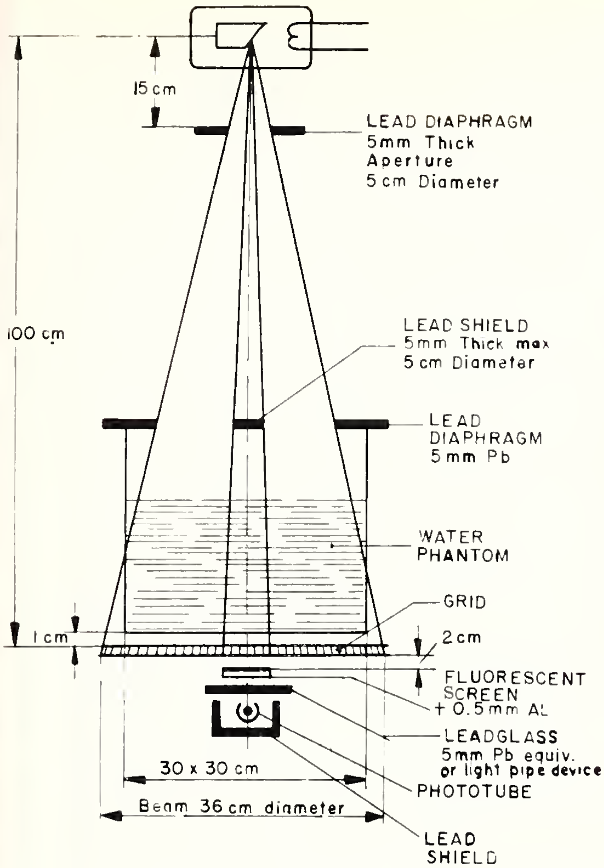

FIGURE III.1. *Setup for measuring scattered and total radiation.*

a.= Grid to be measured.
b.= Light-tight 0.5 mm thick aluminum cover.
c.= Calcium tungstate, intensifying screen 2 cm diameter.
d.= 5 mm thick lead diaphragm with 2 cm diameter aperture.
e.= Light-tight, 5 mm thick lead lined box.
f.= Lead-glass, 5 mm Pb equiv.
g.= Photomultiplier tube with minimal noise and with a minimum sensitivity of 20 am/lm.
h.= Tube holder with incorporated voltage divider.

using lead glass between screen and multiplier tube as shown in figure III.1, or by a light-pipe device.

The scattered radiation alone is measured when the multiplier and fluorescent screen are shielded from all primary radiation. The total radiation is measured when this shield is removed (fig. III.1). To improve the measurement accuracy when determining scattered radiation, it is recommended that the diameter of the lead shield be decreased stepwise from 5 cm and extrapolated to zero.

When the primary radiation incident on the grid is to be measured, the same set-up can be used with appropriate diaphragms in order to obtain a narrow beam. Measurement inaccuracies may be caused, however, by small amounts of scattered rays from the water phantom. Therefore it is recommended that the measurements be carried out as indicated in figure III.2.

An illustration of the arrangement of equipment described here is shown in figure III.3.

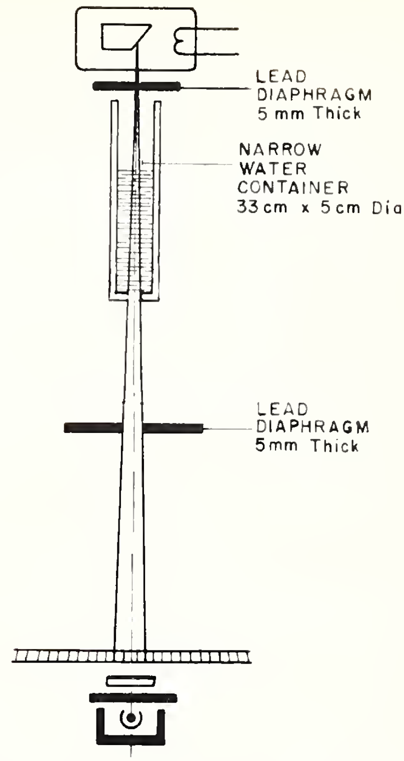

FIGURE III.2. *Setup for measuring primary radiation.*

4. *Measurements:* Using the arrangement described, the following values (see section A) can be measured with and without grid:

a. *Primary radiation:* I_p and I'_p without and with grid, respectively.

b. *Total radiation:* I_t and I'_t without and with grid, respectively.

c. *Scattered radiation:* I_s and I'_s without and with grid, respectively.

5. *Derived values:* From these measured values the following values can be calculated:

Transmission factors

$$T_p = I'_p/I_p;\ T_s = I'_s/I_s;\ T_t = I'_t/I_t.$$

Bucky factor

$$B = I_t/I'_t.$$

Selectivity

$$\Sigma = T_p/T_s.$$

Contrast improvement factor

$$K = T_p/T_t.$$

6. *Standard check:* To check the grid transmission-measuring-equipment an electrolytic-copper filter of thickness equal to 0.89 ± 0.01 g/cm^2 (i.e., approximately 1 mm) is recommended. This filter is used in place of the grid in the arrangement shown in figure III.2. The distance between the filter and screen is then 2 cm. With such an

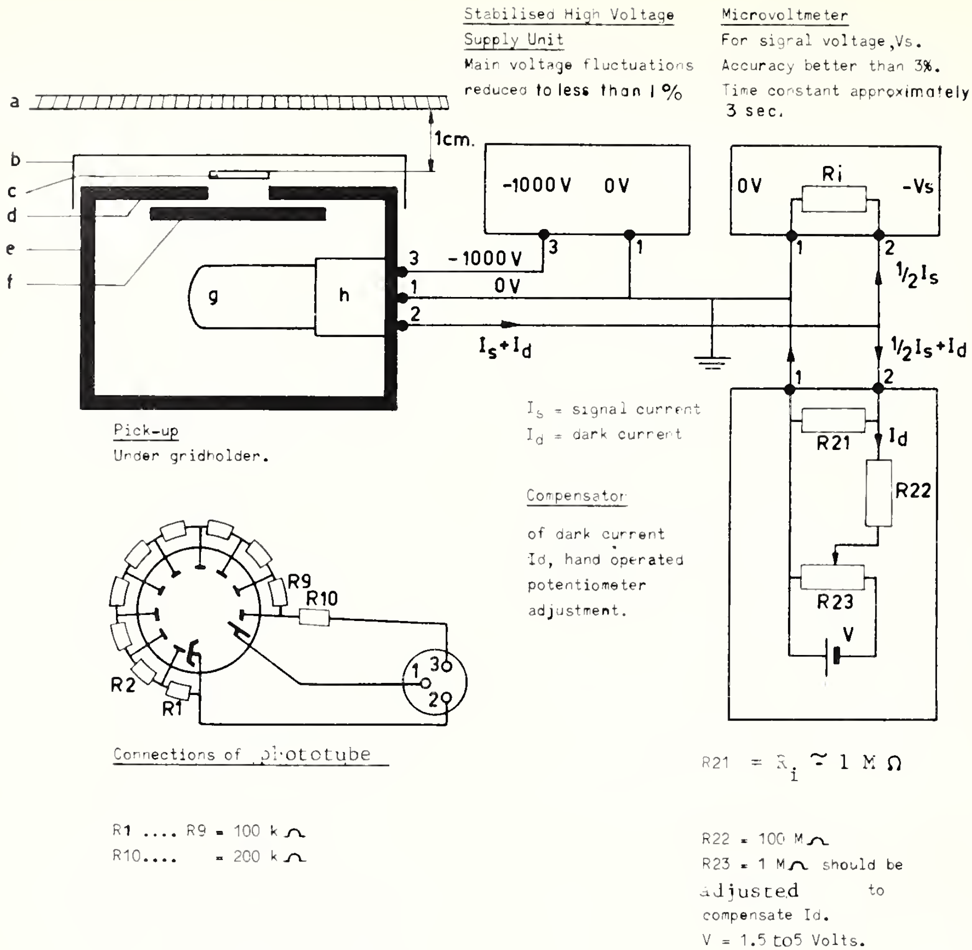

FIGURE III.3. *Grid transmission measuring equipment.*

arrangement the transmission of this copper filter should fall within the limits given below:

Tube potential kv	*Transmission limits* %
60	6½– 7
75	15½–17
100	33–37
125	42–46
150	48–53

Transmissions outside of these limits indicate a defect in the measuring equipment or in the voltage calibration.

C. Grid Characteristics

The performance of a grid is determined by the combined effect of various physical factors, most of which are mutually dependent, whereas the choice of the most important among them depend on the widely diverging exposure conditions in medical practice.

ICRU recommends that for indicating these factors, the terms and symbols given in section A of these recommendations and measurements according to the method described in part B should be used.

The terms to be considered for this purpose are as follows:

N=number of strips per cm (in.)
r=grid ratio
f_1–f_2=focus-grid distance limits
P=lead content
T_p=transmission of primary radiation
T_s=transmission of scattered radiation
T_t=transmission of total radiation
B=Bucky factor=$1/T_t$
Σ=selectivity=T_p/T_s
K=contrast improvement factor T_p/T_t.

The significance of these factors, in so far as their influence on the quality of the radiographic image is concerned is as follows:

N= *The number of strips per cm* (in.) is important for the radiographic result. A small value of N is generally associated with a more favorable combination of the last eight factors given above.

When the grid is used either stationary, or moving in a Potter-Bucky that does not completely fade the shadows of the strips under all exposure conditions, a larger value of N will reduce the visibility of their shadows and so improve the *perceptibility in the radiographs.*

The following factors all deal with the more restricted, but important quality discrimination; viz, the *contrast improving capacity of the grids.*

r=The *grid ratio* is often considered as a quality factor in the above sense. However, this holds good only when comparing grids with approximately the same value of N(2). Of two (efficiently designed) grids with the same strip material and the same ratio r but different N, the one with the lowest value of N may have a better combination of quality factors.

f_1-f_2=The *focus-grid distance limits* are closely connected with the ratio and also dependent on the focusing distance (3). A deviation of the focus-grid distance, f, from the focusing distance, f_0, causes a decrease of the transmission of the primary radiation to the sides of the grid center and so less blackening and lower contrast at the image edges. A high ratio is unsuitable when the conditions of the x-ray examination do not permit exact positioning of the focus relative to the grid.

The limits are so defined, using the nominal or stated values of "f_0" and "r" and an average lateral decentering of 1 cm of the tube focus, that a loss of 50 percent (for cross grids 25 percent) is obtained at 15 cm from the central line of the grid.

The (nominal) focus-grid distance limits f_1 and f_2 are calculated for the above standard conditions; they serve only for general information and comparison. For field widths smaller than 30 cm wider focus-grid distance limits can be applied or a higher ratio be chosen.

P=For "efficiently" designed grids, that is for grids where a fairly optimum transmission of the primary radiation is obtained, the *lead content* (or the equivalent mass per unit surface of another absorbing material (4, 5)) is a rough but easily determinable measure of the contrast improvement of the grid. Therefore one can speak of "light" and "heavy" grids when groups of small and large contrast factors are concerned (6). An extreme example of the opposite of an efficiently designed grid is a lead sheet, which may have a large lead content but is a very inefficient selective absorber of scattered radiation.

T_p=A high *transmission* (low absorption) of *primary radiation* is advantageous. The greater this factor, the smaller is the Bucky factor.

T_s=A low value of the *transmission of scattered radiation* is important as it indicates the degree of removal of this undesirable part of the total radiation.

T_t=The *transmission of the total radiation* is of importance since the higher the transmission the shorter the exposure time. Moreover, it influences (reciprocally) the value of the contrast improvement factor, K.

B=The *Bucky factor* is the reciprocal value of T_t and therefore of the same importance.

These last two factors vary markedly with field size, object thickness and voltage (7).

Σ=The *selectivity* is the quotient of T_p and T_s. A high value of this ratio is advantageous, as a large T_t and a small T_s are desirable.

K=The *contrast improvement factor* is a direct measure of the maximum obtainable contrast *with* grid divided by the corresponding contrast *without* grid. Therefore it is an important factor for comparing grids in *medical practice* as far as their contrast improving capacity is concerned. There is a close connection between K and B as $K=T_p/T_t=T_pB$ and T_p is only slightly dependent on the voltage (8).

Where K is specified it should be observed that this value is by no means an evaluation of the contrast improving capacity of the grid under all exposure conditions. As a matter of fact the factor K is dependent on object thickness, field size and voltage. It may even be that in low density parts of the image a grid with a lower value of K or no grid at all may give a better contrast.

It may be of scientific interest to express the factors B, Σ, and K as a function of various exposure conditions. However, for *comparison* of grids, it is hardly necessary to give so much information; generally *one single exposure condition* will do for this purpose.

When only *one value* of each quality factor is to be given this should be determined at 100 kv, with a 20-cm thick water phantom and with the arrangement shown in figures III.1 and III.2. In this case, the corresponding symbols shall include the subscript n, e.g., K_n.

D. Limitations in Use of Measured Grid Characteristics

In relating measured grid characteristics to the quality of the medical radiographic image the following limitations must be kept in mind:

1. It is assumed that no scattered radiation is produced below the grid.

2. All factors are determined for the *center of the grid*, and for a focus-grid distance of 100 cm.

3. Maximum contrast improvement is only obtained when all the image elements concerned are in the *linear part of the characteristic film blackening curve*. In the toe of this curve, corresponding to weakly exposed image details, better contrast may be obtained with grids having a lower contrast improving capacity, and even without a grid at all (9).

4. The factors T_s, T_t, B, Σ, and K are determined with a simple, rectangular *water phantom*, 20x30 cm wide, which is quite different from any medical object. It is necessary to base the grid characteristics on such a phantom because this is the only way to obtain reproducible and comparable values. With a narrower phantom (object) and/or with a narrower x-ray beam other values will be found.

As a result of these limitations the values of the factors given in grid specifications are suitable for *comparison* of grids and for the *choice* of their field of application and should not be used for calculation of actual contrast improvement, etc., in routine medical radiography.

E. Grid Labeling and Specifications

To permit rapid selection of the grids having characteristics suitable for their task it is recommended that manufacturers use the terms, symbols and units given in section A. For the same reason it is recommended that grid characteristics be measured in the manner described in section B. It is also desirable that grid marking and specifications be uniform in order to provide useful information to the user.

a. *Grid labeling:*

For labeling grids, the following items, related to the requirements of application, are recommended:

N=number of strips per cm (in.)
f_0=focusing distance
f_1-f_2=focus grid distance limits
f_0, f_1 and f_2 to be rounded off to multiples of 5 cm (2 in.) up to 75 cm (30 in.), to multiples of 10 cm (4 in.) above 75 cm.

The Bucky factor, B_n, is not recommended for grid labeling due to its great dependence on exposure conditions.

It is recommended that f_0, f_1, and f_2 be stated in both cm and inches and that the tube side of the grid be indicated by a diagrammatic image of an X-ray tube. The central line should also be marked on the tube side of the grid. For cross grids the central point should be indicated by a cross.

b. *Grid specifications:*

For grid specifications the same items as recommended for grid labeling should be included. In addition K_n should be specified. All other items included in section A, such as B, r, T_p, and T_s may be given. If additional specifications for other than the exposure conditions recommended by item 24 of section A are given, e.g., various voltages and phantom thicknesses, then the values recommended in section B should be used. In order to avoid misinterpretation of the given values, the limitations mentioned in section D could be stressed. For example, it could be explained that for narrower beams the focus grid distance limits are considerably wider than the specified values f_1-f_2. The given values are suitable for comparison and choice in view of their application but they do not express strict limits nor absolute quality values applicable to the numerous and diverging conditions of medical practice.

F. References

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IV. Measurements of the Characteristics of Body-Section Radiographic Equipment

A. Symbols and Abbreviations

For the purpose of this report, the following will be used:

A=focus-film distance, FFD.
D=focus-plane distance; focus-objective plane distance; long arm of lever.
d=plane-film distance; objective plane-film distance; short arm of lever.
f=film; recording medium.
L=lever, or equivalent means of moving tube and film; ratio=D/d.
OP=objective plane.
E=fulcrum, or object axis.
e=film axis.
F=focus of x-ray tube.
h=section "thickness", section interval, section depth, or depth of visibility.
S=an absolute or arbitrary factor for calculating h; the movement of a redundant image on the film.
S_{eff}=efficiency coefficient.
B=diameter in the objective plane.
M=magnification.
T=tube shift; movement of focus during exposure.
α=exposure angle; angle through which central ray moves during exposure; angle through which film and object rotate during exposure.
CR=central ray, or optical axis of beam.
I=separation of planes in the body.
i=distance between film and mechanical axis of film; separation of films in multisection.
j=connecting link between object axis and mechanical axis of film in fixed or moving tube systems.
U=total unsharpness.
U_g=unsharpness due to non-point source.
U_m=unsharpness due to movement of subject during exposure.
U_f=unsharpness due to recording medium.
U_x=unsharpness due to tomographic movement.
C=distance moved by film, or angular movement of film.
t=time.
v=velocity.
ma=tube current.
mas=milliampere-seconds.

B. Nomenclature

It is thought desirable to adopt one term to describe all types of body section technique as all techniques are bound by the same principles. The term TOMOGRAPHY is now so universally used and established that it has been chosen, although PLANIGRAPHY and STRATIGRAPHY have priority, chronologically and historically. A radiograph of a body section is called a TOMOGRAM.

1. *Apparatus:*

There are now so many different designs and makes of tomographic apparatus that it is not feasible to refer to each one by name. The list would seldom be complete as new apparatus is introduced each year. Systems are therefore listed according to the methods involved.

2. *Movements:*

a. *Unidirectional:* the system moves in such a way that the "central ray" moves in a single plane.

b. *Pluridirectional:* the system moves in such a way that the "central ray" moves in more than one plane.

3. *Moving tube systems:*

a. *Rectilinear:* tube and film move in a straight line and movement can be described as plane-parallel (fig. IV.1).

b. *Curvilinear:* tube and film move in an arc within a given plane (fig. IV.2).

c. *Pluridirectional:* tube and film move in a circular, elliptic, hypocycloidal, spiral, sinusoidal, or random path (fig. IV.3a).

d. *Perpendicular:* tube and film move in opposite directions on a line substantially perpendicular to the objective plane (fig. IV.3b). Movement need not be truly perpendicular (right hand diagram for fig. IV.3b).

4. *Fixed focus systems:*

a. Object and film rotates on separate vertical axes; equivalent to a curvilinear movement (fig. IV.4a).

b. Object and film rotate on separate horizontal axes; equivalent to a curvilinear movement (fig. IV.4b).

c. Object and film rotate on separate vertical axes; degree of rotation unlimited (fig. IV.4c).

d. Object and film gyrate on separate vertical axes; object and film remain parallel to fixed reference line (fig. IV.4d).

e. Same as c but in opposite directions (fig. IV.4e).

5. *Parts of apparatus:*

a. *Angle switches:* exposure switch, or switches, to determine angle of exposure: tomographic switches.

b. *Fulcrum:* pivot about which lever rotates; object axis; can be adjustable or fixed; the equivalent point in a stationary tube system.

c. *Film driving pin:* film axis; means of connecting lever to film carrier; can be adjustable for scale correction, or for selection of a plane.

d. *Lever:* usually of the first order; couples tube and film carrier to ensure correct proportional movement: long arm=focus to fulcrum, short arm=fulcrum to film axis: lever is of fixed

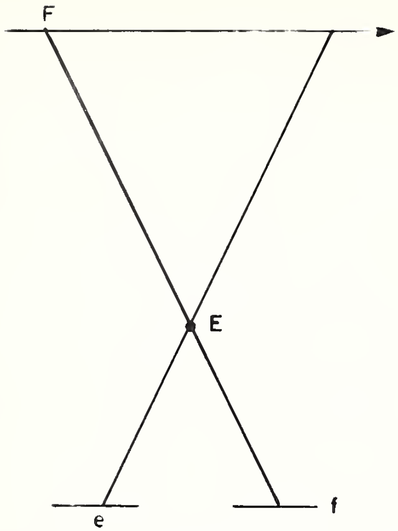

FIGURE IV.1. *Diagram illustrating rectilinear moving tube system.*

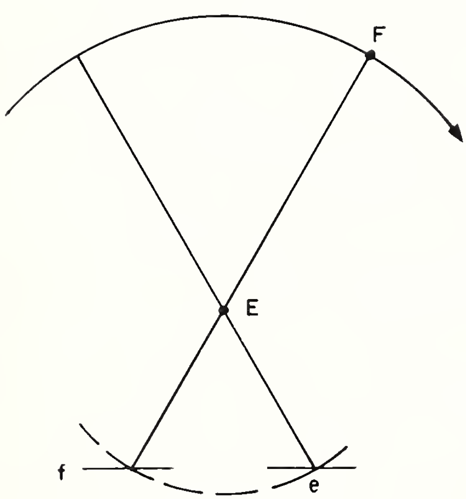

FIGURE IV.2. *Diagram illustrating curvilinear moving tube system.*

length for curvilinear movement, and telescopic for plane parallel movements.

e. *Scale:* tomographic, a means for indicating height of fulcrum above a given level: indicates site of objective plane in relation to resting plane or object support.

f. *Plane indicator:* pointer or light beam to indicate site and position of objective plane.

g. *Tube axis:* point about which x-ray tube rotates: not necessarily on the tube focus.

6. *Technique:*

a. *Sequential:* tomograms taken in sequence, one at a time.

b. *Simultaneous multisection:* two or more films exposed at the same time to produce a different body level on each film.

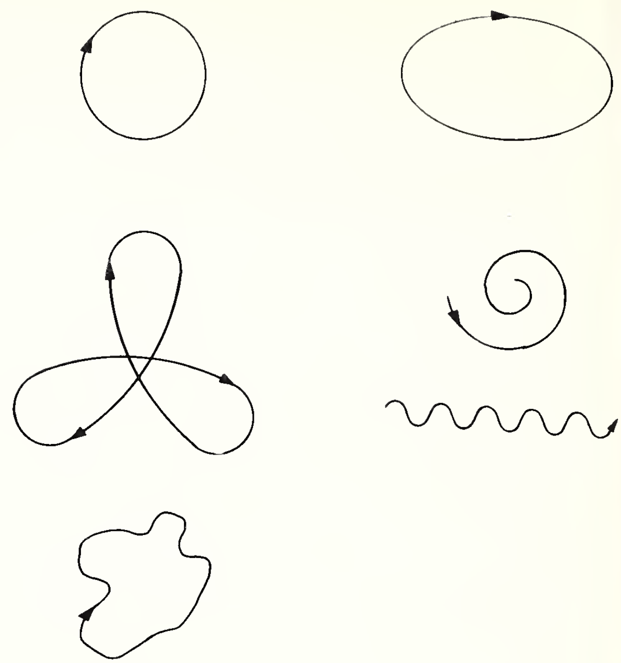

FIGURE IV.3a. *Diagram illustrating circular, elliptical, hypocycloidal, spiral, sinusoidal or random path of moving tube of film.*

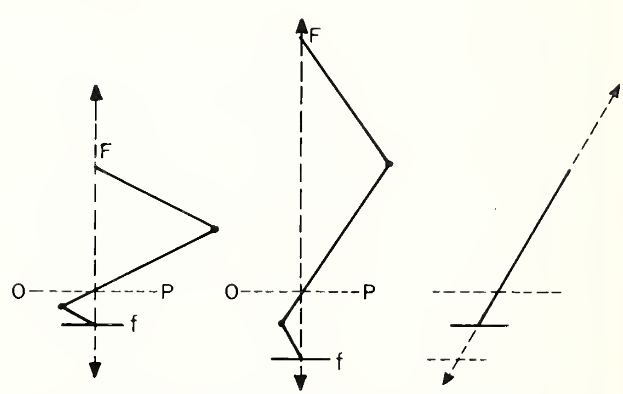

FIGURE IV.3b. *Diagram illustrating perpendicular moving tube system.*

c. *Lever ratio:* focus-plane to plane-film distance: focus-fulcrum to fulcrum-film distance: tomographic ratio.

d. *Objective plane:* a preselected plane: a plane whose shadow is stationary relative to the film. It is permissible to call a curved surface an objective "plane" when recorded on a curved film.

e. *Film plane:* the surface of a film.

f. *Section:* a slice or layer of the body as recorded in a tomogram.

g. *Focus-film distance:* distance between the tube focus and the film.

h. *Focus-plane distance:* distance between the focus and an objective plane.

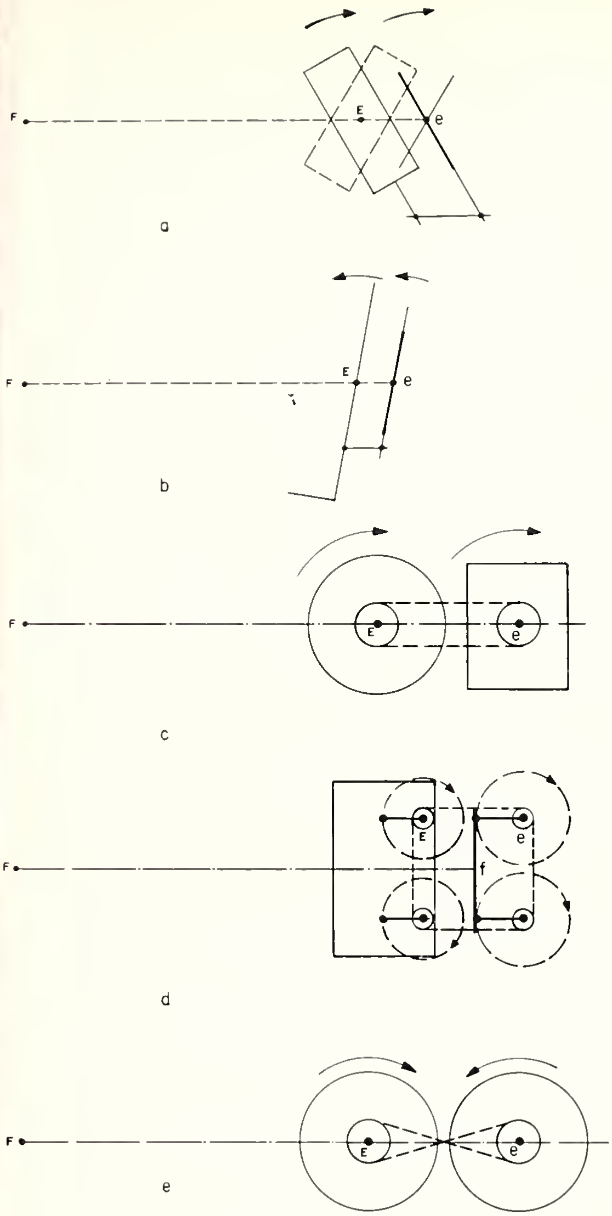

FIGURE IV.4. *Fixed focus systems.*

a. Object and film rotate on separate vertical axes.
b. Object and film rotate on separate horizontal axes.
c. Object and film rotate in same direction on separate vertical axes.
d. Object and film gyrate on separate vertical axes but object and film remain parallel to fixed reference line.
e. Object and film rotate in opposite direction or separate vertical axes.

i. *Plane-film distance:* distance between objective plane and the film.

j. *Magnification:* enlargement of image relative to the objective plane.

k. *Section "thickness":* the effective thickness of a section determined by an absolute or an arbitrary factor. The term *thickness* alone is not recommended.

l. *Section interval:* distance between one objective plane and the next: calculated in the same way as for section "thickness".

m. *Selectivity:* refers to the degree of elimination of redundant shadows: inversely proportional to section "thickness": can be calculated in the same way as for section interval. (See section E.)

n. *Efficiency coefficient:* the excursion of a shadow on the film related to a point 1 cm above the objective plane: determines section intervals. (See section E.)

o. *Redundant shadows:* unwanted images of planes above and below the objective plane.

p. *Interspace:* space between one objective plane and another; space between one film and another.

q. *Intersection:* the line where a film intersects an objective plane when the film is not parallel to it: not to be used to describe interspace. (See fig. IV.6.)

r. *Film:* in the text the recording medium is always referred to as film.

s. *Film axis:* the axis on which a film, or films, rotate: can be adjacent and coaxial with film, or remote and at an angle to film

t. *Patient axis:* object axis: axis on which patient rotates: can be adjacent or remote.

u. *Operative angle:* the angle through which the apparatus moves: not the exposure angle.

v. *Exposure angle:* the angle through which the x-ray beam (or central ray) moves during exposure: the equivalent movement of object and film with a fixed tube system.

w. *Exposure:* submitting patient and film to x radiation for a predetermined time.

x. *Localization:* determination of depth of a plane or situation of a pathological region, either before or by means of tomography.

y. *Focusing:* setting the fulcrum, or the equivalent means, to cause a particular plane, or section, to be recorded.

z. *Stereotomography:* process of taking a pair of sterographic tomograms.

aa. *Orthotomography:* tomographic system for recording an image, by means of a narrow beam of X-rays substantially perpendicular to the objective plane.

bb. *Zonography:* tomography with a very small exposure angle.

cc. *Macrotomography:* tomography with a fine focus tube and a magnification of the order of 2 to 5.

dd. *Microtomography:* tomography with a fine focus tube and subsequent optical magnification of the order of 20.

7. *Multisection:*

a. *Multisections:* a series of tomograms recorded on two or more films in one exposure.

b. *Multisection radiography:* multitomography: the taking of multisections: *not* "multisection tomography".

c. *Multisection cassette:* a means for holding two or more films and screens in spatial relation, suitable for multisection radiography.

d. *Multisection magazine:* a light-tight box for holding more than three films with intensifying-screens in spatial relation: sometimes called box or container.

e. *Multisection screens:* a series of intensifying screens with a graded sensitivity to give equal density on a series of films in one exposure.

f. *Interspacing material:* sheets of x-ray transparent material of a specified thickness for maintaining film spacing.

g. *Magnification:* enlargement of an objective plane: determined by lever ratio, or ratio of plane-film to focus-plane distance.

h. *Constant magnification:* enlargement is constant when lever ratio is constant, as in multisections recorded in one exposure.

i. *Rapid serial multisections:* tomograms taken in rapid succession of different sections of the body.

j. *Stereosynthesis:* reconstruction of the third dimension by the superposition of a series of tomograms.

C. Exposure Angle

Exposure angle should be capable of exact repetition, and be predetermined by the setting of switches operated by the tomographic movement.

a. For rectilinear motion the switches should be independently adjustable so as to produce an unsymmetrical exposure angle when required (fig. IV.5). Exposure angle should vary from 5 degrees up to the maximum possible.

b. Pluridirectional systems should have a switch, or switches, to ensure a complete cycle movement, or a multiple of the cycle.

c. Fixed tube systems require the same type of switching as in a.

d. Apparatus for axial transverse tomography should have switches to ensure at least a complete 360 degree exposure.

1. *Testing exposure angle:*

With the switching devices mentioned above, the desired angle can be preset to a scale, but it is advisable to test the exposure angle and compare with the scale markings. The test should also show whether the exposure angle is symmetrical or not. A repeat test for consistency is advisable. A universal method is described under D.1.

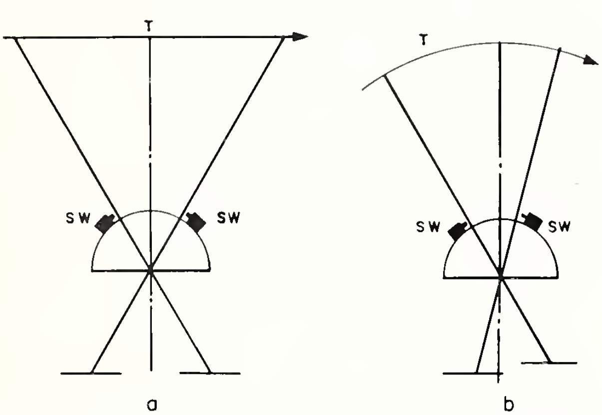

FIGURE IV.5. *Tomographic switches set for (a) symmetrical and (b) unsymmetrical exposure angles for rectilinear or curvilinear motion tomography.*

a. *Rectilinear systems:*

A non-screen film is placed directly under the focus of the x-ray tube. The film should be nearly perpendicular to the objective plane, parallel to the direction of movement, and so placed as to intersect the objective plane. An inclination of not more than 15 degrees is satisfactory (fig. IV.6). Alternatively, the film can be perpendicular and slightly off center.

A slit diaphragm 1 mm wide is fitted to the tube at a distance of 40 cm from the focus, the slit being at right angles to the direction of movement. A correspondingly narrower slit at a shorter distance, such as with the conventional light-beam diaphragm, is also suitable.

A tomographic exposure will record the exposure angle as film blackening in the shape of two opposed triangles (fig. IV.7). Unless the mechanism is perfect and there is no penumbra, the apices of the triangle will be cut off. The angle can be measured by first extending the sides of one triangle to form an apex, figure IV.8, or by measuring the obtuse angle and subtracting from 180 (fig. IV.9). From this test the symmetry or asymmetry of the exposure angle can be seen (fig. IV.10).

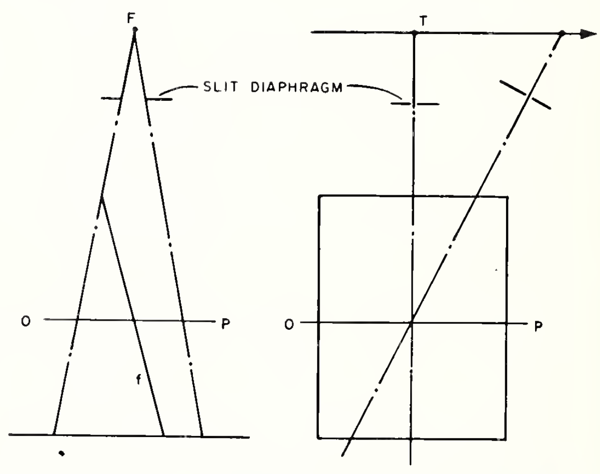

FIGURE IV.6. *Arrangement for testing exposure angle of rectilinear system.*

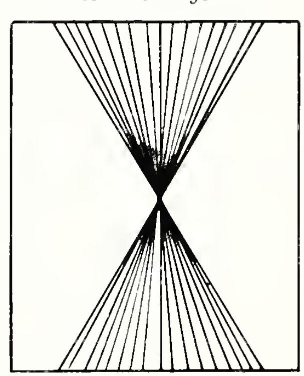

FIGURE IV.7. *Film record of test conducted by arrangement shown in figure IV.6.*

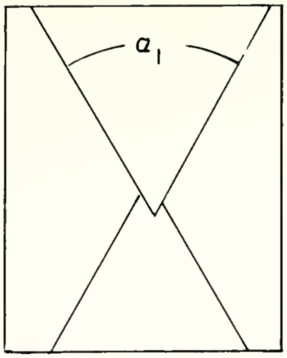

Figure IV.8. *Exposure angle measured from film record of figure IV.7.*

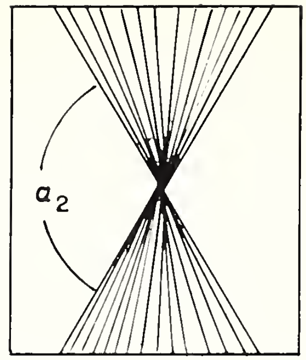

Figure IV.9. *Alternate method of determining exposure angle,* $\alpha_1(=180-\alpha_2)$.

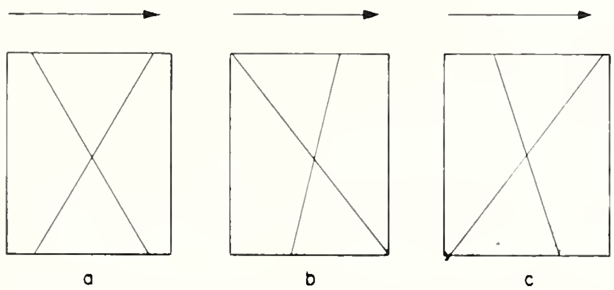

Figure IV.10. *Film record showing (a) symmetrical, (b) advanced unsymmetrical, and (c) retarded unsymmetrical motion indicated by tests conducted with arrangement shown in figure IV.6.*

b. *Pluridirectional systems:*

A moving tube circular motion can be tested for exposure angle by the single film method as in C.1.a. Proof that the movement is truly circular requires a second exposure on another film at right angles to the first.

The two angles of an elliptic movement can be obtained by the same two-film method. The more complex movements require more than two films.

An alternative is to use two films parallel to the objective plane, one above and one below, separated by a known distance. A pinhole diaphragm with 1 mm diameter aperture is fitted to the tube at 40 cm distance, or a variable diaphragm is closed down to less than 1 mm square, and a tomographic exposure is made. To indicate a center a brief exposure is made with the central ray perpendicular to the center of the plane. The pattern of the movement will be similar in both films and from these the various angles can be measured after plotting as in figure IV.11.

c. *Fixed focus systems:*

The same methods as in C.1.a. and C.1.b. can be used, the film or films being placed in the correct relation to the objective plane.

d. *Axial transverse systems:*

A pinhole diaphragm is fitted to the tube so as to project its image off center of the film turntable. The exposure on a film will show whether rotation was complete. It will also indicate whether intermittent exposures were correctly placed, and the degree of cut-off (fig. IV.12).

It is most important in axial transverse tomography that the two axes of the system should be exactly aligned with the focus. A test for this can be done with a wire placed vertically on the patient turntable. A tomogram will show a cross section of the wire when the alignment is correct; when not correct, the image of the wire will appear as a ring. A correction is made by moving the x-ray tube laterally a distance found from

$$T=\frac{SD}{d}$$

when

T=correction

S=radius of ring.

The direction of tube movement for correction is indicated by making a second exposure with the apparatus stationary. The result will then appear as in figure IV.13 where 3 is the stationary image and 4 the tomographic image. The tube or the film axis is always to be moved in the direction of the static image.

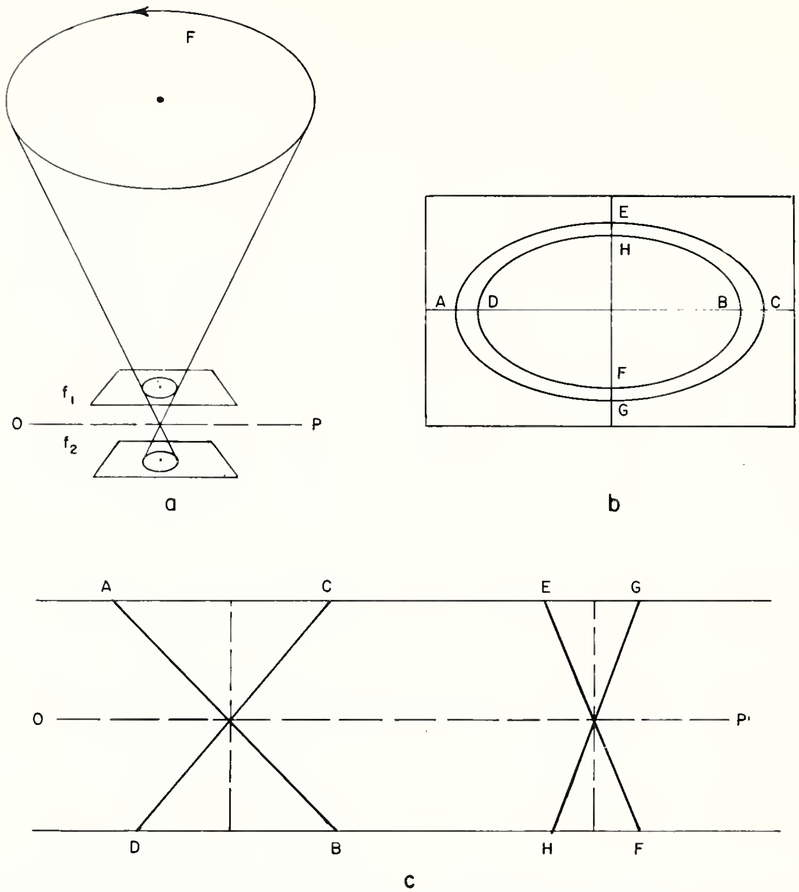

Figure IV.11. *(a) Arrangement, (b) superimposed film record, and (c) construction necessary to determine exposure angle for circular or elliptical movement.*

(See subsection C 1 b for details.)

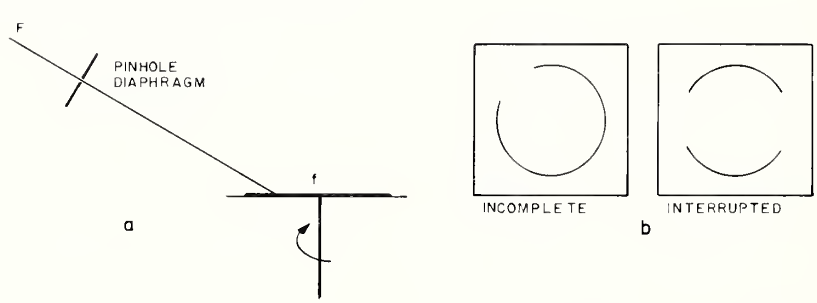

Figure IV.12. *(a) Arrangement and (b) film record of incomplete and interrupted test exposure for determining exposure angle of an axial transverse system.*

(See subsection C 1 d.)

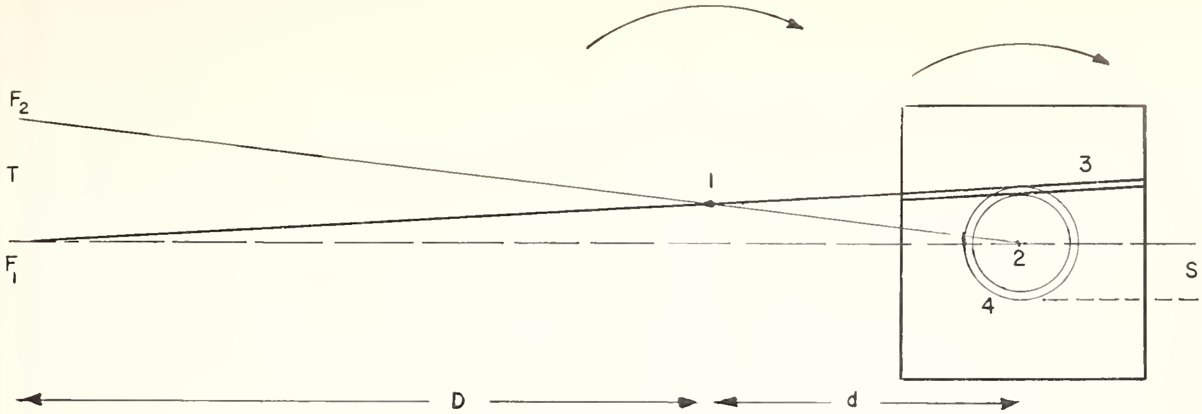

FIGURE IV.13. *Arrangement for determining misalinement in axial transverse system.*
(See subsection C 1 d.)

D. Location of Objective Plane

The objective plane is the plane whose shadow is stationary relative to the film. It is often assumed to be at the level of the fulcrum, but it can be shown that this is not always so. The objective plane is at the level of the fulcrum when the film axis is at the level of the film. Where the film is above or below the film axis the objective plane is above or below the fulcrum. This has no effect on the quality of a tomogram, but it does complicate the depth-scale marking which will be non-linear. It is advisable therefore to fix the film axis at the right level; even to allow for a degree of adjustment where absolute accuracy in localization is required. It may be remarked that the thickness of a cassette can account for a slight discrepancy.

In most tomographic equipments the selection of an objective plane is by moving the fulcrum, and by doing this the magnification varies at different levels, unless the FFD is adjusted. To obtain constant magnification without altering the FFD it is more convenient to have the fulcrum fixed, and to move the patient in order to select different planes. This means that a recumbent body has to be raised and lowered. In fixed tube systems where the patient is erect this method of selection is easily carried out, usually by moving the patient forward or backward.

Scales are usually set with the zero marking at the level of a supporting plane; table top or backrest. For axial transverse tomography the zero has to be set at an anatomical level chosen by the radiologist. All scales should be checked by means of a test object.

1. *Test of depth scale and exposure angle: rectilinear and curvilinear systems:*

A test object (see fig. IV.14) of X-ray transparent material is marked with opaque wires at 0.25 cm intervals up to a height of 20 or 25 cm. The scale portion is at 45 degrees. The test object is set so that a linear movement is at right angles to the length of the scale, and a tomographic exposure is made. The tomogram will appear similar to figure IV.15, indicating the site of the objective plane at the intersection of the longitudinal wire images, *E*. There will be more or less faint images of this wire extending above and below point *E*. If not distinct the images at *E* can be extended by scribing on the film; the transverse wire images will be easily visible.

The true exposure angle can be obtained by reconstruction as in figure IV.16. The height of the triangle can be any suitable one as measured on the scale in the tomogram. In the example the height selected (b) was 4 cm, and the base (c) was measured on the tomogram at 4 cm from *E*. The exposure angle was then found from

$$\frac{\alpha}{2}=\tan^{-1}\frac{c}{2b}.$$

2. *Test of depth scale and exposure angle: for all rectilinear and pluridirectional systems:*

The test object (see fig. IV.17) consists of a sloping sheet of 0.5 mm thick lead foil containing a narrow slit marked off with opaque wires across the slit at 1 cm intervals in height. There is also a horizontal sheet of lead foil perforated with a single hole, *R*, in line with the slit. The angle of slope can be between 30 and 45 degrees.

The test object is placed so that the single hole is directly under the focus. For a rectilinear motion the slit can conveniently be at a right angle to the movement, though it is not necessary. An exposure is first made with the central ray perpendicular to the objective plane, this will record an image of the single hole and of the slit. The tube and the system is then moved to any angle, the hole is covered with lead and another exposure is made. Finally, the slit is covered with lead,

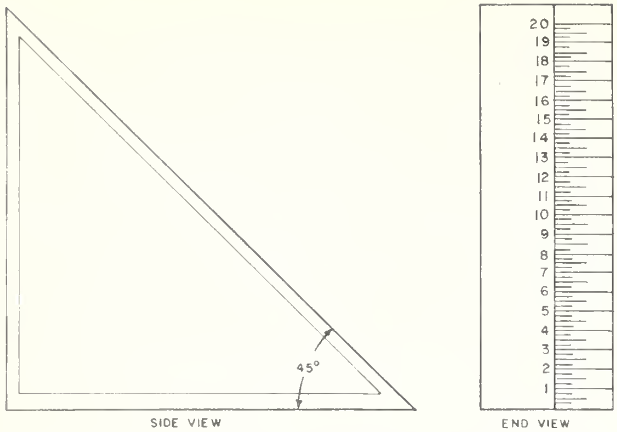

FIGURE IV.14. *Two views of test object used with rectilinear and culvilinear systems for determining the exposure angle and the location of the objective plane.*

The end view shows the radiation opaque wires spaced 0.25 cm apart.

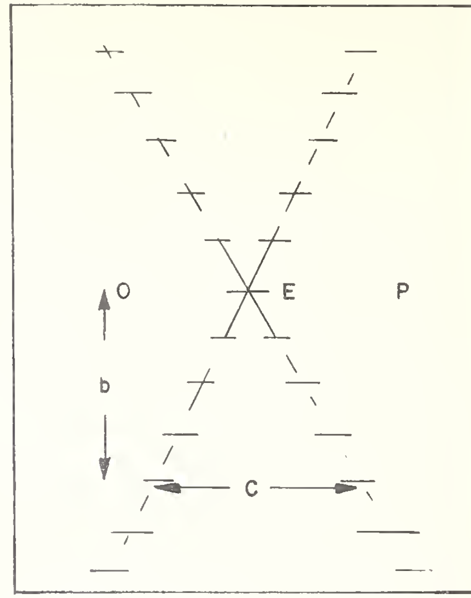

FIGURE IV.15. *Tomogram obtained with test object shown in figure IV.14.*

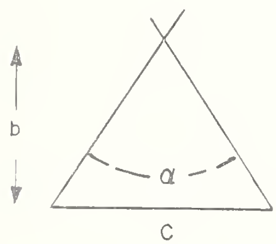

FIGURE IV.16. *Diagram for determining the exposure angle,* α, *where* $\alpha = 2\ \tan^{-1}(c/2b)$.

The distances b and c are obtained from a tomogram such as shown in figure IV.15.

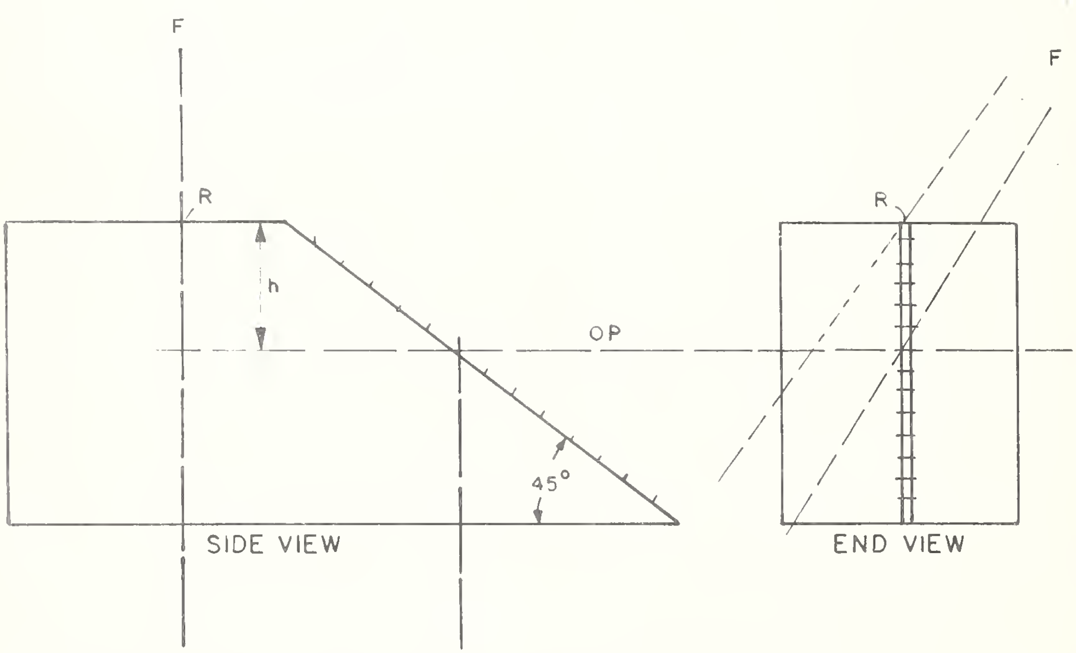

FIGURE IV.17. *Two views of test object used with all systems for determining the exposure angle and the location of the objective plane.*

A sheet of lead covers the top and sloping side. A slit shown in the end view is crossed by radiation opaque wires spaced 1 cm apart. The top has a small hole, R, in line with the slit.

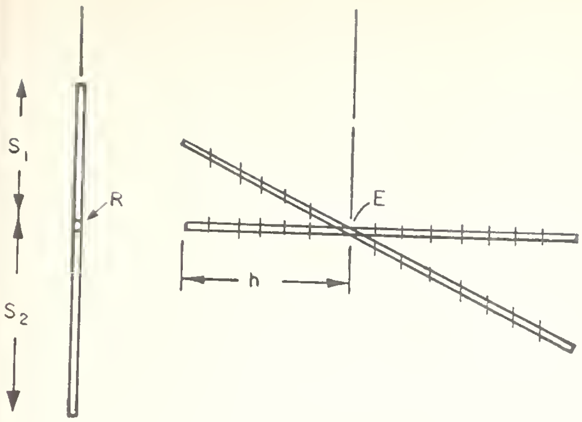

Figure IV.18. *Tomogram obtained for rectilinear motion with the test object shown in figure IV.17.*

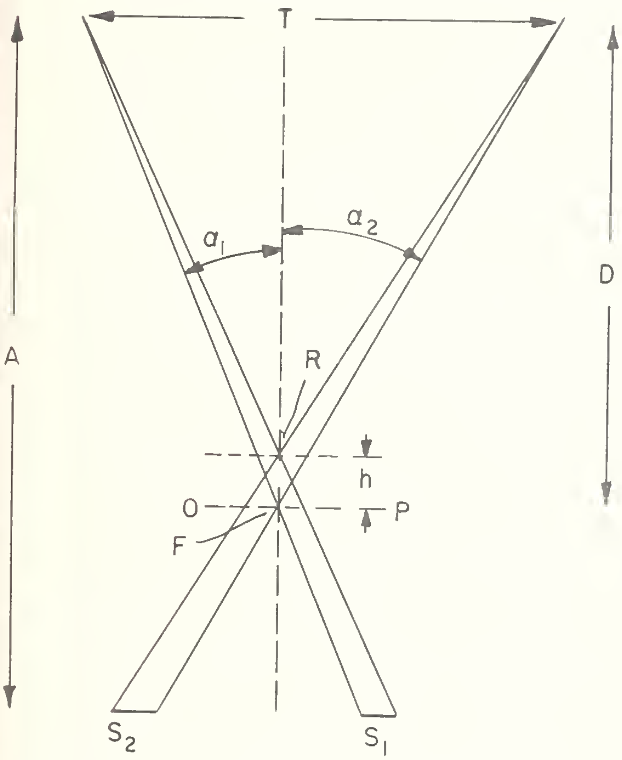

Figure IV.19. *Diagram for determining the exposure angles for rectilinear motion.*

The values of h, S_1, and S_2 and the position E are obtained from a tomogram such as figure IV.18. A is the focus-film distance and T is the distance moved by the focus during exposure.

the hole is uncovered, and a tomographic exposure is made. The resulting radiograph will appear similar to figure IV.18, and from this can be obtained the height of the objective plane, E, and the exposure angle, as shown in figure IV.19. The exposure angle is derived from the excursion of the image of the hole to each side of the central dot, where:

$$\tan \alpha_1 = (D-h)S_1/Ah$$

$$\tan \alpha_2 = (D-h)S_2/Ah.$$

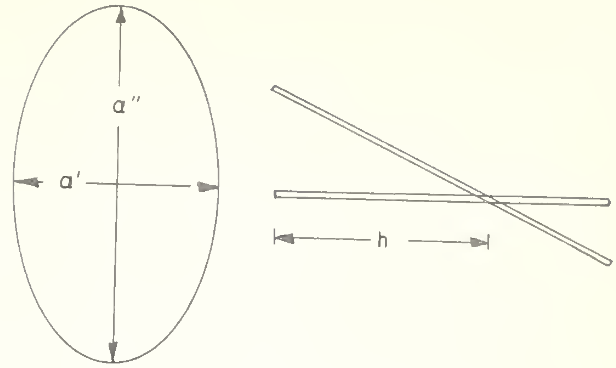

Figure IV.20. *Tomogram obtained for elliptical motion with the test object shown in figure IV.17.*

(See subsection D 2 and subsection E 2.)

The same procedure with a pluridirectional system will produce a radiograph similar to figure IV.20 which shows the trace of the dot with an elliptic motion. It may be shown that the exposure angles for the long and short axis of the ellipse can be derived in the same manner as for a rectilinear motion. The exposure angles for a more complex system can be obtained in the same manner.

E. Estimation of Section Interval, Selectivity, or Efficiency Coefficient

In any tomographic system only a plane is exactly in focus. A plane having no thickness does not cast a shadow, so what is seen in a tomogram is a number of planes sufficiently superimposed to produce a recognizable image. Definition is at a maximum in the objective plane and fades out more or less rapidly on either side of it. As the fade-out line is indefinable, "section thickness" is not a suitable term. Selectivity, interval factor, and efficiency coefficient have been suggested as better terms.

1. *Tomographic resolution on the film:*

Resolution is reduced by the varying inclination of image-forming rays in relation to the thickness of the film, or film-screen combination. An image point of the objective plane is elongated by a rectilinear-movement, and with a circular movement such an image point becomes a diffused annulus. This additional unsharpness is integrated with the unsharpness due to the crystalline structure of the screens, and therefore the total unsharpness is greater than that of the screens alone.

To find the tomographic resolution a suitable test object (see fig. IV.21) consists of a number of wires separated by their own diameter. A simple way of obtaining the wire separation is to wind two wires close together on a flat sheet of x-ray transparent material, then to remove one wire. The remaining wire on one side of the sheet can then be fixed with a cellulose adhesive. The wire on the other side is then removed by shearing the edges of the sheet. Wires from 0.5 to 3 mm in

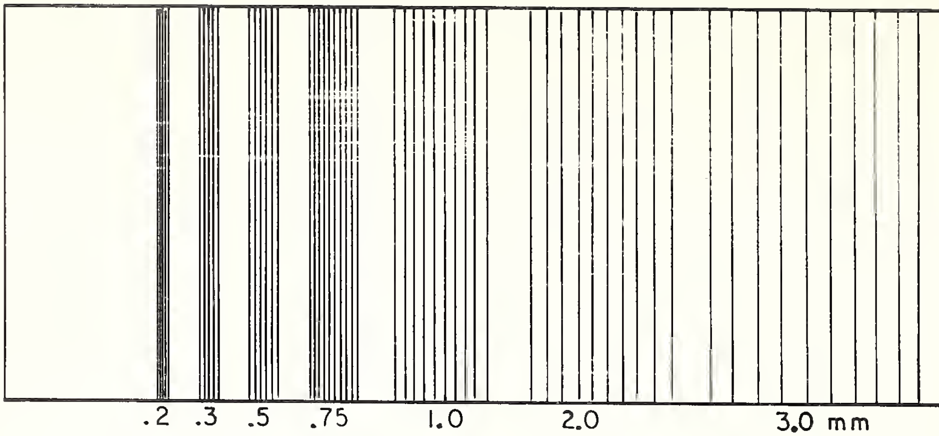

FIGURE IV.21. *Test object for determining tomographic resolution on the film.*

Paralleled radiation-opaque wires are spaced by a distanee equal to their own diameter. Groups of wires with diameters from 0.2 to 3 mm are suitable. Wire solder can be used for the 1 to 3 mm diameters.

diameter in groups of five are suitable and the support can be about 10 cm wide.

The test object is placed at an angle so each wire intersects the objective plane and a tomograph is taken with the movement at right angles to the length of the wires. The minimum diameter of wires which can be seen is a measure of the tomographic resolution. This test evaluates also the mechanical stability and alignment of the apparatus, and when these are bad even the 2 mm wires will be indistinguishable.

2. *Estimation of the section interval:*

a. To proceed to a complete study of a body region which could contain anatomic details of a given image diameter, Φ, the interval between each section is given in accordance with figure IV.19 where:

S=permissible blurring

α=exposure angle

h=section interval

$\Phi = S/2$.

b. In order to study contrast surfaces oblique to the section interval, the section interval may be increased to, for example, 1 cm. This is permissible if it is acceptable to ignore some anatomic details with a smaller diameter than Φ.

c. In the two cases E.2.a. and E.2.b. it seems that calculations could be made easier with the acceptance of an efficiency factor (S_{eff}) which should replace the previous concept of section thickness.

Starting again from the formula

$$\tan \alpha_1 = \frac{(D-h)}{Ah} S_1$$

$$\tan \alpha_2 = \frac{(D-h)}{Ah} S_2$$

we obtain

$$S_1 + S_2 = S = A(\tan \alpha_1 + \tan \alpha_2) \frac{h}{D-h},$$

α_1 and α_2 can be read at once on an angle indicator, placed around the fulcrum on which tangents are marked. The curve of $S = g\left(\frac{h}{D-h}\right)$ is practically rectilinear as long as h is small relative to D, which is generally the case. The error is not very important if it is assumed that there is an efficiency factor S_{eff} which is a constant and corresponds to the blurring of a point located at 1 cm above the objective plane.

$$S_{eff} = A \frac{(\tan \alpha_1 + \tan \alpha_2)}{D-1} = K,$$

a constant in calculation $S_h = S'_{eff} h$.

This example is given for unidirectional tomography but it can be used also for pluridirectional tomography, taking into account the fact that in the latter case there are two efficiency factors, S'_{eff} and S''_{eff}.

3. *Mean projection angle:*

The sharpness of the tomographic image in pluridirectional tomography is not only dependent on the exposure angle mentioned in E.2 but also on the path of movement of the focus; for instance spiral, circular, elliptic, and hypocycloidal. Therefore, it will be useful to introduce the concept "mean projection angle", or "effective exposure angle".

4. *Demonstration of depth of exploration:*

a. *Experimental demonstration of selectivity: unidirectional systems:*

Although it is difficult to devise an ideal test object for this purpose, a demonstration of the effect of different variables can be obtained by means of a test object composed of wires and strips of lead foil, as shown in figure IV.22a, b.

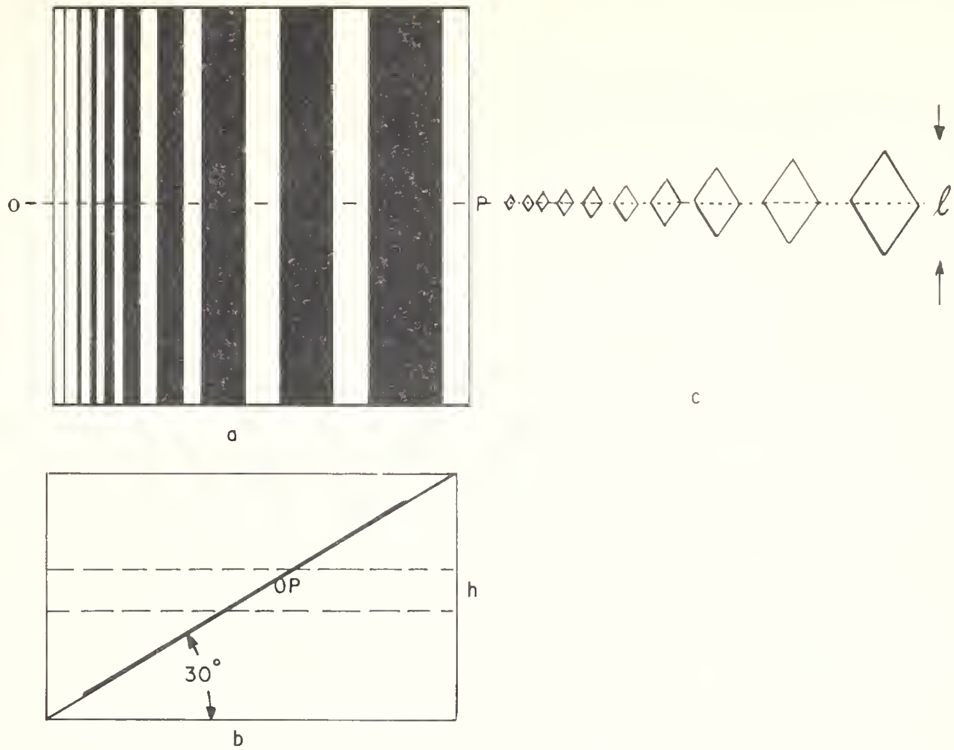

FIGURE IV.22. *Selectivity determination for unidirectional systems.*

a is a plan view of the test object—a flat sheet covered with radiation-opaque, paralleled, 0.5- to 10-mm wide strips and *b* is a side view. *c* indicates the type of tomogram obtained with rectilinear motion.

This is placed so each wire or strip intersects the objective plane at 30 degrees and with the strips at right angles to the movement during the exposure. A tomogram will look similar to figure IV.22c where l is the distance between the two extreme points of the diamond shaped images, and

$h=l/[2M/\sin 30°]=\dfrac{l}{4M}.$

Additional blocks of scatter material can be placed above and below the test object to simulate practical conditions.

b. Experimental demonstration of selectivity: pluridirectional systems:

A cylinder 7 cm in diameter and 12 cm long is wound with wire to a pitch of 1 cm (fig. IV.23a). The wire can be 1 mm or more in diameter. For diameters of more than 2 mm metal tubing is more suitable. The depth of negative images can be determined by winding with an x-ray transparent tubing of from 1 to 5 mm in diameter, and the whole test object immersed in tissue equivalent material (fig. IV.24).

A wire is also fixed coaxially in the cylinder which serves as a test for alignment of focus with the object and film axes.

The cylinder is placed with its long axis normal to the objective plane, which should intersect the cylinder. A tomogram will show a portion of one of the wire turns more distinctly, the arc of which can be related to 360 degrees and 1 cm pitch. For example, $90\times 10/360=2.5$ mm (fig. IV.23b).

$$h=\text{arc}\times\text{pitch}/360.$$

The 1 cm pitch can be increased to 2 or 4 cm where conditions are such that a whole turn of the 1 cm pitch is resolved.

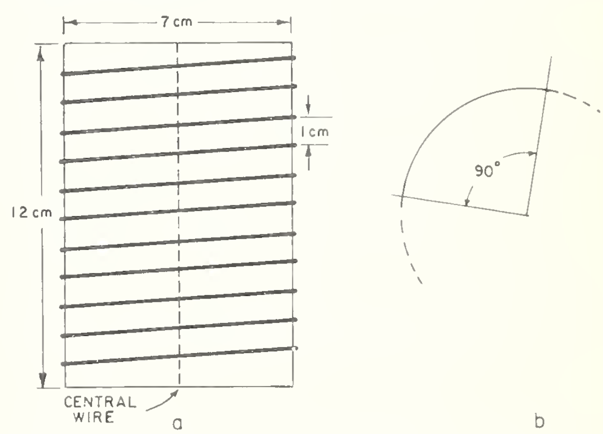

FIGURE IV.23. *Selectivity determination for pluridirectional systems and positive contrast.*

a shows the cylindrical test object with a helically-wound wire of 1 mm or more diameter, and b the resultant tomogram.

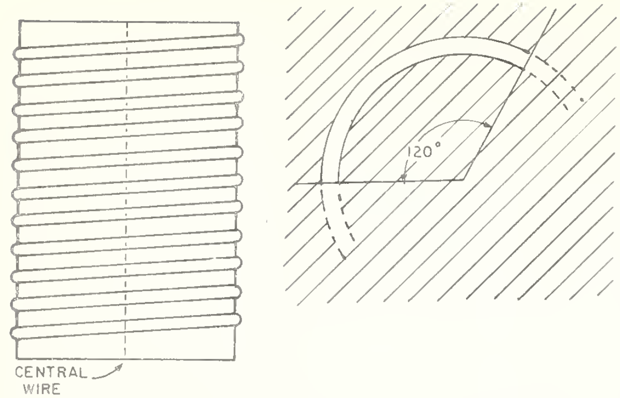

FIGURE IV.24. *Selectivity determination for pluridirectional systems and negative contrast.*

a shows the cylindrical test object with helically-wound transparent tubing. The test object is immersed on a tissue equivalent material. *b* shows the resultant tomogram.

F. Magnification

In tomography, it is important to know the magnification factor because different machines produce different degrees of magnification in an otherwise similar tomogram. Also the degree of magnification is generally greater than in conventional radiography because of the greater distance between object and film. This is especially so in axial transverse tomography. It is therefore necessary to be able to measure this factor in order to make comparisons, or to derive the exact size of object.

1. *Measurement of magnification:*

The magnification of the image in relation to the object can simply be derived from:

$$\frac{A}{A-d} \text{ or } \frac{A}{D} \text{ or } 1+\frac{d}{D}.$$

The magnification can also be measured by attaching a marker of known dimensions to the object so that an image of the marker is recorded in the tomogram. The relative size of the image to the marker gives the magnification factor. The actual size of the object is, $\frac{\text{diameter of image}}{M}$.

A circumferential marker, or scale, recorded in the transverse tomogram indicates the exact outline of the contour of the body recorded in the tomogram.

A further use of the marker recorded in a tomogram is that an optically reduced image of the tomogram can easily be projected to the actual size.

G. Dose Considerations in Tomography

In addition to the precautions taken and the adjustment of factors in conventional radiography, such as filtration, diaphragming, quality of radiation, focus-skin distance, and magnification, tomography requires some special consideration in view of the fact that the exposure is repeated a number of times in the taking of a series of tomograms. The number of exposures may be reduced by preliminary localization of a pathological area or plane.

1. *Single film systems:*

It has been found that the surface dose of radiation involved in the taking of a single tomogram, with a moving tube system or its equivalent, is only about 30 percent greater than that required for conventional radiography. The dose required at the film for a given film density is the same. The dose distribution at a given depth in the body during tomographic exposure shows wider variations than during conventional radiographic exposure. However, there is no dose concentration in tomography such as is evident for the much larger angles of rotation sometimes employed in therapy.

A comparison of the gonadal dose received during a tomographic and during a conventional radiographic exposure indicate that the two techniques give similar results [1]. Such comparisons are difficult on a general basis because the conventional radiograph often requires a larger field but there are usually several exposures per tomographic examination (figs. 10 and 11 in reference 1).

2. *Multitomography* (*multisection radiography*):

There is no great difference between the dose in multitomography with intensifying screens and the same number of sequential tomograms, where the tomographic qualities are similar. The exposure for a set of multitomograms is determined by the speed of the top screens, which are normally very slow, thus accounting for a heavy exposure. Dose reduction is not one of the advantages of multitomography with intensifying screens. However, where non-screen film is used, as for extremities, the dose reduction is proportional to the number of films which are exposed simultaneously (fig. 8 in reference 1).

H. References

References to publications prior to 1959 can be found in Index Stratigraphicus, from the Institut di Radiologica dell Universita di Genova, G. L. Besie (Edizione Universitaie).

1. Stieve, F. E., Radiation exposure to body section radiography, Acta Radiol. (38 references), **55,** 465–485 (1961).
2. Widenmann, L., Comparative investigations of the picture quality and dose needs in simultaneous and individual-layer photography, (in German), Fortschr. Rontgrenstr. (21 references), **89,** 613–623 (1958).
3. Vallebona, A., Lights and shadows in stratigraphy, (in Italian), Nuntius Radiologious, Estratto, June 1959 (Sec. Ed. Universo., Roma).

Appendix I*

Radiation Quantities and Units

International Commission on Radiological Units and Measurements (ICRU) Report 10a 1962

1. Introduction

There has recently been much discussion of the fundamental concepts and quantities employed in radiation dosimetry. This has arisen partly from the rapid increase in the number of individuals using these concepts in the expanding field of nuclear science and technology, partly because of the need for extending the concepts so that they would be of use at higher photon energies and for particulate as well as for photon radiation, but chiefly because of certain obscurities in the existing formulation of the quantities and units themselves.

The roentgen, for example, was originally defined to provide the best quantitative measure of exposure to medium energy x radiation which the measuring techniques of that day (1928) permitted. The choice of air as a standard substance was not only convenient but, also appropriate for a physical quantity which was to be correlated with the biological effect of x rays, since the effective atomic number of air is not very different from that of tissue. Thus a given biological response could be reproduced approximately by an equal exposure in roentgens for x-ray energies available at that time. Since 1928 the definition of the roentgen has been changed several times, and this has reflected some feeling of dissatisfaction with the clarity of the concept.

The most serious source of confusion was the failure to define adequately the radiation quantity of which the roentgen was said to be the unit.[1] As a consequence of this omission the roentgen had gradually acquired a double role. The use of this name for the unit had become recognized as a way of specifying not only the magnitude but also the nature of the quantity measured. This practice conflicts with the general usage in physics, which permits, within the same field, the use of a particular unit for all quantities having the same dimensions.

Even before this, the need for accurate dosimetry of neutrons and of charged particles from accelerators or from radionuclides had compelled the International Commission on Radiological Units and Measurements (ICRU) to extend the number of concepts. It was also desired to introduce a new quantity which could be more directly correlated with the local biological and chemical effects of radiation. This quantity, *absorbed dose*, has a generality and simplicity which greatly facilitated its acceptance, and in a very few years it has become widely used in every branch of radiation dosimetry.

The introduction of absorbed dose into the medical and biological field was further assisted by defining a special unit—the *rad*. One rad is approximately equal to the absorbed dose delivered when soft tissue is exposed to one roentgen of medium voltage x radiation. Thus in many situations of interest to medical radiology, but not in all, the numbers of roentgens and rads associated with a particular medical or biological effect are approximately equal and experience with the earlier unit could be readily transferred to the new one. Although the *rad* is merely a convenient multiple of the fundamental unit, erg/g, it has already acquired, at least in some circles, the additional connotation that the only quantity which can be measured in rads is absorbed dose. On the other hand, the rad has been used by some authors as a unit for a quantity called by them *first collision dose;* this practice is deprecated by the Commission.

Being aware of the need for preventing the emergence of different interpretations of the same quantity, or the introduction of undesirable, unrelated quantities or units in this or similar fields of measurement, the ICRU set up, during its meeting in Geneva in September 1958, an *Ad Hoc* Committee. The task of this committee was to review the fundamental concepts, quantities, and units which are required in radiation dosimetry and to recommend a system of concepts and a set of definitions which would be, as far as possible, internally consistent and of sufficient generality to cover present requirements and such future requirements as can be foreseen. The committee was instructed to pay more attention to consistency and rigor than to the historical development of the subject and was authorized to reject any existing quantities or units which seemed to hinder a consistent and unified formulation of the concepts.

Bertrand Russell[2] in commenting on the use and abuse of the concept of infinitesimals by mathematicians remarks: "But mathematicians did not at first pay heed to (these) warnings. They went ahead and developed their science, and it is well that they should have done so. It is a peculiar fact about the genesis and growth of new disciplines that too much rigor too early imposed stifles the imagination and stultifies invention. A certain freedom from the strictures of sustained formality tends to promote the development of a

*Taken from Radiation Quantities and Units, International Commission on Radiological Units and Measurements, Report 10a, National Bureau of Standards Handbook 84 (1962). (Numbers refer to paragraphs in the original report.)

[1] Fränz, H. and Hübner, W. Concepts and Measurement of Dose, Proceedings of Second International Conference on the Peaceful Uses of Atomic Energy, Geneva 1958, P/971 21, 101, United Nations, Geneva (1958).

[2] Russell, B., Wisdom of the West, p. 280 (Doubleday and Co., Inc. New York, 1959).

subject in its early stages, even if this means the risk of a certain amount of error. Nonetheless, there comes a time in the development of any field when standards of rigor have to be tightened."

The purpose of the present reexamination of the concepts to be employed in radiation dosimetry was primarily "to tighten standards of rigor". If, in the process some increased formality is required in the definitions in order to eliminate any foreseeable ambiguities, this must be accepted.

2. General Considerations

The development of the more unified presentations of quantities and units which is here proposed was stimulated and greatly assisted by mathematical models of the dosimetric field which had been proposed by some members of the committee in an effort to clarify the concepts. It appeared, however, that the essential features of the mathematical models had been incorporated into the definitions and hence the need for their exposition in this report largely disappeared. The mathematical approach is published elsewhere.[3]

As far as possible, the definitions of the various fundamental quantities given here conform to a common pattern. Complex quantities are defined in terms of the simpler quantities of which they are comprised.

The passage to a "macroscopic limit" which has to be used in defining point quantities in other fields of physics can be adapted to radiation quantities and a special discussion of this is included in the section headed "limiting procedures."

The general pattern adopted is to give a short definition and to indicate the precise meaning of any special phrase or term used by means of an explanatory note following the definition. There has been no attempt to make the list of quantities which are defined here comprehensive. Rather the Commission has striven to clarify the fundamental dosimetric quantities and a few others (such as activity) which were specifically referred to it for discussion.

It is recognized that certain terms for which definitions are proposed here are of interest in other fields of science and that they are already variously defined elsewhere. The precise wording of the definition and even the name and symbol given to any such quantity, may at some future date require alteration if discussions with representatives of the other interested groups of scientists should lead to agreement on a common definition or symbol. Although the definitions presented here represent some degree of compromise, they are believed to meet the requirements in the field of radiation dosimetry.

3. Quantities, Units, and Their Names

The Commission is of the opinion that the definition of concepts and quantities is a fundamental matter and that the choice of units is of less importance. Ambiguity can best be avoided if the defined quantity which is being measured is specified. Nevertheless, the special units do exist in this as in many other fields. For example, the hertz is restricted, by established convention, to the measurement of vibrational frequency, and the curie, in the present recommendations, to the measurement of the activity of a quantity of a nuclide. One does not measure activity in hertz nor frequency in curies although these quantities have the same dimensions.

It was necessary to decide whether or not to extend the use of the special dosimetric units to other more recently defined quantities having the same dimensions, to retain the existing restriction on their use to one quantity each, or to abandon the special units altogether. The Commission considers that the addition of further special units in the field of radiation dosimetry is undesirable, but continues to recognize the existing special units. It sees no objection, however, to the expression of any defined quantity in the appropriate units of a coherent physical system. Thus, to express absorbed dose in ergs per gram or joules per kilogram, exposure in coulombs per kilogram or activity in reciprocal seconds, are entirely acceptable alternatives to the use of the special units which, for historical reasons, are usually associated with these quantities.

The ICRU recommends that the use of each special unit be restricted to one quantity as follows:

The rad—solely for absorbed dose
The roentgen—solely for exposure
The curie—solely for activity.

It recommends further that those who prefer to express quantities such as absorbed dose and kerma (see below) in the same units should use units of an internationally agreed coherent system.

Several new names are proposed in the present report. When the absorbed dose concept was adopted in 1953, the Commission recognized the need for a term to distinguish it from the quantity of which the roentgen is the unit. In 1956 the Commission proposed the term *exposure* for this latter quantity. To meet objections by the ICRP, a compromise term, "exposure dose" was agreed upon.[4] While this term has come into some use since then, it has never been considered as completely satisfactory. In the meantime, the basic cause of the ICRP objection has largely disappeared since most legal codes use either the units rad or rem.

Since in this report the whole system of radiological quantities and units has come under critical review, it seemed appropriate to reconsider the 1956 decision. Numerous names were examined as a replacement for exposure dose, but there were serious objections to any which included the word dose. There appeared to be a minimum of objection to the name *exposure* and hence this term has been adopted by the Commission with

[3] Rossi, H. H., and Roesch, W. C., Field Equations in Dosimetry, Radiation Res. 16, 783 (1962).

[4] For details see ICRU, 1956 Report, NBS Handb. 62, p. 2 (1957).

the hope that the question has been permanently settled. It involves a minimum change from the older name exposure dose. Furthermore, the elimination of the term "dose" accomplishes the long-felt desire of the Commission to retain the term dose for one quantity only—the absorbed dose.

The term "*RBE* dose" has in past publications of the Commission not been included in the list of definitions but was merely presented as a "recognized symbol." In its 1959 report the Commission also expressed misgivings over the utilization of the same term, "*RBE*", in both radiobiology and radiation protection. It now recommends that the term *RBE* be used in radiobiology only and that another name be used for the linear-energy-transfer-dependent factor by which absorbed doses are to be multiplied to obtain for purposes of radiation protection a quantity that expresses on a common scale for all ionizing radiations the irradiation incurred by exposed persons. The name recommended for this factor is the *quality factor*, (*QF*). Provisions for other factors are also made. Thus, a *distribution factor*, (*DF*), may be used to express the modification of biological effect due to non-uniform distribution of internally deposited isotopes. The product of absorbed dose and modifying factors is termed the *dose equivalent*, (*DE*). As a result of discussions between ICRU and ICRP the following formulation has been agreed upon:

The Dose Equivalent

1. For protection purposes it is useful to define a quantity which will be termed the "dose equivalent", (*DE*).
2. (*DE*) is defined as the product of absorbed dose, *D*, quality factor, (*QF*), dose distribution factor, (*DF*), and other necessary modifying factors.

$$(DE)=D(QF)(DF). \; . \; . \; .$$

3. The unit of dose equivalent is the "rem". The dose equivalent is numerically equal to the dose in rads multiplied by the appropriate modifying factors.

Although this statement does not cover a number of theoretical aspects (in particular the physical dimensions of some of the quantities) it fulfills the immediate requirement for an unequivocal specification of a scale that may be used for numerical expression in radiation protection.

Another new name is that for the quantity which represents the kinetic energy transferred to charged particles by the uncharged particles per unit mass of the irradiated medium. This is the same as one of the common interpretations of a concept "first collision dose," that has proved to be of great value in the dosimetry of fast neutrons. The concept is also closely related to the energy equivalent of exposure in an x-ray beam. The name proposed, *kerma*, is based on the initials of *k*inetic *e*nergy *r*eleased in *m*aterial.

Still another new name is the *energy fluence* which is here attached to the quantity in the 1953 ICRU report called *quantity of radiation*. The latter term was dropped in the 1956 ICRU report but the concept—time integral of intensity—remains a useful one and the proposed term appears to be acceptable in other languages as well as English. A related quantity, *particle fluence*, which is equivalent to the quantity *nvt* used in neutron physics, is included to round out the system of radiation quantities.

The quantity for which the curie is the unit was referred to the committee for a name and definition. Hitherto the curie has been defined as a *quantity of the radioactive nuclide* such that 3.7×10^{10} disintegrations per second occur in it. However, it has never been specified what was meant by quantity of a nuclide, whether it be a number, mass, volume, etc., Meanwhile the custom has grown of identifying the number of curies of a radionuclide with its transformation rate. Because of the vagueness of the original concept, because of the custom of identifying curies with transformation rate and because it appeared not to interfere with any other use of the curie, the Commission recommends that the term *activity* be used for the transformation rate, and that the curie be made its unit. It is recognized that the definition of the curie is of interest to other bodies in addition to the ICRU, but by this report we recommend that steps be taken to redefine it as $3.7\times10^{10}s^{-1}$, i.e., as a unit of activity and not of quantity of a nuclide.

It is also recommended that the term *specific gamma ray constant* be used instead of *specific gamma ray emission* for the quotient of the exposure rate at a given distance by the activity. The former term focuses attention on the *constancy* of this quotient for a given nuclide rather than the *emission* of the source.

4. Detailed Considerations

A. Limiting Procedures

Except in the case of a uniform distribution of sources throughout a large region, radiation fields are in general non-uniform in space. They may also be variable in time. Many of the quantities defined in this report have to be specified as functions of space or time, and in principle they must therefore be determined for sufficiently small regions of space or intervals of time by some limiting procedure. There are conceptual difficulties in taking such limits for quantities which depend upon the discrete interactions between radiations and atoms. Similar difficulties arise with other macroscopic physical quantities such as density or temperature and they must be overcome by means of an appropriate averaging procedure.

To illustrate this procedure we may consider the measurement of the macroscopic quantity "absorbed dose" in a non-uniform radiation field. In measuring this dose the quotient of energy by mass must be taken in an elementary volume in the medium which on the one hand is so small that a further reduction in its size would not appreciably change the measured value of the quotient energy by mass and on the other hand is still large enough to contain many interactions and be traversed by many particles.[5] If it is impossible to find a mass such that both these conditions are met, the dose cannot be established directly in a single measurement. It can only be deduced from multiple measurements that involve extrapolation or averaging procedures. Similar considerations apply to some of the other concepts defined below. The symbol Δ precedes the symbols for quantities that may be concerned in such averaging procedures.

In the measurement of certain material constants such as stopping power, absorption coefficient, etc., the limiting procedure can be specified more rigorously. Such constants can be determined for a given material with any desired accuracy without difficulties from statistical fluctuations. In these cases the formula quoted in the definitions are presented as differential quotients.

B. Spectral Distributions and Mean Values

In practice many of the quantities defined below to characterize a radiation field and its interaction with matter are used for radiations having a complex energy spectrum. An important general concept in this connection is the *spectral concentration* of one quantity with respect to another. The spectral concentration is the ordinate of the distribution function of the first quantity with respect to the second. The independent quantity need not always be energy or frequency; one can speak of the spectral concentration of flux density with respect to quantum energy or of the absorbed dose with respect to linear energy transfer. The interaction constants (such as μ, S and W) referred to in this report are often mean values taken over the appropriate spectral distributions of the corresponding quantities.

C. Units

For any of the quantities defined below the appropriate unit of an internationally agreed coherent system can be used. In addition certain special units are reserved for special quantities:

the rad for absorbed dose
the roentgen for exposure
the curie for activity.

D. Definitions

(1) *Directly ionizing particles* are charged particles (electrons, protons, α-particles, etc.) having sufficient kinetic energy to produce ionization by collision.

(2) *Indirectly ionizing particles* are uncharged particles (neutrons, photons, etc.) which can liberate directly ionizing particles or can initiate a nuclear transformation.

[5] In interpreting radiation effects the macroscopic concept of absorbed dose may not be sufficient. Whenever the statistical fluctuations around the mean value are important, additional parameters describing the distribution of absorbed energy on a microscopic scale are necessary.

(3) *Ionizing radiation* is any radiation consisting of directly or indirectly ionizing particles or a mixture of both.

(4) The *energy imparted* by ionizing radiation to the matter in a volume is the difference between the sum of the energies of all the directly and indirectly ionizing particles which have entered the volume and the sum of the energies of all those which have left it, minus the energy equivalent of any increase in rest mass that took place in nuclear or elementary particle reactions within the volume.

NOTES: (a) The above definition is intended to be exactly equivalent to the previous meanings given by the ICRU to "energy retained by matter and made locally available" or "energy which appears as ionization, excitation, or changes of chemical bond energies". The present formulation specifies what energy is to be included without requiring a lengthy, and possibly incomplete, catalogue of the different types of energy transfer.

(b) Ultimately, most of the energy imparted will be degraded and appear as heat. Some of it, however, may appear as a change in interatomic bond energies. Moreover, during the degradation process the energy will diffuse and the distribution of heat produced may be different from the distribution of imparted energy. For these reasons the energy imparted cannot always be equated with the heat produced.

(c) The quantity *energy imparted to matter* in a given volume is identical with the quantity often called *intergral absorbed dose* in that volume.

(5) The *absorbed dose* (D) is the quotient of ΔE_D by Δm, where ΔE_D is the energy imparted by ionizing radiation to the matter in a volume element, Δm is the mass of the matter in that volume element and Δ has the meaning indicated in section 4.A.

$$D=\frac{\Delta E_D}{\Delta m}$$

The special unit of absorbed dose is the *rad*.

$$1 \text{ rad}=100 \text{ erg}/g=\frac{1}{100}\ J/kg.$$

NOTE: J is the abbreviation for Joule.

(6) The *absorbed dose rate* is the quotient of ΔD by Δt, where ΔD is the increment in absorbed dose in time Δt and Δ has the meaning indicated in section 4.A.

$$\text{Absorbed dose rate}=\frac{\Delta D}{\Delta t}$$

A special unit of absorbed dose rate is any quotient of the rad by a suitable unit of time (rad/d, rad/min, rad/h, etc.).

(7) The *particle fluence* [6] or *fluence* (Φ) of particles is the quotient of ΔN by Δa, where ΔN is the number of particles which enter a sphere [7]

[6] This quantity is the same as the quantity, *nvt*, commonly used in neutron physics.

[7] This quantity is sometimes defined with reference to a plane of area Δa, instead of a sphere of cross-sectional area Δa. The plane quantity is less useful for the present purposes and it will not be defined. The two quantities are equal for a unidirectional beam of particles perpendicularly incident upon the plane area.

of cross-sectional area Δa and Δ has the meaning indicated in section 4.A.

$$\Phi=\frac{\Delta N}{\Delta a}$$

(8) The *particle flux density* or *flux density* (φ) of particles is the quotient of $\Delta\Phi$ by Δt where $\Delta\Phi$ is the particle fluence in time Δt and Δ has the meaning indicated in section 4.A.

$$\varphi=\frac{\Delta\Phi}{\Delta t}$$

NOTE: This quantity may also be referred to as particle fluence rate.

(9) The *energy fluence* (F) of particles is the quotient of ΔE_F by Δa, where ΔE_F is the sum of the energies, exclusive of rest energies, of all the particles which enter a sphere [8] of cross-sectional area Δa and Δ has the meaning indicated in section 4.A.

$$F=\frac{\Delta E_F}{\Delta a}$$

(10) The *energy flux density* or *intensity* (I) is the quotient of ΔF by Δt where ΔF is the energy fluence in the time Δt and Δ has the meaning indicated in section 4.A.

$$I=\frac{\Delta F}{\Delta t}$$

NOTE: This quantity may also be referred to as energy fluence rate.

(11) The *kerma* [9] (K) is the quotient of ΔE_K by Δm, where ΔE_K is the sum of the initial kinetic energies of all the charged particles liberated by indirectly ionizing particles in a volume element of the specified material, Δm is the mass of the matter in that volume element and Δ has the meaning indicated in section 4.A.

$$K=\frac{\Delta E_K}{\Delta m}$$

NOTES: (a) Since ΔE_K is the sum of the initial kinetic energies of the charged particles liberated by the indirectly ionizing particles, it includes not only the kinetic energy these charged particles expend in collisions but also the energy they radiate in bremsstrahlung. The energy of any charged particles is also included when these are produced in secondary processes occurring within the volume element. Thus the energy of Auger electrons is part of ΔE_K.

(b) In actual measurements Δm should be so small that its introduction does not appreciably disturb the radiation field. This is particularly necessary if the medium for which kerma is determined is different from the ambient medium; if the disturbance is appreciable an appropriate correction must be applied.

(c) It may often be convenient to refer to a value of kerma or of kerma rate for a specified material in free space or at a point inside a different material. In such a case the value will be that which would be obtained if a small quantity of the specified material were placed at the point of interest. It is, however, permissible to make a statement such as: "The kerma for air at the point P inside a water phantom is . . ." recognizing that this is a shorthand version of the fuller description given above.

(d) A fundamental physical description of a radiation field is the intensity (energy flux density) at all relevant points. For the purpose of dosimetry, however, it may be convenient to describe the field of indirectly ionizing particles in terms of the kerma rate for a specified material. A suitable material would be air for electromagnetic radiation of moderate energies, tissue for all radiations in medicine or biology, or any relevant material for studies of radiation effects.

Kerma can also be a useful quantity in dosimetry when charged particle equilibrium exists at the position and in the material of interest, and bremsstrahlung losses are negligible. It is then equal to the absorbed dose at that point. In beams of x or gamma rays or neutrons, whose energies are moderately high, transient charged-particle equilibrium can occur; in this condition the kerma is just slightly less than the absorbed dose. At very high energies the difference becomes appreciable. In general, if the range of directly ionizing particles becomes comparable with the mean free path of the indirectly ionizing particles, no equilibrium will exist.

(12) The *kerma rate* is the quotient of ΔK by Δt, where ΔK is the increment in kerma in time Δt and Δ has the meaning indicated in section 4.A.

(13) The *exposure* (X) is the quotient of ΔQ by Δm, where ΔQ is the sum of the electrical charges on all the ions of one sign produced in air when all the electrons (negatrons and positrons), liberated by photons in a volume element of air whose mass is Δm, are completely stopped in air and Δ has the meaning indicated in section 4.A.

$$X=\frac{\Delta Q}{\Delta m}$$

The special unit of exposure is the roentgen (R).

$$1R=2.58\times10^{-4}C/kg$$ [10]

NOTES: (a) The words "charges on all the ions of one sign" should be interpreted in the mathematically absolute sense.

(b) The ionization arising from the absorption of bremsstrahlung emitted by the secondary electrons is not to be included in ΔQ. Except

[8] See footnote 7.

[9] Various other methods of specifying a radiation field have been used, e.g., for a neutron source the "first collision dose" in a standard material at a specified point (see Introduction).

[10] This unit is numerically identical with the old one defined as 1 e.s.u. of charge per .001293 gram of air. C is the abbreviation for coulomb.

for this small difference, significant only at high energies, the exposure as defined above is the ionization equivalent of the kerma in air.

(c) With present techniques it is difficult to measure exposure when the photon energies involved lie above a few Mev or below a few kev.

(d) As in the case of kerma (4D(11), note (c)), it may often be convenient to refer to a value of exposure or of exposure rate in free space or at a point inside a material different from air. In such a case the value will be that which would be determined for a small quantity of air placed at the point of interest. It is, however, permissible to make a statement such as: "The exposure at the point P inside a water phantom is . . . ."

(14) The *exposure rate* is the quotient of ΔX by Δt, where ΔX is the increment in exposure in time Δt and Δ has the meaning indicated in section 4.A,

$$\text{Exposure rate} = \frac{\Delta X}{\Delta t}$$

A special unit of exposure rate is any quotient of the roentgen by a suitable unit of time (R/s, R/min, R/h, etc.).

(15) The *mass attenuation coefficient* $\left(\frac{\mu}{\rho}\right)$ of a material for indirectly ionizing particles is the quotient of dN by the product of ρ, N, and dl where N is the number of particles incident normally upon a layer of thickness dl and density ρ, and dN is the number of particles that experience interactions in this layer.

$$\frac{\mu}{\rho} = \frac{1}{\rho N}\frac{dN}{dl}$$

Notes: (a) The term "interactions" refers to processes whereby the energy or direction of the indirectly ionizing particles is altered.

(b) For X or gamma radiations

$$\frac{\mu}{\rho} = \frac{\tau}{\rho} + \frac{\sigma}{\rho} + \frac{\sigma_{\text{coh}}}{\rho} + \frac{\kappa}{\rho}$$

where $\frac{\tau}{\rho}$ is the mass photoelectric attenuation coefficient, $\frac{\sigma}{\rho}$ is the total Compton mass attenuation coefficient, $\frac{\sigma_{\text{coh}}}{\rho}$ is the mass attenuation coefficient for coherent scattering, and $\frac{\kappa}{\rho}$ is the pair-production mass attenuation coefficient.

(16) The *mass energy transfer coefficient* $\left(\frac{\mu_K}{\rho}\right)$ of a material for indirectly ionizing particles is the quotient of dE_K by the product of E, ρ and dl where E is the sum of the energies (excluding rest energies) of the indirectly ionizing particles incident normally upon a layer of thickness dl and density ρ, and dE_K is the sum of the kinetic energies of all the charged particles liberated in this layer.

$$\frac{\mu_K}{\rho} = \frac{1}{E\rho}\frac{dE_K}{dl}$$

Notes: (a) The relation between fluence and kerma may be written as

$$K = F\frac{\mu_K}{\rho}$$

(b) For X or gamma rays of energy $h\nu$

$$\frac{\mu_K}{\rho} = \frac{\tau_a}{\rho} + \frac{\sigma_a}{\rho} + \frac{\kappa}{\rho}$$

where

$$\frac{\tau_a}{\rho} = \frac{\tau}{\rho}\left(1 - \frac{\delta}{h\nu}\right)$$

$\left(\frac{\tau}{\rho}\right.$=the photoelectric mass attenuation coefficient, δ=average energy emitted as fluorescent radiation per photon absorbed.$\left.\right)$

and

$$\frac{\sigma_e}{\rho} = \frac{\sigma}{\rho}\frac{E_e}{h\nu}$$

$\left(\frac{\sigma}{\rho}\right.$=total Compton mass attenuation coefficient, E_e=average energy of the Compton electrons per scattered photon.$\left.\right)$

and

$$\frac{\kappa_a}{\rho} = \frac{\kappa}{\rho}\left(1 - \frac{2mc^2}{h\nu}\right)$$

$\left(\frac{\kappa}{\rho}\right.$=mass attenuation coefficient for pair production, mc^2=rest energy of the electron.$\left.\right)$

(17) The *mass energy-absorption coefficient* $\left(\frac{\mu_{en}}{\rho}\right)$ of a material for indirectly ionizing particles is $\frac{\mu_K}{\rho}$ (1-G) where G is the proportion of the energy of secondary charged particles that is lost to bremsstrahlung in the material.

Notes: (a) When the material is air, $\frac{\mu_{en}}{\rho}$ is proportional to the quotient of exposure by fluence.

(b) $\frac{\mu_K}{\rho}$ and $\frac{\mu_{en}}{\rho}$ do not differ appreciably unless the kinetic energies of the secondary particles are comparable with or larger than their rest energy.

(18) The *mass stopping power* $\left(\frac{S}{\rho}\right)$ of a material for charged particles is the quotient of dE_s by the product of dl and ρ, where dE_s is the average energy lost by a charged particle of specified energy in traversing a path length dl, and ρ is the density of the medium.

$$\frac{S}{\rho} = \frac{1}{\rho}\frac{dE_s}{dl}$$

NOTE: dE_s denotes energy lost due to ionization, electronic excitation and radiation. For some purposes it is desirable to consider stopping power with the exclusion of bremsstrahlung losses. In this case $\frac{S}{\rho}$ must be multiplied by an appropriate factor that is less than unity.

(19) The *linear energy transfer* (L) of charged particles in a medium is the quotient of dE_L by dl where dE_L is the average energy locally imparted to the medium by a charged particle of specified energy in traversing a distance of dl.

$$L=\frac{dE_L}{dl}$$

NOTES: (a) The term "locally imparted" may refer either to a maximum distance from the track or to a maximum value of discrete energy loss by the particle beyond which losses are no longer considered as local. In either case the limits chosen should be specified.

(b) The concept of linear energy transfer is different from that of stopping power. The former refers to energy imparted within a limited volume, the latter to loss of energy regardless of where this energy is absorbed.

(20) The *average energy* (W) *expended in a gas per ion pair formed* is the quotient of E by N_W, where N_W is the average number of ion pairs formed when a charged particle of initial energy E is completely stopped by the gas.

$$W=\frac{E}{N_W}$$

NOTES: (a) The ions arising from the absorption of bremsstrahlung emitted by the charged particles are not to be counted in N_W.

(b) In certain cases it may be necessary to consider the variation in W along the path of the particle, and a differential concept is then required, but is not specifically defined here.

(21) A *nuclide* is a species of atom having specified numbers of neutrons and protons in its nucleus.

(22) The *activity* (A) of a quantity of a radioactive nuclide is the quotient of ΔN by Δt where ΔN is the number of nuclear transformations which occur in this quantity in time Δt and Δ has the meaning indicated in section 4.A.

$$A=\frac{\Delta N}{\Delta t}$$

The special unit of activity is the curie (c).

$$1c=3.7\times10^{10}s^{-1}\text{(exactly)}$$

NOTE: In accordance with the former definition of the curie as a unit of quantity of a radioactive nuclide, it was customary and correct to say: "Y curies of P–32 were administered . . . . ." It is still permissible to make such statements rather than use the longer form which is now correct: "A quantity of P–32 was administered whose activity was Y curies."

(23) The *specific gamma ray constant* (Γ) of a gamma-emitting nuclide is the quotient of $l^2\frac{\Delta X}{\Delta t}$ by A, where $\frac{\Delta X}{\Delta t}$ is the exposure rate at a distance l from a point source of this nuclide having an activity A and Δ has the meaning indicated in section 4.A.

$$\Gamma=\frac{l^2\Delta X}{A\Delta t}$$

Special units of specific gamma ray constant are $Rm^2h^{-1}c^{-1}$ or any convenient multiple of this.

NOTE: It is assumed that the attenuation in the source and along l is negligible. However, in the case of radium the value of Γ is determined for a filter thickness of 0.5 mm of platinum and in this case the special units are $Rm^2h^{-1}g^{-1}$ or any convenient multiple of this.

TABLE 4.1. *Table of quantities and units*

No.	Name	Symbol	Dimensions[a]	Units		
				MKSA	cgs	Special
4	Energy imparted (integral absorbed dose).		E	J	erg	g. rad.
5	Absorbed dose	D	EM^{-1}	J kg^{-1}	erg g^{-1}	rad.
6	Absorbed dose rate.		$EM^{-1}T^{-1}$	J kg^{-1}s^{-1}	erg g^{-1}s^{-1}	rad s^{-1} etc.
7	Particle fluence or fluence.	Φ	L^{-2}	m^{-2}	cm^{-2}	
8	Particle flux density.	φ	$L^{-2}T^{-1}$	m^{-2}s^{-1}	cm^{-2}s^{-1}	
9	Energy fluence	F	EL^{-2}	J m^{-2}	erg cm^{-2}	
10	Energy flux density or intensity.	I	$EL^{-2}T^{-1}$	J m^{-2}s^{-1}	erg cm^{-2}s^{-1}	
11	Kerma	K	EM^{-1}	J kg^{-1}	erg g^{-1}	
12	Kerma rate		$EM^{-1}T^{-1}$	J kg^{-1}s^{-1}	erg g^{-1}s^{-1}	
13	Exposure	X	QM^{-1}	C kg^{-1}	esu g^{-1}	R (roentgen).
14	Exposure rate		$QM^{-1}T^{-1}$	C kg^{-1}s^{-1}	esu g^{-1}s^{-1}	Rs^{-1} etc.
15	Mass attenuation coefficient.	$\frac{\mu}{\rho}$	L^2M^{-1}	m^{2}kg^{-1}	cm^{2}g^{-1}	
16	Mass energy transfer coefficient.	$\frac{\mu_K}{\rho}$	L^2M^{-1}	m^{2}kg^{-1}	cm^{2}g^{-1}	
17	Mass energy absorption coefficient.	$\frac{\mu_{en}}{\rho}$	L^2M^{-1}	m^{2}kg^{-1}	cm^{2}g^{-1}	
18	Mass stopping power.	$\frac{S}{\rho}$	EL^2M^{-1}	J m^{2}kg^{-1}	erg cm^{2}g^{-1}	
19	Linear energy transfer.	L	EL^{-1}	J m^{-1}	erg cm^{-1}	kev(μm)$^{-1}$.
20	Average energy per ion pair.	W	E	J	erg	ev.
22	Activity	A	T^{-1}	s^{-1}	s^{-1}	c (curie).
23	Specific gamma-ray constant.	Γ	QL^2M^{-1}	Cm^{2}kg^{-1}	esu cm^{2}g^{-1}	Rm^{2}h^{-1}c^{-1} etc.
	Dose equivalent	DE				rem

[a] It was desired to present only one set of dimensions for each quantity, a set that would be suitable in both the MKSA and electrostatic-cgs systems. To do this it was necessary to use a dimension Q, for the electrical charge, that is, not a fundamental dimension in either system. In the MKSA system (fundamental dimensions M, L, T, I) Q represents the product IT; in the electrostatic-cgs system (M, L, T) it represents $M^{1/2}L^{3/2}T^{-1}$.

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Recommendations[a] of International Commission on Radiological Units and Measurements (ICRU)

ICRU Report Number	Reference[b]
1	Discussion on International Units and Standards for X-ray work Brit. J. Radiol. **23,** 64 (1927)
2	International X-ray Unit of Intensity Brit. J. Radiol. (new series) **1,** 363 (1928)
3	Report of Committee on Standardization of X-ray Measurements Radiology **22,** 289 (1934)
4	Recommendations of the International Committee for Radiological Units Radiology **23,** 580 (1934)
5	Recommendations of the International Committee for Radiological Units Radiology **29,** 634 (1937)
6	Report of International Commission on Radiological Protection and International Commission on Radiological Units National Bureau of Standards Handbook 47, Washington, D.C. (1951)
7	Recommendations of the International Commission for Radiological Units Radiology **62,** 106 (1954)
8	Report of International Commission on Radiological Units and Measurements (ICRU) 1956 National Bureau of Standards Handbook 62, Washington, D.C. (1957)
9	Report of International Commission on Radiological Units and Measurements (ICRU) 1959 National Bureau of Standards Handbook 78, Washington, D.C. (1961)
10a	Radiation Quantities and Units National Bureau of Standards Handbook 84, Washington, D.C. (1962)
10b	Physical Aspects of Irradiation National Bureau of Standards Handbook 85, Washington, D.C. ([c])
10c	Radioactivity National Bureau of Standards Handbook 86, Washington, D.C. ([c])
10d	Clinical Dosimetry National Bureau of Standards Handbook 87, Washington, D.C. (1963)
10e	Radiobiological Dosimetry National Bureau of Standards Handbook 88, Washington, D.C. (1963)
10f	Methods of Evaluating Radiological Equipment and Materials National Bureau of Standards Handbook 89, Washington, D.C. (1963)

[a] Current recommendations are included.
[b] References given are in English. Many of them were also published in other languages.
[c] In preparation.