GILBERT H. FLETCHER, M.D.
Professor and Head, Department of Radiotherapy,
the University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute, Houston, Texas

TEXTBOOK OF RADIOTHERAPY

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Preface

The basic pattern of the previous editions has not been changed, being inspired by Paterson's first edition of TREATMENT OF MALIGNANT DISEASES BY RADIOTHERAPY, with its subtitle, A Practice of Radiotherapy. Quoting Paterson, *"The presentation of a definite outlook is of more value than the discussion of different principles and practices. It does not mean to imply that it is the only way, or the best, or the most correct method but may leave the reader with something concrete."*

As in the first and second editions, the major part of the TEXTBOOK is based on the practices in existence at the M. D. Anderson Hospital. The radiotherapists at other institutions who were asked to participate have followed the same pattern of presentation.

Important conceptual changes had happened between the first and second editions, published respectively in 1966 and 1973. In the third edition, emphasis is placed on the better correlation that has developed between radiobiology and the basic parameters of radiotherapy. Consequently, the section on Biologic Basis of Radiotherapy has been expanded by the present staff of the Section of Experimental Radiotherapy. Although there was some mention of the time factor in the second edition, it is now discussed in depth because of the increasing use of isoeffect formulae.

The various possibilities of combining irradiation and surgery in head and neck, breast, bladder, rectosigmoid, and soft tissue cancers are abundantly illustrated.

Because of the controversy over the place of irradiation and the claims of elective chemotherapy, the chapter on Breast Cancer has been augmented by data beyond the scope of merely describing irradiation techniques.

Serious consideration was given to use of the word *rad*. It would have been easy to make the changes in the text but it would have meant redoing many glossies. Dr. Peter Almond, Chief of the Section of Clinical Physics, has the following comments:

> *In the United States the unit for absorbed dose is rad. This stands for both the name of the unit and the symbol. When it is used as the name, the plural form may be used;*

e.g., it is allowable to give the absorbed dose as 6,000 rads. If it is used as the symbol, then only rad must be used. It is often difficult to know whether the name or the symbol is intended and by common usage rad and rads have become interchangeable. In line with this common practice, no distinction is made in this textbook and frequently rads is used when more strictly speaking rad should have been used, e.g., in labelling graphs.

I wish to acknowledge Miss Joan McCay, who edited and indexed this edition as well as the previous editions, and Mrs. Barbara Foremsky, who managed the secretarial logistics for both the second and this edition. My family and the office force, headed by Mrs. Helen Atterbury, have borne the manifestations of a temperament worn thinner by the third edition than the first two.

Gilbert H. Fletcher, M.D.

Houston, Texas

Contributing and Collaborating Authors

Jerry L. Barker, M.D.
Radiotherapist
Radiation Medicine Associates of Dallas;
Clinical Assistant Professor of Radiotherapy
Southwestern Medical School
Dallas, Texas

H. Thomas Barkley, Jr., M.D.
Associate Radiotherapist and
Associate Professor of Radiotherapy,
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Fernando G. Bloedorn, M.D.*
Chairman and Professor,
Radiotherapist-in-Chief
Department of Therapeutic Radiology,
Tufts University School of Medicine,
New England Medical Center Hospitals,
Boston, Massachusetts

Jean Bouchard, M.D.*
Professor of Radiology Emeritus
McGill University
Montreal, Canada

Robert M. Byers, M.D.
Surgeon, Department of Head and Neck Surgery,
Associate Professor of Surgery,
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

William L. Caldwell, M.D.*
Professor of Human Oncology and Radiology
Director, Division of Radiation Oncology
Wisconsin Clinical Cancer Center
University of Wisconsin
Madison, Wisconsin

Rafael C. Chan, M.D.
Assistant Radiotherapist and
Assistant Professor of Radiotherapy
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

*Deceased

Thomas E. Daly, D.D.S.*
Dental Oncologist (General Dentistry)
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute;
Professor
Restorative Dentistry
Maxillofacial Prosthetics
The University of Texas Dental Branch at Houston,
Houston, Texas

Luis Delclos, M.D.
Radiotherapist and Professor of Radiotherapy,
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Alon J. Dembo, M.B., F.R.C.P. (C)
Staff Radiotherapist
The Princess Margaret Hospital;
Lecturer in Radiotherapy,
University of Toronto,
Toronto, Canada

Helmuth Goepfert, M.D.
Associate Surgeon (Otolaryngology)
Department of Head and Neck Surgery,
Associate Professor of Surgery,
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Oscar M. Guillamondegui, M.D.
Surgeon, Department of Head and Neck Surgery,
Associate Professor of Surgery,
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Leonard L. Gunderson, M.D.
Assistant Radiation Therapist
Massachusetts General Hospital;
Assistant Professor of Radiation Therapy
Harvard Medical School,
Boston, Massachusetts

Augusto E. Gutierrez, M.D.
Medical Director Radiotherapy
St. Joseph Hospital;
Associate Clinical Professor
University of Texas Medical School,
Houston, Texas

Arthur D. Hamberger, M.D.
Associate Radiotherapist and
Associate Professor of Radiotherapy,
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

David H. Hussey, M.D.
Radiotherapist and
Professor of Radiotherapy,
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Richard H. Jesse, Jr., M.D.
Head, Department of Head and Neck Surgery, and Professor of Surgery,
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Carol Heaton Johnson, B.S., R.D.
Research Dietitian, Department of Radiotherapy,
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Manuel Lederman, M.D., D.M.R.E., F.F.R.
Radiotherapist Emeritus
Royal Marsden Hospital
London, England

Robert D. Lindberg, M.D.
Radiotherapist and
Professor of Radiotherapy
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Rodney R. Million, M.D.
Chief, Radiotherapy Division,
J. Hillis Miller Health Center;
Professor, Full Time,
University of Florida,
Gainesville, Florida

Eleanor D. Montague, M.D.
Radiotherapist and
Professor of Radiotherapy
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Carlos A. Perez, M.D.
Director, Division of Radiation Oncology,
Mallinckrodt Institute of Radiology;
Professor of Radiology, Full Time,
Washington University School of Medicine,
St. Louis, Missouri

Lester J. Peters, M.D.
Associate Radiotherapist and
Associate Professor of Radiotherapy
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Felix N. Rutledge, M.D.
Professor and Head,
Department of Gynecology
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Vincent A. Sampiere
Assistant Physicist
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Julian P. Smith, M.D.
Director, Department of Gynecologic Oncology,
Hutzel Hospital;
Professor of Gynecology
Wayne State University
Detroit, Michigan

Wataru W. Sutow, M.D.
Pediatrician and Professor of Pediatrics
Division Chief, Division of Solid Tumors
Department of Pediatrics
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Norah duV. Tapley, M.D.
Radiotherapist and
Professor of Radiotherapy
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Carl F. von Essen, M.D.
Radiotherapist
Zurich, Switzerland

Charles Votava, Jr., M.D.
Chief, Radiation Therapy
Methodist Hospital;
Associate Clinical Professor
Department of Radiology,
Texas Tech University School of Medicine
Lubbock, Texas

Kent C. Westbrook, M.D.
Associate Professor of Surgery
University of Arkansas for Medical Sciences
Little Rock, Arkansas

J. Taylor Wharton, M.D.
Alternate Head, Department of Gynecology and
Professor of Gynecology
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

J. Rodney Withers, M.D.
Radiotherapist and
Professor of Radiotherapy
Chief, Section of Experimental Radiotherapy
The University of Texas System Cancer Center,
M. D. Anderson Hospital and Tumor Institute,
Houston, Texas

Contents

1. General Considerations

Radiation Measurement and Dosimetric Practices

VINCENT A. SAMPIERE

This chapter presents the methods of dose measurement and dose calculation in radiation therapy employed at MDAH. Those practical aspects of dosimetry that are required to implement the clinical methods described in this book will be emphasized. Since the methods utilized in interstitial and intracavitary radium treatment have already been published,[25,26] the discussions relating to these subjects will be limited and most of the chapter will be devoted to the external beam techniques.

Basic Dose Determination

The radiation doses delivered by various therapy machines are measured at regular intervals with calibrated ionization chambers.

These ionization chambers have calibration factors that are directly traceable to the National Bureau of Standards (NBS). This means that the instruments have been calibrated either at NBS or at the Regional Calibration Laboratory at MDAH at the appropriate orthovoltage x-ray energy or for ^{60}Co gamma rays.

Measurement of Radiation Dose from Kilovoltage X-Ray Machines and ^{60}Co Teletherapy Units

CALIBRATION JIGS

X-ray and gamma-beam units have jigs constructed to ensure that chambers are held in identical relationship to the source each time a calibration is done. The jig is designed so that with the widest field, there is no chance of the primary beam hitting the jig itself.

TIME ERRORS

It is particularly important to determine the relationship of the exposure time to that set on the timer. The difference arises from "end-errors" at the beginning and at the end of exposures. These end-errors produce the same time error regardless of the total length of exposure. While this error is usually insignificant compared to the time required for a clinical treatment, it may be appreciable in the small time required for the exposure of a 25-R or 100-R chamber.

Time error α can be determined by a numeric method that is independent of instrument linearity. It involves exposing the instrument to a single exposure yielding a reading M_1, in time t_1, and also to several (n) shorter exposures accumulating a total reading M_2 in a total time t_2. The 2 times should be equal or approximately so. Since the exposure rates in the 2 cases are equal, then

$$\frac{M_1}{t_1 + \alpha} = \frac{M_2}{t_2 + n\alpha}$$

$$\alpha = \frac{M_2 t_1 - M_1 t_2}{NM_1 - M_2}$$

If α is positive, it is added to the timer setting; if α is negative, it is subtracted from the timer setting.

The timer error may also be determined in a graphic manner by setting a number of different small times on the clock and taking readings with a dosimeter. Results are plotted and extrapolated to 0 dose (Fig. 1–1). The slope is proportional to the exposure rate, and the intercept on the horizontal axis is the time to be added to or subtracted from the time set on the timer. The graphic method depends upon the linearity of the measuring instrument. It is best to use a least squares fit to the data to obtain the slope and intercept.

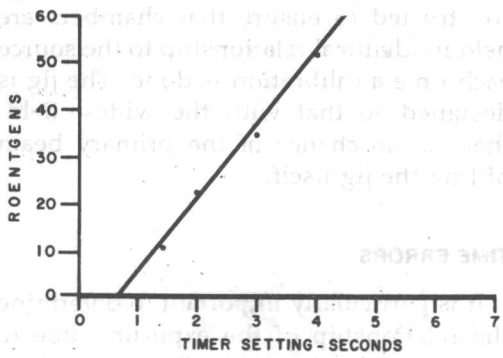

FIG. 1–1. Method of deriving timer errors. Timer errors arise from difference between true exposure time and that recorded by the timer. A plot of radiation dose versus timer setting extrapolated to 0 dose indicates the amount of time to be added to or subtracted from the time set on the timer to give the true exposure time. In the instance shown, the intercept, 0.7 seconds, should be subtracted from the time set on the timer in order to obtain the time of irradiation.

FIELD-SIZE DEPENDENCE

The radiation exposure in air varies with field size, trimmer position, and type of collimating system. Scattering from collimating parts will be different for various field settings; this can result in measured output variations of up to 15%.

In the initial calibration of a radiation therapy unit, it is usual to measure the variation of exposure for the full range of treatment fields. It is convenient to express the change in output with change in field area as a ratio to the standard calibration field area (Fig. 1–2). The exposure rate for any field is then calculated by multiplying the exposure rate for the standard field by the appropriate ratio.

KILOVOLTAGE X-RAY MACHINES

The kilovoltage x-ray machines are calibrated on a weekly basis. The weekly calibration is compared with the current standard field exposure rate in use. When a change of more than 2% is observed, a repeat calibration is made. If the change persists, new exposure rates are assigned to the machine. When the change is substantial, an estimate of the actual dose received by the patient is made and a daily calibration routine is initiated to observe whether any further fluctuations occur in the machine's performance and to determine when the output has become stable.

^{60}CO

The cobalt units are calibrated monthly and the readings compared with the decay curve. If the readings and the decay curves differ by more than 2%, an investigation is made to determine the cause. The most

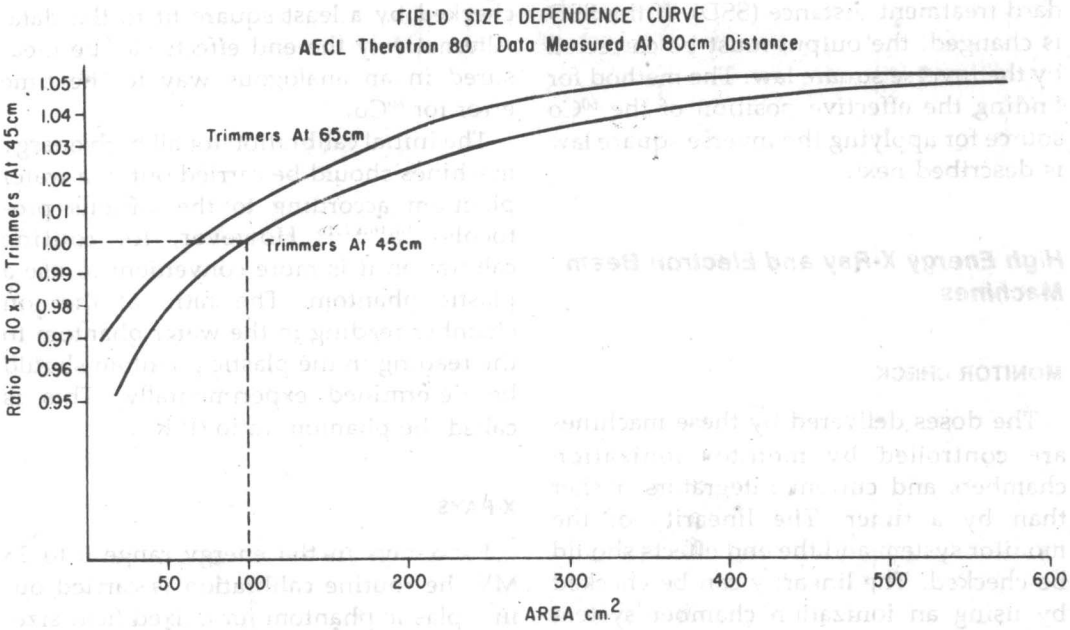

FIG. 1–2. The relative exposure rate for 2 trimmer positions and various collimator settings. Measurements are at 80-cm distance for a particular AECL Theratron 80. The standard field size in this example is 10 × 10 cm with the trimmers at 45 cm.

likely cause is a change in timer error caused by the shutter mechanism moving at a new speed. Otherwise the promulgated output of the machine is corrected to the first day of each month in accordance with radioactive decay.

With the ^{60}Co irradiators, the exposure is measured in air for a 10 x 10 cm field with the center of the secondary standard ionization chamber at the point of measurement. The secondary standard is either a Farmer instrument or a Victoreen 100-R thimble chamber. The chamber is fitted with the appropriate buildup cap (approximately 4 mm Lucite). The machines are calibrated in terms of exposure for a 10 x 10 cm field at 80.5 cm. The output is given by the following equation:

$$X = M \times N_c$$

where X is the exposure in roentgens at the position of the center of the chamber in the absence of the chamber, since the effects of its walls are included in the calibration factor; M is the chamber reading corrected for temperature and pressure with the timer error taken into account; N_c is the exposure calibration factor for ^{60}Co gamma rays assigned to the chamber by NBS with the stem correction taken into account. The given dose (D_{max}) for any field size is then given by

$$D_{med} = M \times N_c \times A \times f_{med} \times BSF \times FD$$

where D_{med} is the dose in rads in the medium; A is a factor that allows for the gamma-ray attenuation from the surface to 0.5-cm depth in tissue, A = 0.985; f_{med} is the rads per roentgen conversion for the medium (f_{med} = 0.957 rads/R for muscle); BSF is the backscatter factor for the field size being used; and FD is the field size dependence measured in air (normalized to 10 x 10 cm field). The dose at depth is then obtained by multiplying by the percentage depth dose for the appropriate field size and source to skin distance (SSD). The output is measured for a stan-

dard treatment distance (SSD). If the SSD is changed, the output must be corrected by the inverse square law. The method for finding the effective position of the ^{60}Co source for applying the inverse square law is described next.

High Energy X-Ray and Electron Beam Machines

MONITOR CHECK

The doses delivered by these machines are controlled by monitor ionization chambers and current integrators, rather than by a timer. The linearity of the monitor system and the end effects should be checked. The linearity can be checked by using an ionization chamber system with known linearity. With the ion chamber in a water phantom placed at the treatment distance, ion chamber reading versus monitor units can be taken (Fig. 1–3). Linearity and end effects can be checked by a least square fit to the data. Alternatively the end effects can be measured in an analogous way to the time error for ^{60}Co.

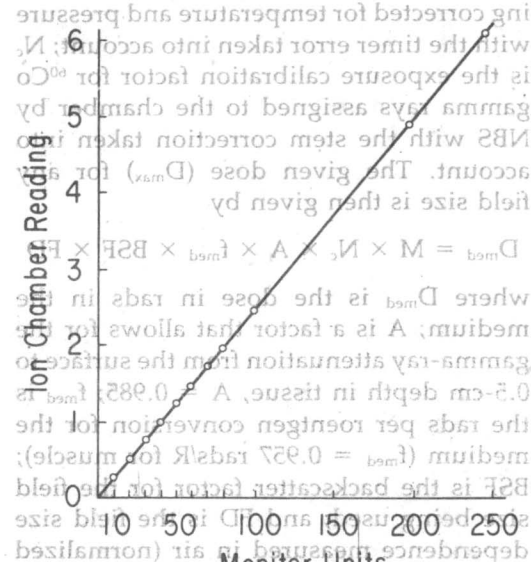

FIG. 1–3. Plot of ion chamber readings versus monitor units show the linearity of the monitor system. Corrections are made to the ionization readings for any instrument nonlinearity.

The initial calibration for all high energy machines should be carried out in a water phantom according to the various protocols.[1,13,16,17,18] However, for routine calibration it is more convenient to use a plastic phantom. The ratio of the ion chamber reading in the water phantom to the reading in the plastic phantom should be determined experimentally. This is called the phantom ratio (P.R.).

X-RAYS

For x-rays in the energy range 6 to 25 MV the routine calibration is carried out in a plastic phantom for a fixed field size. Either a Victoreen 100-R or a Farmer chamber is used. It is important that the same field size be employed for each calibration since the stem effect for some 100-R Victoreen chambers may contribute up to 5% of the observed reading, although for new chambers the stem effect is generally reduced to about 1%.

Routine checks are made of the beam's uniformity by exposing RPM film perpendicular to the central axis and at a depth in plastic beyond the dose maximum depth. An evaluation is made of the densities along the perpendicular axis of the beam's cross-section and these densities are normalized to the center field density and plotted (Fig. 1–4).

The absorbed dose will be given by

$$D = M \times N_c \times C\lambda \times P.R. \times 0.99$$

where the symbols have the same meaning as before and $C\lambda$ is the conversion factor to absorbed dose in water, which depends on energy. P.R. is the previously mentioned phantom ratio and 0.99 converts to absorbed dose in muscle from absorbed dose in water. For 25 MV x-rays, $C\lambda = 0.90$, for 18 MV x-rays $C\lambda = 0.91$, and for 6 MV x-rays $C\lambda = 0.94$.

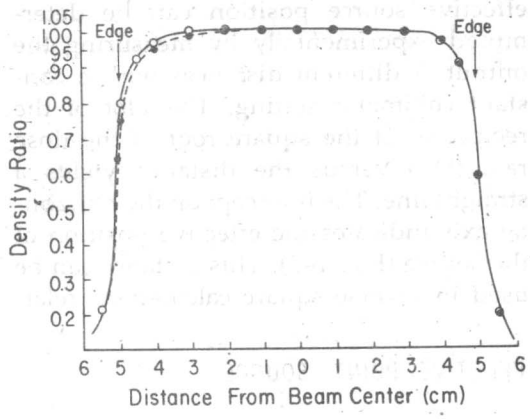

FIG. 1–4. Uniformity of dosage across a beam can be routinely checked by exposing a film, such as type M, perpendicular to the central axis of a beam in a solid phantom at a depth beyond D_{max}. Plotting densities at centimeter increments as ratios to the mid-beam density along the perpendicular axis of the beam provides a graphic display of the beam's uniformity. The radiation beam in this example is 6 MV x-ray.

This is for a standard field size. The variation in output with field size is determined during the initial calibration. Because some high energy machines show an appreciable shift in the position of the dose maximum with field size, the field size dependence may have to be measured at depth in an H_2O phantom. For the Sagittaire Linear Accelerator this was done at 10-cm depth.

ELECTRONS

For electrons

$$D = M \times N_c \times C_E \times P.R.$$

where C_E are the electron beam factors for different energies. No correction is made for water to muscle rads at this time.

The calibrations are made with a constant field size to avoid varying stem effects. For most machines the output may be checked once a week, but if the measured output varies appreciably (more than 3%), the output should be measured more frequently (at least twice a week).

Full details for the electron calibration can be found in another publication.[2]

IN VIVO DOSIMETRY

In vivo measurements for verification of an individual treatment plan may be accomplished by the use of thermoluminescent dosimeters (TLDs) if the area of interest is accessible. The placement of such dosimeters via a nasogastric (Levin) tube is an example. Access to oral areas has been achieved by placing TLD in dental stents. The TLD is available in a variety of forms and configurations.

Usually, the dose to a given point delivered by a full treatment cycle is desired. To achieve a dose proportional to the total from a cycle, a selected fraction of the given dose to each field is treated on the same day with the dosimeter in place.

A carefully carried out measurement of this type may achieve ±3% overall precision. When lithium fluoride (LiF) powder is used, the weighing of the sample and the assignment of volts per unit weight to its subsequent reading have been found to improve the accuracy of the system. The TLD is used in a relative sense, with a number of the dosimeters irradiated to a known dose. The dosimeters must either belong to a "batch" with known homogeneous response or have a known response relation to each other determined from prior irradiations.

Geometric Parameters of the Beam

Source to Skin Distance (SSD)

The source to skin distance (also known as TSD and FSD for target and focal skin distance respectively) refers to the distance, usually along the central ray, from the source of radiation to that part of the patient's skin that the beam enters.