AAPM REPORT No. 7

PROTOCOL FOR NEUTRON BEAM DOSIMETRY

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Task Group No. 18
Fast Neutron Beam Dosimetry Physics
Radiation Therapy Committee

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FOREWORD

This document "Protocol for Neutron Beam Dosimetry," is the seventh in the series of AAPM reports. It contains recommendations for the dosimetry of high energy neutron beams. These recommendations are intended to serve the immediate needs of particular neutron therapy centers and to be used as a common basis for clinical neutron basic calibration. This report was prepared by Task Group 18 under the direction of Peter Wootton and was reviewed for the Publications Committee by Stephen Graham to whom we are indebted for his careful review and comments. Issuance of such reports is one of the means that the AAPM employs to carry out its responsibility to prepare and disseminate technical information in medical physics and related fields.

John S. Laughlin, Ph.D. FACR
Chairman, Publications Committee

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1. Introduction

The purpose of this document is to provide specific recommendations the dosimetry of high energy neutron beams to be used for biological and medical applications. It is generally recognized that a common basis of neutron dosimetry is one of the essential requirements for the comparison of the clinical results obtained in radiotherapy trials at the increasing number of fast neutron therapy facilities in the United States. To date, all groups in the U.S. involved in, or planning, fast neutron therapy programs have directly or indirectly participated in the activities of Task Group No. 18: Fast Neutron Beam Dosimetry Physics, Radiation Therapy Committee of the American Association of Physicists in Medicine (AAPM). This Group was established to ensure compatibility of the physical dosimetry between centers engaged in national trials of fast neutron beam therapy, and to assist the cancer therapy effort in general by the development of a fast neutron beam dosimetry protocol based on field requirements and the best available physics experience. The Task Group is an outgrowth and extension of an ad hoc group of physicists (the Neutron Dosimetry Physics Group) from the original three fast neutron therapy facilities in the U.S. (M.D. Anderson Hospital-Texas A&M University (MDAH-TAMU), Naval Research Laboratory (NRL), and the University of Washington (UW)), which met to agree on methodology and to engage in dosimetry intercomparisons. Since the formation of the Group, three additional fast neutron therapy facilities have become operational in the U.S.: Fermi National Accelerator Laboratory Cancer Therapy Facility (FNAL-CTF), the Cleveland Clinic (CC), and the University of Chicago (UC).

The Group has drawn up a number of recommendations concerning dose calculations, dose-measurements procedures, the nature of the phantoms to be used in depth-dose measurements, and determination of displacement correction factors. It has generated studies of such items as the elemental composition of the tissue-equivalent (TE) plastic used in ion chambers, effects of phantom size on depth dose and gamma-ray contamination, and the accuracy attainable in neutron beam dosimetry. Some of these matters,

together with reports of experimental intercomparisons, have been presented at various meetings and at national and international conferences and have appeared in the literature (Smith, et al, 1975).

In addition, the Group has profited from the exchange of information and the experience gained by participating groups in the International Neutron Dosimetry Intercomparison (INDI), sponsored by the International Commission on Radiation Units and Measurements at the Radiological Research Accelerator Facility of the Brookhaven National Laboratory (Goodman, et al., 1975; Caswell et al., 1975; ICRU, 1978), and the European Neutron Dosimetry Intercomparison Project (ENDIP), performed in 1975 partly at the Institute für Strahlenschutz GSF, Neuherberg, the Federal Republic of Germany, and partly at the Radiobiological Institute TNO, Rijswijk, the Netherlands, as an effort at coordination of neutron dosimetry in Europe (Broerse and Mijnheer, 1976; Broerse et al., 1977; Broerse et al., 1978). Both of these extensive intercomparison projects have yielded the same conclusions, namely, that to reduce systematic dosimetry differences, it will be necessary to standardize the basic physical parameters and the experimental instruments and techniques employed.

The publication, “Neutron Dosimetry for Biology and Medicine," Report No. 26, by the International Commission on Radiation Units and Measurements (ICRU, 1977) has provided extensive information on methodology and physical data. It should be emphasized that efforts are being continued to improve the data base of neutron dosimetry. Some of the tables in the present protocol will therefore contain recommended values which differ from the ones quoted in ICRU Report 26 and may be subject to changes in the future. In the meanwhile, the recommendations within this document are intended to serve the immediate needs of participating neutron therapy centers and they can be used as a common basis for clinical neutron dose calibrations.

The U.S. protocol has been developed in parallel with the European neutron dosimetry protocol, in a cooperative effort with the European

Clinical Neutron Dosimetry Group (ECNEU). (Broerse and Mijnheer, 1976; Broerse and Mijnheer, 1979). The ECNEU was established in Europe by the Fast Particle Therapy Project Group under the sponsorship of the European Organization for Research on Treatment of Cancer (EORTC). The main items of the two protocols were summarized by Broerse et al. (1979). The differences in the dosimetry procedures between the European and U.S. groups will be indicated where relevant.

Any commercial products that are mentioned by name or model number in this report are included for information only; and this does not constitute a recommendation for their use by the Task Group.

2. Principles of Dosimetry in Mixed Fields

Neutron fields are always accompanied by gamma rays originating from the neutron-producing target, the primary shielding, the field-limiting or collimating system, the biological object or phantom being irradiated, and from the surroundings. The proportion of the total absorbed dose due to the photon component of the mixed neutron-photon field increases markedly with increasing depth of penetration of the incident beam in a phantom, and with the field size at a fixed depth.

Because of the differences in biological effectiveness (the magnitude of which can depend on the specific biologic end-point) of these two radiation components, it is necessary to determine the separate neutron absorbed dose, D_N, and the gamma-ray absorbed dose, D_G, of the radiation field at all points in tissue.

An evaluation of the separate dose components can be made with a single instrument, such as a proportional counter. This method requires the unfolding of the energy deposition events caused by protons and heavy ions from those caused by electrons (Caswell, 1960; Weaver, et al., 1977; August et al., 1978; Stinchcomb et al., 1980). Generally, however, two instruments, with different relative neutron sensitivities, are used for the evaluation of the component radiations (Almond, 1973; Rossi, 1956). One of the dosimeters, such as a TE ion chamber or a TE calorimeter, will have approximately the same sensitivity to neutrons and photons; the second instrument is chosen for its reduced neutron sensitivity relative to photons. The dose components at a point in the mixed field can be computed from the response of the two dosimeters by the use of the following simultaneous equations:

$$R'_T = k_T D_N + h_T D_G \qquad (2.1\text{-a})$$

$$R'_U = k_U D_N + h_U D_G \qquad (2.1\text{-b})$$

The separate neutron absorbed dose, D_N, and the gamma-ray absorbed dose, D_G, are then given by:

$$D_N = \frac{h_U R'_T - h_T R'_U}{h_U k_T - h_T k_U} \qquad (2.2\text{-a})$$

$$D_G = \frac{k_T R'_U - k_U R'_T}{h_U k_T - h_T k_U} \qquad (2.2\text{-b})$$

In these equations the subscript T refers to the TE or neutron-sensitive device and the subscript U refers to the neutron insensitive device. R_T and R_U are the quotients of the responses of the two dosimeters in the same mixed beam relative to their sensitivities (the ratio of the dosimeter response to dose measured) to the gamma rays used for the photon calibration. Similarly, k_T and k_U are ratios of the sensitivities of each dosimeter to neutrons relative to their sensitivity to the gamma rays used in calibration, and h_T and h_U are the ratios of the sensitivities of each dosimeter to the photons in the mixed field relative to their sensitivity to the gamma rays used for calibration. The values of h_T and h_U are close to unity and the simplifying assumption $h_T = h_U = 1$ can usually be made. The evaluation and numerical values of k_T and k_U for specific devices Will be discussed in Section 4.

3. Dosimetric Methods

The use of calibrated TE ionization chambers with TE gas filling is recommended as the principal practical method for measuring neutron beam tissue kerma in air and the absorbed dose in a TE phantom. This recommendation is based on the fact that homogeneous TE chambers have ken successfully used as the principal. dose measuring instrument by the neutron therapy projects in the U.S., Europe, and Span which are currently regularly treating patients. The dosimetry system of all the U.S. groups is based on the use of two sizes of commercial ion chambers manufactured with A-150 TE plastic as the chamber wall and collector material. Typically, a 1.0-cm^3 spherical chamber is used as the principal dose-rate calibration instrument, for measurements of neutron beam tissue kerma in air and total absorbed dose in a TE-liquid phantom, and a 0.1-cm^3 thimble chamber is used for spatial dose distribution measurements in a TE phantom. Preferably, the TE chambers should be used with TE gas filling because air-filled TE chambers have a larger gas-to-wall dose conversion factor than do the more homogeneous TE-TE gas chambers. There are also larger uncertainties in the physical parameters for the air-filled chambers. However, it should be noted that results of dosimetry intercomparison measurements have indicated very close agreement between TE-TE gas and TE-air chambers. TE-air chambers may be very convenient to use for daily constancy checks of machine output.

The principal beam calibration chamber should have applied to it a calibration factor for ^{60}Co gamma rays. This calibration should be directly traceable to the National Bureau of Standards (NBS), Washington. For purposes of this protocol, this means that the instrument either has been calibrated directly at NBS, or against a reference instrument or transfer standard, which has itself been calibrated at NBS. This reference instrument should be capable of performing with high precision. The calibration should be checked annually or more frequently if possible. The dependence upon a national reference standard has the advantage of providing for uniform dose calibrations.

Where feasible, it is suggested that the calibrated ion chamber method be compared with an absolute dosimeter which does not itself require

calibration in a known radiation field. A dosimeter which has been used for this purpose for high energy photon and electron beams, and neutron beams as well, is the TE calorimeter (McDonald et al., 1976; ICRU, 1977; Holt et al., 1980).

3.1 Ionization Chambers

The general dosimetric methods have received extensive discussion in the literature and will only be summarized (Burlin, 1968; ICRU, 1977).

For an ion chamber exposed in a radiation beam, the dose, $D_{g,x}$(Gy), to the gas (g) of the ion chamber is given by

$$D_{g,x} = Q_x \cdot \frac{\bar{W}_x}{e} \cdot \frac{1}{M_g} \qquad (3.1\text{-}1)$$

where Q_x, is the measured ionization charge (coul) corrected for recombination losses, W_x/e is the average energy required to produce an ion pair in the gas cavity (J/coul), e is the charge of the electron (1.60×10^{-19} coul), and M_g is the mass of the gas in the cavity (kg); the subscript x shall refer to the radiation quality (x = N and G for neutrons and photons, respectively, in the user's mixed beam (for which x = NG = T), and x = C for the calibration photon beam, e.g., ^{60}Co gamma rays).

The dose to the wall material (w) of the chamber, $D_{w,x}$, is then obtained by the following relation,

$$D_{w,x} = (S_{w,g})_x \, D_{g,x} \qquad (3.1\text{-}2)$$

where $(S_{w,g})_x$ is the gas-to-wall dose conversion factor and is commonly referred to as the effective mass stopping power ratio. For a cavity whose size is negligible compared to the mean range of the secondary charged particles generated in the wall, the value of $(S_{w,g})_x$ is given by,

$$(S_{w,g})_x = S_{w,x}/S_{g,x} \qquad (3.1\text{-}3)$$

where $S_{w,x}$ and $S_{g,x}$ are the values of the respective mass collision stopping powers of the wall and gas, the ratio of which is averaged over the corresponding equilibrium secondary charged particle

spectrum produced in the wall. For photon radiation, this charged-particle fluence will consist of electrons, while for neutrons it will, in general, consist of electrons, protons, helium nuclei, and heavy recoils such as carbon, oxygen and nitrogen nuclei (Caswell and Coyne, 1972). For a cavity whose size is not negligible in relation to the mean range of the secondary charged particles generated in the wall, it is necessary to make more detailed calculations for the values of $(S_{w,g})_x$. In these calculations, the points of origin of the secondary charged particles and their separate energy depositions in the wall and gas cavity must be considered explicitly. Values of this parameter will be discussed in Section 4.2.

Finally, if muscle tissue or tissue-equivalent phantom medium (t) is substituted for the material of the dosimeter, the dose to the tissue medium is given by,

$$D_{t,x} = K_x D_{w,x} \qquad (3.1\text{-}4)$$

where K_x is the ratio of the mass-energy absorption coefficient of muscle tissue to that of the material of the ion chamber, respectively, for the specific radiation field. For photons (x = G), $K_G = [(\mu_{en}/\rho) / (\mu_{en}/\rho)_w]_G$, where (μ_{en}/ρ) are the mass-energy absorption coefficients for photons. For neutrons (x = N), $K_N = K_t/K_w$, the ratio of neutron kerma factors. Mass-energy absorption coefficients for photons are given by Hubbell, 1977. The calculation of kerma factors for neutrons will be discussed in Section 4.3.

When the ionization chamber is exposed in a mixed radiation (neutron plus photon) beam, the total collected ionization charge is due to both neutrons and photons. Combining Equations (3.1-1) to (3.1-4) yields

$$D_{t,NG} = \frac{Q_{NG}}{M_g} \cdot \frac{\overline{W}_{NG}}{e} \cdot (S_{w,g})_{NG} \cdot K_{NG} \cdot d_{NG} \qquad (3.1\text{-}5)$$

where the subscript NG refers to the total (NG = T) mixed beam. The factor d_{NG} has been included in this expression as the chamber displacement correction factor for measurements in a TE phantom (see Section 5.3). This factor corrects for the perturbation of the radiation field in the phantom and for the phantom material that replaces the ionization chamber when it is removed from the phantom. If measurements are made in free space, this factor equals unity.

The mass of the gas in the cavity, M_g, can be determined explicitly from a calibration of the ion chamber in a ^{60}Co gamma-ray beam of known exposure (see Appendix A).

The use of Equation (3.1-5) would not appear to be very practical because it requires that the physical parameters used for the mixed beam reflect the presence of a varying fraction of photons as a function of depth in the phantom, since the photon dose fraction in the beam increases with depth. As an approximation, the parameters can be assumed to be constant with depth, or parameters appropriate to neutrons only can be applied for the mixed beam. This approximation might appear to be severe at first glance. In actual fact, it introduces a negligible error for a TE ionization chamber (see Appendix B), and such a procedure has been adopted by the neutron therapy groups in the U.S. because the "errors" introduced by its use are much less than the uncertainties in the physical parameters themselves. The total collected ionization charge can then readily be interpreted in terms of the total dose. With this approximation, and in terms of the ion chamber's calibration, N_c, in R/coul in the standard calibration photon beam (see Appendix A), the total dose to tissue in the neutron beam is given by:

$$D_{t,T} = N_C \cdot A_{w,C} \cdot f_{t,C} \cdot d_{NG} \cdot \frac{(S_{w,g})_N}{(S_{w,g})_C} \cdot \frac{\overline{W}_N}{\overline{W}_C} \cdot \frac{K_N}{K_C} \cdot Q_T \qquad (3.1\text{-}6)$$

where

$D_{t,T}$ = total dose to muscle tissue in the neutron beam (1 Gy = 10^2 rads) ($D_{t,T} = D_T = D_{NG} = D_N + D_G$);

N_c = ionization chamber ^{60}Co calibration factor (R/coul) at the calibration temperature and pressure, (0°C = 273.15°K and 760 Torr); traceable to NBS;

$A_{w,c}$ = attenuation and scattering correction factor for ^{60}Co for the TE chamber of equilibrium wall thickness (conventionally, $A_{w,c}$ = 0.985 for 0.5 g/cm^2 A-150 TE wall);

$f_{t,c}$ = ICRU muscle tissue dose-to-exposure conversion factor for ^{60}Co photons ($f_{t,c}$ = 0.00957 Gy/R);

d_{NG} = chamber displacement correction factor for dose measurements in the TE liquid phantom in the neutron beam (See Section 5.3);

$(S_{w,G})_x$ = ionization chamber gas-to-wall dose conversion factor for the secondary charged particles created in the neutron beam (x = N) or by ^{60}Co photons (x = C);

W_x/e = average energy (J/coul) required to create an ion pair in the chamber gas by secondary charged particles created in the neutron beam (x = N) or by ^{60}Co photons (x = C);

$K_c = [(\mu_{en}/\rho)_t / (\mu_{en}/\rho)_w]_C$; quotient of mass energy absorption coefficients for muscle tissue to A-150 TE plastic for ^{60}Co photons (K_c = 1.004);

K_N = neutron kerma factor ratio for muscle tissue relative to A-150 TE plastic;

Q_T = total corrected ionization charge (coul) in the neutron beam at the standard temperature and pressure for the chamber's mass calibration.

Alternatively, Equations (3.1-1) to (3.1-4) can be applied to a neutron sensitive and neutron-insensitive ion chamber pair and the dose components, and the total dose, in the mixed field can be computed by the use of Equations (2.1-a,b) and (2.2-a,b).

3.2 Materials

3.2.1 Tissue-equivalent plastic

A common electrically conductive plastic used in the construction of TE ionization chambers has been a particular muscle-equivalent formulation designated A-150*. A-150 plastic can be obtained as small chips or granules suitable for use in molding, or in various sizes of stock and custom-molded shapes for n-ore direct use. It consists of a homogeneous mixture of polyethylene,

* Supplied by Physical Sciences Laboratory, Illinois Benedictine College, Lisle, Illinois, 60532, U.S.A.

nylon (duPont Zytel 69), carbon, and calcium fluoride (Smathers et al., 1977). Based on extensive experimental and computational analyses of the end product, Smathers et al. have arrived at the elemental weight composition for A-150 plastic given in Table I (see also Goodman, 1978). Ideally, each new batch of mixture which is intended for fabrication of instrument components for which the elemental composition is critical should be analyzed thoroughly either at its source or by the user. In particular, the user should be aware that the accuracy of the measured neutron dose is very strongly dependent on the exact hydrogen content of the materiel.

The A-150 TE plastic is not identical in elemental composition to ICRU muscle tissue because of the large admixture of carbon in the plastic formulation (see Table I). Deviations from muscle equivalence will thus necessarily be reflected in a kerma factor ratio for the two mediums which is different from unity.

The density of molded A-150 plastic is 1.127 ± .005 g/cm^3 and does not appear to depend on the molding batch (Goodman, 1978). If the density is deemed to be important for a particular application, it should be measured accurately for a representative sample of the finished part.

3.2.2 Tissue-equivalent gas

Tissue-equivalent gas is recommended for use in homogeneous TE ionization chambers for measuring the total absorbed dose. The recommended formulation and composition of the TE gas are given in Table II. The composition of the gas should be verified by analysis since impurities in the gas may have a significant effect on the chamber response.

4. Physical Parameters

Variations in the results obtained in neutron dosimetry intercomparisons can be traced, in part, to differences in some of the basic physical parameters which are used to convert specific dosimeter response to tissue kerma in free air or to absorbed dose in a phantom (Almond and Smathers, 1977; ICRU, 1978; Broerse et al., 1978). These parameters are: a) the average energy required to create an ion pair in the chamber gas, W_x; b) the dose conversion factor from the gas to the chamber wall, $(s_{w,g})_k$; c) the relative neutron sensitivities, k_T and k_U, of the dosimeters used to measure the neutron and photon components of the total dose, according the Eqns. (2.1-a,b); and d) the neutron kerma factor ratio, K_N.

In order to achieve consistency in neutron dosimetry, it will be necessary to use a set of these basic parameters which is appropriate for the given neutron spectrum and which is traceable to a common source. It can be anticipated that new data on these parameters will become available from time to time and will be evaluated. The recommendations in the present document are based primarily on the field experience of the current U.S. neutron therapy groups, and partly on ICRU Report 26 (ICRU, 1977).

The values of the physical parameters are generally dependent on the neutron energy and therefore on the neutron energy spectrum at the point of interest or measurement. Neutron spectral measurements are thus a necessary source input for computation of the basic physical parameters appropriate for a given neutron spectrum. It should be noted that the neutron spectrum, and therefore the values of the physical parameters, can charge with the depth of penetration of the beam in a phantom. Measurements of neutron spectra have been accomplished by many methods, including proton-recoil counters, time-of-flight, and foil activation. Foil activation tends to give results having less energy resolution than the others mentioned. The techniques are those developed over a period of years for use in reactor and neutron physics research. All of these methods have been used to characterize the neutron beams presently used in treatment. For neutrons above 20 MeV, deficiencies in the cross section information tend to increase the uncertainty in the data with increasing energy.

4.1 Energy Required to Create an Ion Pair, $\overline{W}_x$

The parameter which is used to convert the corrected observed response of an ion chamber (the ionization charge) to energy deposited in the gas is $\overline{W}_x$, the mean energy required to form an ion pair in the chamber gas. The magnitude of this parameter depends on the nature and spectrum of the secondary charged particles, and on the chemical composition of the gas.

The values of $\overline{W}_x$ which have been used for neutron-beam dosimetry by the U.S. institutions engaged in neutron therapy are listed in Table III; the footnote to this Table indicates the methods of arriving et the tabulated values from the W-values adopted by the ICRU (1964) for high-energy electrons, alpha particles, and protons.

For methane-based TE gas, the value of $\overline{W}_N/e$ which has been widely used for a broad range of neutron spectra is 30.5 J/coul (Smith, et al., 1975). However, there are strong indications that a slightly higher value is more appropriate. Bichsel and Rubach (1978) have used recent W-data for protons, alpha particles, and ions of carbon, nitrogen, and oxygen nuclei to calculate $\overline{W}_N$ values for spherical ion chambers with walls of A-150 plastic, filled with TE gas, and irradiated with monoenergetic neutrons. Similar calculations were made by Goodman (1978), and Coyne and Goodman (1978) for the energy range 0.1 to 20 MeV. The results of these calculations are essentially in agreement and a value of 31.0 $\pm$ 1.5 J/coul for $\overline{W}_N/e$ is indicated for the neutron energy range between 1 and 14 MeV (ICRU, 1979). With this value of $\overline{W}_N/e$, and if $\overline{W}_c/e$ = 29.2 J/coul is the value for the mean energy required per ion pair for the energetic secondary electrons from the calibration gamma-rays, (Leonard and Boring, 1973), then the ratio $\overline{W}_N/\overline{W}_c$ becomes 1.06 $\pm$ 0.05. This value is about 1.5% higher than the value used previously by the U.S. groups, and IX higher than the value recommended in ICRU Report 26 (ICRU, 1977), and presently adopted by the European protocol. The most satisfying method would consist in evaluating a $\overline{W}_N$ value for each neutron spectrum. If this is not possible, then the value 31.0 J/coul suggested above can be taken for a broad neutron spectrum, with an uncertainty of approximately $\pm$ 5%.

Bichsel and Rubach (1978) have also calculated a charge of $\overline{W}_N$ with the volume of the chamber between 0.1- and 1-cm^3 of no more than 0.2%.

The value of $\overline{W}_N/e$ for TE-air chambers is less well known than for TE gas. The value which has been widely used is 35.8 J/coul (Smith, et al., 1975). A more recently indicated value for TE-air is $\overline{W}_N/e$ = 36.1 J/coul with an uncertainty of 6 - 8% (Bichsel and Rubach, 1978). This value is about 1% higher than the value used previously.

4.2 Dose conversion factor, $(S_{w,g})$

The gas-to-wall dose conversion factors for neutrons have been compromised in their accuracy due to several basic problems: a) adequate measurements of stopping power have not been made for the secondary charged particles generated by fast neutrons in A-150 plastic or in TE gas, b) the equilibrium charged particle spectrum created by the neutron beams is not well-known, and c) the range of the low-energy heavy recoils is limited. The latter results in particles existing in the various categories of "starters", "stoppers", "insiders", and "crossers" relative to the chamber cavity (Caswell, 1966), and the dose conversion factor is therefore in principle dependent on cavity size as well as on neutron energy.

Values of the gas-to-wall dose conversion factors for photons and neutrons presently used by the U.S. neutron therapy groups are listed in Smith et al., (1975) and in Table IV. The values for neutrons have been calculated assuming a Bragg-Gray chamber only, and for the indicated weighting of the secondary charged particle spectra produced by the reactions in the footnote to Table III. Values presently used by the European therapy groups can be found in Broerse et al., (1978).

More recently, Bichsel and Rubach (1978) have calculated dose conversion factors for a TE plastic-TE gas ionization chamber which approximates a Bragg-Gray cavity and for chambers of finite volume, taking into account the ionization due to primary charged particles (protons, helium nuclei, and C, N, and O nuclei) originating in the gas (i.e., starters and insiders), relative to the total ionization; they also accounted for the possible differences in the stopping powers for solids and gases (Williamson and Watt, 1972; Whillock and Edwards, 1978). The results for the 1 cm^3-volume TE plastic-TE gas chamber used currently as the principal dose calibration instruments by the U.S. neutron therapy groups indicate a value of $(S_{w,g})_N$ = 0.98 - 0.99 for the neutron beams being presently used. This new value would be about

1 - 2% lower than that used previously. It should be realized, however, that Bichsel and Rubach recommend an uncertainty of 4 to 5 percent. Therefore, until further experimental results become available there appears to be no severe need to deviate from the values used previously by the existing groups. New groups coming on line will find their values of the physical parameters from the existing sets, or from the new data, when these have been tested and evaluated.

Intercomparisons of the ionization chamber technique with calorimetry indicate good agreement between the two dosimetric methods (McDonald, 1979), when either the previous values or the newly indicated values of the physical parameters $\overline{W}_N$ and $(S_{w,g})_N$ are used, since it is the product of these two factors which appears in the expression for the total dose (Eqn. 3.1-6). The magnitude of the dose is thus maintained invariant from that indicated previously.

The newly indicated value for the 1 cm^3-volume TE plastic-air chamber is $(S_{w,g})_N = 1.18$ (± 5%), (Bichsel and Rubach, 1978); this is also not much different from that used previously.

4.3 Neutron kerma factor ratio, K_N

The relevant quantity for medical and biological applications of fast neutrons is the absorbed dose in tissue. However, it should be kept in mind that the elemental composition of tissue depends on the tissue type and this is often not well defined in a clinical situation. Absorbed dose measurements are made with instruments which are usually only approximately "tissue-equivalent" and hardly ever have the exact composition of the tissue in which the kerma or absorbed dose is desired. This is the case for A-150 tissue-equivalent plastic and ICRU “muscle” tissue (ICRU, 1964), whose composition, in general practice, is taken as a suitable approximation for most soft tissues (Table I). If the neutron energy spectrum is known at the point of measurement, then the appropriate neutron kerma factors (quotient of neutron kerma by fluence) can be used to determine the kerma or absorbed dose in the tissue from the measured kerma or absorbed dose in the instrument. The appropriate conversion factor is the ratio of neutron kerma factors in the two media. Because of the differences in the oxygen and carbon neutron cross sections, the differences in the oxygen and carbon

content of TE plastic and muscle tissue result in a muscle TE plastic (A-150) kerma factor ratio, K_N, which deviates from unity. This ratio is approximately 0.95 (Table V) for the neutron therapy beams which have been used for treating patients in the U.S. (Smith et al., 1975).

It is recommended that the kerma factor ratios be calculated on the basis of the kerma factors published in ICRU Report 26 (ICRU, 1977) for neutron energies up to 30 MeV. For neutron energies of more than 30 MeV kerma factors have been calculated by Alsmiller ard Barish (1976), and by Wells (1978). The calculations of the kerma factor ratios will obviously be dependent on the neutron energy spectrum at the point of measurement and attempts should be made to obtain information concerning the energy spectrum for different irradiation configurations (e.g., with beam filtration) and for different depths in a phantom (Wells, 1978). The uncertainty associated with the kerma factor ratio increases with neutron energy beyond 20 MeV because of the sparsity of neutron cross section information in the higher energy range.

4.4 Relative neutron sensitivities, k_T and k_U

The neutron and photon components of absorbed dose in the mixed beam can be assessed, in accordance with Eqns. (2.1-a,b) and (2.2-a,b), from measurements with a TE dosimeter and with a non-hydrogenous dosimeter which is relatively insensitive to neutrons.

The relative neutron sensitivity, k_T, of the TE dosimeter is given by

$$k_T = \frac{\overline{W}_C}{\overline{W}_N} \cdot \frac{(S_{w,g})_C}{(S_{w,g})_N} \cdot \frac{K_C}{K_N} \qquad (4.4\text{-}1)$$

where K_C and K_N are the kerma factor ratios for tissue to chamber wall material, for the calibration gamma rays and for neutrons respectively.

The relative neutron sensitivity, k_U, of the neutron insensitive dosimeter cannot, in general, be reliably calculated because existing data are fragmentary. The values of k_U vary with the neutron energy, and consequently, with the depth of measurement in a phantom. The relative neutron sensitivity for a number of specific neutron insensitive devices for a range of neutron energies, is presented in ICRU 26. The values in ICRU 26

have been supplemented by more recent measurements (Waterman et al., 1979; Mijnheer et al., 1979). Independent experimental methods have been reported by Attix et al. (1975) and Ito (1978) which directly determine k_u for ionization chambers in neutron beams having broad energy distributions. Care must be exercised in using the lead filtration method of Attix et al. to assure that the background radiation is invariant under the three beam conditions required (Waterman et al., 1977). In using the LET spectrum method of Ito, care must be exercised to ensure exact linearity of the pulse height analysis electronic system over a dynamic range of five decades, since small departures from this exact linearity result in large uncertainties in the gamma dose (Stinchcomb et al., 1979).

An experimental method developed by Kuchnir et al. (1975) determines the relative neutron sensitivity as a function of neutron energy. Neutron sensitivity functions were recently obtained by this method over the neutron energy range from 1 to 50 MeV for C-CO_2, Mg-Ar, and TE-TE ion chambers (Waterman et al.., 1979). Using such functions, the average value of $\overline{k_u}$ or $\overline{k_T}$ for any neutron beam in this energy region can be obtained by,

$$\overline{k_U} = \frac{\int k_U(E)\,\phi\,(E)K(E)dE}{\int \phi(E)K(E)dE} \qquad (4.4\text{-}2)$$

where ϕ(E) is the neutron differential spectrum of fluence and K(E) is the tissue kerma factor. If these data are used it is recommended that both $\overline{k_u}$ and $\overline{k_T}$ be determined by use of Eqn. (4.4-2) and that the quotient of $\overline{k_U}/\overline{k_T}$ be multiplied by the value of k_T obtained by calculation. This recommendation is made to compensate for a probable systematic error in the experimental data which is suggested by the fact that the experimental values for k_T(E) are systematically lower than calculated values. The uncertainty in the ratio $\overline{k_U}/\overline{k_T}$ is estimated to be $\pm$ 10%.

A Mg-Ar chamber or miniature G-M counter should be used in preference to the C-CO_2 chamber for determination of the photon dose component. This recommendation is made in view of the lower neutron sensitivity of the

Mg-Ar chamber, the different saturation characteristics of the graphite chamber to neutrons and photons (Maier and Burger, 1978), and the anomalous response obtained with the graphite chamber due to the diffusion of gas through the walls (Attix, et al., 1978; Maier and Burger, 1978; Pearson, et al., 1979).

It should be recognized that ionization chambers constructed with the same wall and gas materials, but of markedly different size, configuration, gas pressure, or shield construction, may have different relative neutron sensitivities. It should also be recognized that measurements of k_u made in air cannot be applied to determine the photon dose component in phantom unless the walls of the chamber are thick enough to stop the most energetic recoil protons generated in the phantom.

5. Calibration and Absorbed Dose Specification

5.1 Calibration

5.1.1 Gamma-ray calibration

The use of Equations (3.1-5) and (3.1-6) to determine the absorbed dose in the neutron beam requires that a ^{60}Co calibration factor, N_c, traceable to NBS, be available for the ion chamber that is to be used for the neutron beam absorbed dose calibration. This ^{60}Co calibration factor should also be verified routinely at each institution by means of a local transfer standard which is traceable to NBS. Two types of such transfer standards have been maintained at neutron institutions: (1) an ion chamber which has been calibrated at NBS or at a Regional Calibration Laboratory*; or (2) a radioactive source such as ^{137}Cs, which has been standardized with respect to an ion chamber with a ^{60}Co exposure calibration factor traceable to NBS. An ion chamber maintained as a transfer standard should not be used for routine neutron calibrations.

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* Exposure calibration factors for ^{60}Co gamma rays are at present provided by the NBS and the three Regional Calibration Laboratories (RCL) accredited by the AAPM at Cleveland, Houston, and New York. The NBS uses a spherical, graphite ionization chamber to determine the in-air exposure fron a ^{60}Co source, and the methods and procedures have been described by Loftus and Weaver (1974). The three RCL's are referenced to the NBS by transfer-grade ionization chambers, and intercomparisons between the RCL's and the NBS take place at approximately yearly intervals. For the purposes of clinical dosimetry, the calibrations provided by RCL's are equivalent to those provided by the NBS. Exposure calibration factors are expressed in R/coulomb for ionization chambers not supplied with an electrometer. The factors are referenced to the chamber response at normal temperature and pressure (NTP), ie., 0°C and 760 Torr. It is important to recognize that the exposure obtained from the product of the corrected chamber response and the. calibration factor is that which exists in the absence of the chamber, ie., when the chamber no longer perturbs the radiation field.

The following considerations apply to gamma-ray calibration of ion chambers to be used as neutron dosimeters.

a) Position of measurement. The geometrical center of the chamber should be taken as the point of measurement. A build-up cap of the same material as the wall of the chamber should be added if the wall thickness of the dosimeter is not sufficient to achieve charged particle equilibrium.

b) Chamber orientation. The chamber should be oriented at both the standardization laboratory and at the home institution so that the radiation enters the chamber from a specific direction (designated, e.g., by the serial number marking on the chamber stem or on the terminal connector block).

c) Use of a radioactive source as a local standard. If a radioactive source with a reentrant cavity, such as ^{137}Cs, is used as a local standard, a correction must be applied to allow for source decay from the time the source was standardized. The standardization of such a radioactive source in terms of a ^{60}Co calibration factor will only apply to ion chambers of the same type as that used to transfer the ^{60}Co calibration factor from the standardization laboratory.

d) Calibration in a local ^{60}Co therapy unit. The measurement should be performed in sir with the ion chamber centered at the machine isocenter (typically 80 cm or 100 cm). The chamber should be oriented such that its stem is perpendicular to the beam and to one side of the field edge. Stem scattering effects should be investigated and appropriate corrections made, if necessary. The field size should be the same as that used for calibration of the unit with the transfer standard. Where applicable, corrections for source decay and timer errors should be applied.

d) Temperature and pressure correction. The chamber response should be corrected for the specific temperature T_c, and pressure P_c, usually 0°C and 760 Torr, for which the chamber calibration factor is valid. The correction factor is given by

$$K_{TP} = \frac{273.15+T_M}{273.15+T_C} \cdot \frac{P_C}{P_M} \qquad (5.1\text{-}1)$$

where the subscript M denotes the measurement conditions.

It is important to remember that the temperature of the gas in the ionization chamber will not be the same as the room temperature, which is actually measured, unless adequate time is allowed for the chamber and gas supply with which it may be used to come to room temperature.

For measurements with a gas-flushed ion chamber, the chamber should be operated at low gas flow (typically around 5 ml/min) where it is insensitive to minor flow variations. Each chamber should be evaluated for that flow which results in a fiat response vs. flow region and the chamber should be calibrated using that flow rate. The operator should ensure that there are no leaks in the gas supply to the sensitive cavity and that the exhaust port is unrestricted such that the proper gas mixture is achieved in the sensitive cavity and that the cavity pressure is very nearly the same as the measured ambient air pressure.

It is also important to use a calibrated thermometer, capable of being read with an uncertainty of ≤ 0.2°C. The same is true for the barometer, which should be capable of measuring the air pressure with an uncertainty of ≤ 0.5 Torr.

If room air is used, appropriate corrections should be applied for gas density using Eqn. (5.1-1).

f) Ionization charge. Electrical leakage and noise should be investigated with full bias voltage and no incident beam.

Charge saturation correction factors should be determined experimentally by extrapolation of the curve of the reciprocal of the response, I/Q, versus $1/V^2$ to infinite bias voltage, V. The correction factor, K_{sat}, which will be dependent on the high voltage used, the size and dimensions of the chamber, and the absorbed dose rate in the gas, should be applied to the

collected ionization charge. Polarity effects should be checked by reversing the bias voltage and averaging the two results, if necessary.

A number of measurements of the charge-collection rate (exposures) should be made in order to ensure that the chamber conditions have come to equilibrium. The electrometer employed should be periodicaly checked with a charge source whose uncertainty is $\leq 1\%$.

g) Chamber factors. It is recommended that the values of the chamber factors following Equation (3.1-6) be used for calculating the absorbed dose from exposure in ^{60}Co.

h) Uncertainties in gamma-ray calibration:

Source		Uncertainty (%)
Ionization charge	Q_C	0.2
Charge saturation correction	$K_{sat,C}$	0.2
T and P correction	K_{TP}	0.1
Ratio of mass energy absorption coefficients	K_C	0.2
Wall attenuation	$A_{w,C}$	0.5
Dose conversion factor	$(\bar{S}_{w,g})_C$	1.0
Average energy per ion pair	W_C	
TE/TE gas		1.0
TE/Air		0.4
Exposure calibration	N_C	2.0
		RMS 2.5

5.1.2 Neutron beam calibration

Considerations regarding the procedures for neutron beam calibration are similar to those for the chamber's gamma-ray calibration. Some additional considerations include the following:

a) Measuring point. For the assessment of neutron tissue kerma in free sir, the effective measuring point should also be the geometrical

center of the chamber. The chamber should be positioned at the isocenter or at the nominal source-to-skin distance (SSD) of the therapy unit.

b) Wall thickness. The equilibrium wall thickness for neutron beams will generally not be the same as that for ^{60}Co calibration photons. The depth of maximum dose varies from about 2 mm for d(16)Be to 1.4 cm of A-150 plastic for p(66)Be(49) neutron beams. Assessment of neutron tissue kerma in free air will require corrections for wall attenuation and scattering. It is recommended that these corrections be based on the measured attenuation length for the neutron beam in which the chamber is employed.

c) Beam calibration. The primary method of neutron beam dose calibration shall be via absorbed-dose measurements at depth in the primary TE-liquid phantom. The phantom should be located with its beam entrance surface at the nominal SSD of the therapy unit. The center of the chamber should be positioned at a reference point, which in analogy to photon beams, can be chosen at 5-cm depth along the central axis of the beam.

d) Temperature and pressure correction. The chamber response should be corrected for the specific temperature and pressure for which the chamber calibration is valid (Eqn. 5.1-1). Adequate time should be allowed for the TE-liquid phantom to come to room temperature. For gas-flushed chambers, the gas flow rate should be the same as for the in-air photon calibration and the exhaust port should be unrestricted such that the cavity pressure is very nearly the same as the ambient air pressure.

e) Beam output. It is useful to have a method of standardization of the "output" of neutron sources. This procedure is discussed in Section 6.

f) Ionization Charge. The ion chamber saturation characteristics Will generally be different for the neutron beams and for the gamma rays used for the calibration radiation. Therefore, the saturation correction factor must b-s experimentally determined for the neutron beam as well. Charge saturation correction procedures are discussed by Boag (1966), and by Ellis and Read (1969).

g) Uncertainties in neutron-beam calibration:

Source		Uncertainty (%)
Ionization charge	Q_T	0.2
Charge saturation correction	$K_{sat,N}$	0.2
T and P correction	K_{TP}	0.1
Neutron kerma factor ratio	K_N .	2
Displacement correction in phantom	d_{NG}	1
Dose conversion factor:	$(S_{w,g})_N$	
TE/TE gas		4-5
TE/Air		4-5
Average energy per ion pair:	$\overline{W}_N$	
TE/TE gas		5
TE/Air		6-8
	RMS (TE/TE gas)	5-7
	(TE/Air)	7-8.5

5.2 Dose specification

Because of their significantly different biological effectiveness, the separate neutron and gamma-ray dose components, D_N and D_G respectively, of fast neutron beams should be determined as accurately as possible at all positions of relevance to their clinical application, and this should be done for different field sizes and irradiation conditions.

However, because of the uncertainties in the determination of the relatively small photon dose component leading to resultant overall uncertainties in the separate neutron absorbed dose, which can be greater than those of the total absorbed dose, it is recommended that the specification of absorbed dose for clinical applications be in terms of the total dose. That is, the total dose (neutron plus gamma) should be specified, with the relative contribution of the photon dose component, expressed, for example, as a percentage of the total dose, D_T(percent D_G).

In Europe, alternative approaches have been adopted. Either both dose components, $D_N(D_G)$, or the "total effective dose," $D_E = D_N + D_G/\tau$, where τ is a weighting factor indicating the efficiency of the neutron dose component versus the gamma-ray component for relevant effects on tumors and normal tissues, are quoted.

Depth-doses should be reported as measured in 1.07 g/cm^3 density TE liquid phantom, without conversion to unit density medium.

5.3 Displacement correction

For absorbed dose specification as a function of depth and/or position in an extended tissue-equivalent phantom, the analyses of dosimetric measurements with ionization chambers must account for the net displacement of the phantom material brought about by the introduction of the dosimeter. A displacement correction should be applied to compensate for the alteration of scattering and for the decreased attenuation of the incident radiation in the void volume and the increased attenuation in the more dense chamber walls (compared with TE liquid), caused by the displacement of the phantom material by the ion chamber when it is introduced into the dosimetry phantom.

It is recommended that the multiplicative displacement correction factors of 0.970 (for 1-cm^3 volume) and 0.989 (for 0.1-cm^3 volume), suggested by measurements of Shapiro et al. (1976) with air-filled EG&G* spherical IC-17 and IC-18 ion chambers respectively, be used. One then obtains the dose in the homogeneous phantom at the same location in the absence of the chamber. This multiplicative displacement correction factor can only be applied on the descending portion of the depth dose curve at some point beyond the depth of dose maximum. It has been found to have no significant dependence on neutron beam field size.

In Europe, where detectors of significantly different volumes and shapes are in use, it has been found more advantageous to account for the displacement correction by stating the effective measuring point as a certain fraction of the radius of the gas cavity of the ionization chamber

*EG&G, Goleta, California, formerly manufactured spherical and thimble TE ion chambers. At present, Far West Technology, Goleta, California, is producing virtually identical chambers.

upstream of the chamber's geometrical center (Zoetelief et al., 1979). This method of defining an effective chamber center is more general and is not limited to the monotonically decreasing portion of the depth dose curve. The uncertainty in these two displacement correction methods is probably on the order of 1% at present, and the differences in the corrections by the two methods indicate the importance of using small ionization chambers.

5.4 Phantoms

5.4.1 Primary phantom

The primary consideration in the selection of a phantom material for neutron therapy is that the neutron absorption and scattering properties of the material should be similar to those of muscle tissue over the broad range of neutron energies used in neutron therapy. Secondary considerations are that the material should be well-defined, stable, and the ingredients readily available.

Water has proven to be a suitable liquid for photon and electron beams, and has, for these reasons, been adopted by the European groups as the standard material for neutron beams as well. However, water is neither tissue-equivalent in terms of its elemental composition nor in terms of its density. Its neutron absorption and scattering properties cannot be expected to be identical to those of a truly tissue-equivalent liquid over the broad range of neutron energies used in neutron therapy. For these latter reasons and because a considerable amount of depth-dose data already exists in the U.S. based on this medium, it is recommended that the fat-free muscle-equivalent standard-density liquid of Frigerio et al. (1972) be used as the standard tissue-equivalent liquid medium for neutron beam dose calibrations in phantom. The canposition of this liquid mixture is exactly equivalent to ICRU muscle tissue in the major elemental ingredients C H N O (ICRU, 1964), and it reproduces the relative abundance of the trace elements of muscle tissue as wall. Its measured density of 1.07 g/cm^3 compares with the density of fat-free muscle, which has been determined to be 1.066 ± .003 g/cm^3 at 37°C (Allen, 1959). The composition and formulation of the Frigerio liquid mixture are listed in Table VI. The necessary chemicals to be used in its preparation should be of reagent grade

certified by A.C.S., and of known assay. They can be obtained from a general chemical supply house. Because of uncertainties in the composition and purity of tap water in different geographical locations, it is recommended that distilled water be used for this ingredient. The liquid mixture is quite resistant to bacterial growth and experience has shown it to be quite stable and not unpleasant to use. However, the liquid should be stored in a capped container when not in use and its density should be checked periodically.

Tissue-equivalence can be achieved accurately by the Frigerio liquid. However, there is no experimental evidence for the absolute necessity of the trace elements reproduced by the Frigerio liquid. An alternate tissue-equivalent liquid which reproduces the important elements C, H, O, and N, and the density of muscle is the three-component formulation of Goodman (1969). The composition and ingredients for this solution are also listed in Table VI. Although this liquid is considerably easier to prepare than is the Frigerio liquid, it has been found to be somewhat susceptible to bacterial growth and not as pleasant to use. In any case, it is recommended as a possible alternative and acceptable TE liquid medium for the primary phantom.

The phantom container should be made of a plastic material such as perspex (Lucite) or polystyrene and the walls should be made thick enough (about 6 mm) to prevent flexing of the surfaces. A thinner window (about 3 mm) should be provided at the entrance surface for horizontal beams so that ion chamber measurements at shallow depths are not significantly perturbed by the presence of the wall. The container should be deep enough to provide maximum backscatter and large enough in cross-sectional area to provide about a 5-cm margin about the largest field size for measurements of beam isodose contours. Generally, a 30 x 30 x 30 cm^3 container will be large enough to eliminate phantom-size effects.

5.4.2 Secondary Phantoms

Tissue-equivalent plastic and acrylic plastic phantoms can be used to supplement the primary phantom for source output consistency checks, and for intercomparison of ion chambers or other dosimetry systems. Solid plastic

phantoms of this type can provide the neutron absorption and scattering conditions closely approximating those of the primary TE-liquid phantom and, at the same time, the added advantages of exact reproducibility of detector positioning, and speed and convenience of handling. Solid plastic phantoms can also be used as a scattering medium for radiobiological studies. However, it is specifically recommended that an overlying layer of the appropriate thickness of TE plastic be used as the dose buildup material for radiobiology studies if the use of TE liquid is not feasible.

6. Beam Output

The primary method of neutron beam dosimetry should be absorbed-dose measurements at a depth in a TE-liquid phantom. However, it is often convenient and useful to have an easily reproducible method of intercomparing dose rates between institutions. In addition, a convenient method is required for daily checking the constancy of the dose monitor unit on the neutron source.

Measurements of neutron tissue kerma rate in free air performed with ionization chambers have been used extensively for this purpose. however, this method may not be a good choice for the higher energy neutron beams proposed for therapy because of the larger uncertainties associated with the wall attenuation correction for the ion chambers' necessarily greater wall thickness to achieve charged-particle equilibrium in the higher-energy beam. An alternate method would be to measure the response of a calibrated ion chamber placed at a known depth in a solid plastic secondary phantom, and irradiated under specified conditions. A 15-cm cubs of acrylic plastic having provisions for inserting an ionization chamber at a 10-cm depth has been used for this purpose at FNAL-CTF and at TAMVEC (Graves, 1979; and Awschalom, 1979). The phantom should be placed in the beam such that the ion chamber is located at the isocenter (SAD) or at the nominal source to skin distance (SSD) for the therapy unit. A 10 x 10 cm^2 field size at this distance should be used to irradiate the phantom.

The absorbed dose at a depth in the primary phantom can be related to this output measurement.

7. Beam Definitions

In order to facilitate the comparison of biological and clinical results obtained with different neutron sources, it is recommended that the following definition of field size and neutron beam description convention be adopted. In addition, all information relevant to the irradiation conditions should be specified.

7.1. Field size

The field size can be significantly expressed only in terms of the dose distribution achieved in a TE-liquid phantom. It is recommended that field size be defined by the intersection of the 50% dose-decrement lines with a plane normal to the beam central axis at the depth of maximum dose. The field size so defined will generally be a few millimeters larger than that defined by the intersection of the 50% dose-decrement lines with the phantom surface.

7.2 Beam description

A complete description of the neutron beam within a phantom would include the neutron energy spectrum as well as a spectrum of the lineal energy transfer (LET) of the secondary charged particles produced. The initial energy spectrum obtained from the fusion of deuterium and tritium (D-T neutrons) is monoenergetic; however, the neutron spectrum obtained from cyclotrons by accelerating either deuterons or protons onto a beryllium target is quite broad. These spectra can charge with the depth of penetration of the beam within the phantom and it would be desirable to have data on the neutron spectrum and the LET spectrum at various locations in the penumbral region and at a depth in a TE phantom. However, this detailed information has not been necessary for specificaton of the clinical dose. As a minimum, the principal factors affecting the neutron spectrum incident on a phantom should be specified. For accelerator-produced neutron beams, these are: the accelerated particle and its energy, the target material and

its thickness, and the thickness and material (usually polyethylene) inserted in the beam to remove the low energy neutron portion of the spectrum to increase the depth of dose penetration. The notation conventionally used is, p(66)Be(49) 6-cm polyethylene, for the case where 66 MeV protons are incident on a Be target which is 49 MeV thick and the resultant neutron beam is filtered by 6-cm of polyethylene. Knowledge of the accelerated charged particles, their energy, the target material, and the primary filter used in the neutron beam is of value for estimating the approximate depth-dose of the neutron beam.

8. Beam Monitoring

The neutron beam intensity should be monitored continuously. For cyclotron-produced neutron beams, the accelerated charged-particle beam current on the neutron target should be measured. It is desirable to have a dual transmission ionization chamber in the neutron beam. The integrated current in the transmission monitor ionization chamber should determine the total exposure. As with photon or electron therapy beams, it may be desirable to adjust the sensitivity of the monitor ionization chamber to correspond to a particular dose or dose rate at the equilibrium depth in a TE phantom, for a known field size and for a fixed source-skin distance. Saturation of the collection current should be specifically checked. If the chamber is not sealed, a convenient procedure for correction for ambient conditions should be established.

A calibration check of all monitoring detectors should be made regularly. If more than one monitor system is used, the ratio of the monitor readings will give an indication of unsatisfactory performance.

It is essential that linearity and dose-rate independence be estab lished as well as the relationship between monitor response and absorbed dose and beam output with field size. The temporal variation of background reading in the monitor chamber should be established and corrections applied where appropriate.

9. Dosimetry Intercomparisons Between Institutions

Institutions engaged in neutron beam therapy should periodically carry out neutron dosimetry intercomparisons. The purpose of these intercomparisons is to verify the institutions' conformity to quantitative dosimetry standards and to serve as a review of the procedures and techniques employed for neutron dosimetry.

The intercomparison measurements should include the total absorbed dose in the neutron beam, as well as tissue kerma in free air or beam output. The use of a calibrated tissue-equivalent ionization chamber is recommended for this purpose. A dosimeter which is relatively insensitive to neutrons can be used, in combination with the TE ion chamber, to determine the photon absorbed dose component in the neutron beam. An alternate absolute dosimetry technique such as calorimetry can be compared with the ionization chamber technique.

Specific parameters that should be intercompared are:

a) Calibration of the charge measurement of the electrometer employed (measurements of charge, Q = CV, where C = capacitance and V = bias volts, can be easily made to better than 0.1 % accuracy with air capacitors and a digital volt meter);

b) Measurement of the photon calibration of the ion chamber with a gamma-ray source whose calibration is traceable to the NBS;

c) Measurement of the ion chamber's response in the neutron beam (this measurement should be in air or in the plastic phantom used for beam output determinations);

d) Measurement of the absorbed dose at 5-, 10-, and 15-cm depths in the TE-liquid phantom.

The ion chamber's response in air or in the plastic phantom in the neutron beam can be used to calculate the tissue kerma in free air or the beam "output", respectively. Absorbed dose at depth should be determined from measurements in the TE-liquid phantom. A common set of physical parameters appropriate to the neutron beam being intercompared should be used.

Standardized measurement and correction procedures, as outlined in the preceding sections of this document, should be used. All measurements should state relevant uncertainties (one standard deviation) and the distribution of results. It is important to ascertain if significant systematic differences exist in the results.

Documentation of the intercomparison measurements should include a brief description of the facility and neutron beam characteristics, measurement techniques, the dosimeter response, and the physical parameters appropriate to the radiation field which are used to convert the dosimeter response to tissue kerma in free air and to the absorbed dose in the TE-liquid phantom.

TABLE I

Percent Elemental Composition, by Weight, of A-150
Tissue-Equivalent Plastic Compared to ICRU Muscle Tissue

Element	ICRU Muscle[a]	A-150 Plastic[b]
H	10.2	10.2 ± 0.1
C	12.3	76.8 ± .5
O	72.9	5.92 ± .2
N	3.5	3.6 ± .2
Ca	0.007	1.8 ± .1
F	not listed	1.7 ± .1
Total	98.9	100.0 ± .5

a) ICRU (1964)

b) J.B. Smathers et. al. (1977)

TABLE II
Tissue-Equivalent Gas[a]

Percent Partial Pressure		Percent Elemental Weight	
64.4	CH_4	10.2	H
32.4	CO_2	45.6	C
3.2	N_2	3.5	N
		40.7	O

(a) Rossi, H. H. and Failla, G. (1956)

TABLE III
Values of $\overline{W_x}/e$ for Gamma-Ray Calibration
and Neutron Beam Dosimetry

Constant	Chamber/Gas	MDAH-TAMU[a], FNAL-CTF,CC	NRL[b]	UW[c], UC
$\overline{W_c}/e$ (J/coul) [d]	TE/Air	33.7	33.7	33.7
	TE/TE	29.2	29.2	29.2
	C/CO_2	32.9	32.9	32.9
	Mg/Ar	26.2	26.2	26.2
$\overline{W_N}/e$ (J/coul)	TE/Air	35.8	34.98	35.8
	TE/TE	30.5	30.5	30.5
	C/CO_2			34.9
	Mg/Ar			26.4

a) MDAH-TAMU: d(50)Be

FNAL-CTF: p(66)Be(49)

CC: d(25)Be

$\overline{W}_{NG} = 0.86\ W_p + 0.09\ W_\alpha + 0.05\ W_e$

b) NRL: d(35)Be

$\overline{W_N} = W_\alpha$

c) UW: d(21)Be(16)

UC: $d(8.3)D_2$

$\overline{W_N} = 0.85\ W_P + 0.15\ W_\alpha$

d) $\overline{W_C} = W_e$

TABLE IV

Dose Conversion Factor, $(S_{w,g})_x$

Constant	Chamber/Gas	MDAH-TAMU[a], FNAL-CTF	NRL[b]	CC	UW[c], UC
$(S_{w,g})_C$[d]	TE/Air	1.142	1.140	1.142	1.133
	TE/TE	1.001	0.995	1.001	0.994
	C/CO_2	1.009	---	---	---
	Mg/Ar	1.14	---	---	---
$(S_{w,g})_N$	TE/Air	1.157	1.174	1.164	1.188
	TE/TE	1.020	0.995	1.012	1.013

a) $(S_{w,g})_{NG} = 0.77\ (S_{w,g})_p + 0.18\ (S_{w,g})_\alpha + 0.05\ (S_{w,g})_e$ (for d(50)Be)

b) $(S_{w,g})_N = (S_{w,g})_p$

c) $(S_{w,g})_N = (S_{w,g})_p$

d) $(S_{w,g})_C = (S_{w,g})_e$

TABLE V

Neutron Kerma Factor Ratio, K_N, for
ICRU Muscle Tissue Relative to A-150 TE Plastic

Constant	UW	NRL, CC	MDAH-TAMU, FNAL-CTF	UC
K_N	0.954	0.955	0.952	0.962

TABLE VI

Tissue-Equivalent Liquids for ICRU Muscle Tissue

Composition	Percent by Weight	
Element	Frigerio[a]	Goodman[b]
C	12.3	12.0
H	10.2	10.2
O	72.9	74.2
N	3.5	3.6
Ca	0.01	----
P	0.20	----
S	0.32	----
K	0.39	----
Na	0.07	----
Cl	0.08	----
Mg	0.02	----

Formulation of Standard 1.07 g/cm^3 Density TE-Liquid

		Percent by Weight	
Compound	Formula	Frigerio[a]	Goodman[b]
water	H_2O	62.16	65.6
Calcium Phosphate, Monobasic	$Ca(H_2PO_4)_2:6H_2O$	0.06	--
Magnesium Nitrate	$Mg(NO_3)_2:6H_2O$	0.21	--
Phosphoric Acid (85%)	$H_3PO_4:H_2O$	0.60	--
Potassium Bisulfate	$KHSO_4$	1.36	--
Sodium Chloride	NaCl	0.13	--
Sodium Phosphate, Monobasic	$NaH_2PO_4:H_2O$	0.11	--
Ursa	NH_2CONH_2	7.46	7.60
Glycerol	$CH_2OHCHOHCH_2OH$	2.60	26.8
Ethylene Glycol	$HOCH_2CH_2OH$	25.30	--

(a) N.A. Frigerio et al. (1972).
(b) L.J. Goodman (1969).

APPENDIX A
Photon Calibration of Ion Chambers

The use of Equation (3.1-5) or (3.1-6) to determine the absorbed dose in the neutron beam requires that a ^{60}Co exposure calibration factor, N_C, expressed in R/coul, traceable to NBS, be available for the ion chamber that is to be used for the neutron beam absorbed dose calibration. Alternately, this exposure calibration factor can be expressed in terms of the mass of the gas in the chamber, M_g.

A transfer-grade ionization chamber (usually graphite), which has been calibrated at NBS or at an RCL, is used to determine the exposure in air in a ^{60}Co gamma-ray beam. The exposure is that which is obtained at the center of the transfer chamber when the chamber is removed. A calibration factor for the user's ion chamber is then obtained by placing it at the same location as the transfer chamber and giving it a known exposure. If Q_c (coulomb) is the chamber's electrometer reading (corrected for the various experimental conditions such as electrometer factors, charge recombination effects, stem scattering and for the standard calibration conditions of temperature and pressure for which the calibration is to be valid), the ^{60}Co exposure calibration factor, N_c (R/coulomb), for the user's ion chamber, is given by

$$N_C = \frac{X}{Q_C} \qquad \text{(A-1)}$$

This exposure calibration factor is traceable to NBS, and the mass of the gas in the ion chamber, M_g, can now be determined.

It is assumed that the chamber wall and build-up cap are of the same material and of equilibrium thickness for the ^{60}Co calibration gamma rays. It is also assumed that the chamber can be treated as a Bragg-Gray cavity surrounded by the wall material of the chamber (the medium), and that the presence of the cavity in the medium does not disturb the photon flux in the medium.

If the chamber is exposed in sir in ^{60}Co where the exposure is X, and the ionization produced in the gas is Q_c, then Equations (3.1-1) and (3.1-2) can be combined to give the dose to the wall material of the chamber,

$$D_{w,C} = \frac{Q_C}{M_g} \cdot \frac{W_C}{e} \cdot (S_{w,g})_C \qquad (A-2)$$

If the chamber is now replaced by an equilibrium mass of tissue, which is sufficiently similar to the chamber so as not to disturb the external photon flux, then the dose to the tissue is given by

$$D_{t,C} = \left(\frac{(\mu en/\rho)t}{(\mu en/\rho)w}\right)_C \cdot D_{w,C} \qquad (A-3)$$

Now, following the methods suggested by Johns and Cunningham (1969), the dose to the center of the chamber is also given by

$$D_{w,C} = f_{w,C} \cdot X \cdot A_{w,C} \qquad (A-4)$$

where $f_{w,C}$ is the exposure-to-dose conversion factor for the chamber for ^{60}CO gamma rays, and is given by

$$f_{w,C} = 0.869 \cdot \left(\frac{(\mu en/\rho)w}{(\mu en/\rho)air}\right)_C \qquad (A-5)$$

and $A_{w,C}$ is the correction factor for photon attenuation and scattering in the chamber wall and buildup cap. The value of $A_{w,C}$ depends on the thickness of material required to give electronic equilibrium, and its precise value is difficult to determine. Its value has been calculated for

a variety of commonly employed chambers and caps by a Monte-Carlo method developed by Bond, Nath and Schulz (1978). The conventionally accepted value for spherical ion chambers of A-150 TE plastic with 0.5 g/cm^2 wall thickness and gas cavity volumes of about 1 cc or leas, is 0.985, for ^{60}Co gamma rays.

If the chamber wall is replaced by tissue, then

$$D_{t,C} = f_{t,C} \cdot X \cdot A_{t,C} \qquad (A\text{-}6)$$

where it can be assumed that $A_{t,C} = A_{w,C}$ for a TE chamber.

Combining Equations (A-1) to (A-6) by equating the expressions for $D_{t,C}$ and solving for M_g, we obtain

$$M_g = \frac{1}{N_C f_{t,C} A_{w,C}} \cdot \frac{\overline{W}_C}{e} \cdot (S_{w,g})_C \cdot \left\{ \frac{(\mu_{en}/\rho)_t}{(\mu_{en}/\rho)_w} \right\}_C \qquad (A\text{-}7)$$

APPENDIX B
Determination of Total Dose Approximation

The total dose to tissue in the neutron beam ($D_{t,T} = D_T = D_{NG}$) can be written as

$$D_T = D_N + D_G \qquad \text{(B-1)}$$

where the separate dose components could, in principle, be obtained from

$$D_x = \frac{Q_x}{M_g} \cdot \frac{\overline{W}}{e} \cdot S_x \cdot K_x \qquad \text{(B-2)}$$

where the parameters are as defined in the text and $S_x = (S_{w,g})_x$ for convenience of notation.

In practice, it is not possible, with a single chamber, to separate the total ionization charge into components due to neutrons and photons in the mixed beam. Only the total charge

$$Q_T = Q_N + Q_G \qquad \text{(B-3)}$$

can be measured.

From Equations (B-1) and (B-2), the total dose can be written

$$D_T = \frac{Q_N}{M_g} \cdot \frac{\overline{W}_N}{e} \cdot S_N \cdot K_N + \frac{Q_G}{M_g} \cdot \frac{\overline{W}_G}{e} \cdot S_G \cdot K_G \qquad \text{(B-4)}$$

By substituting from Equation (B-3) for Q_N, and performing some straight-forward algebra, it can readily be shown that

$$D_T = K_N \cdot S_N \cdot \overline{W}_N \cdot \frac{Q_T}{M_g} \left[1 - (1-k_T/h_T)\frac{Q_G}{Q_T}\right] \qquad \text{(B-5)}$$

where (Equation 4.4-1)

$$k_T = \frac{\bar{W}_C}{\bar{W}_N} \cdot \frac{S_C}{S_N} \cdot \frac{K_C}{K_N} \qquad \text{(B-6a)}$$

and

$$h_T = \frac{\bar{W}_C}{\bar{W}_G} \cdot \frac{S_C}{S_G} \cdot \frac{K_C}{K_G} \qquad \text{(B-6b)}$$

Except for the factor in brackets, Equation (B-5) is similar to Equation (3.1-S) for neutrons only, ie., NG = N. The charge ratio, Q_G/Q_T, is approximately equal to the photon dose fraction (PDF) in the mixed beams and the assumption $h_T = 1$ is usually made. With this assumption, the expression in brackets becomes

$$[\ ] = 1 - (1-k_T)\,PDF \qquad \text{(B-7)}$$

and, for typical values of the physical parameters for a TE-TE gas ion chamber (Tables III, IV, and V),

$$k_T = \frac{(29.2)\ (1.00)\ (1.004)}{(30.5)\ (1.01)\ (0.95)}$$

$$= 1.002$$

and, for a photon dose fraction (PDF) of 20% in the TE phantom,

$$[\ ] = 1 - (1-1.002)\ (0.20)$$

$$= 1.0004$$

The approximation (Equation 3.1-6) thus introduces a negligible error for the determination of the total dose with a TE-TE gas ion chamber.

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