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Title ABSORBED DOSE FROM X RAYS IN SOFT TISSUE ADJACENT TO BONE MEASURED BY INDUCED CONDUCTIVITY IN POLYETHYLENE FILMS

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The X-ray dose absorbed in soft tissue located near plane bone interfaces and in parallel-slab bone cavities was measured experimentally with a polyethylene-film dosimeter.

The dose absorbed in soft tissue adjacent to plane bone interfaces was measured by placing a 12.7 micron film of polyethylene, upon which were deposited graphite electrodes, in contact with bone-equivalent disks of Shonka plastic. The dose was measured both as a function of X-ray quality and distance of the tissue-equivalent polyethylene away from the bone interface. The dose in finite soft tissue layers next to bone was also measured as a function of X-ray quality. The results indicated that the soft tissue dose in finite layers next to the bone was greater than that remote from the bone. At the lower effective X-ray energies the dose very close to the bone was considerably higher than for the higher

effective X-ray energies. At higher effective X-ray energies, however, the increased dose effect extended to a greater depth in the soft tissue. At effective energies greater than 177 keV electronic equilibrium was obtained, and the dose was the same for the adjacent and distant soft tissue.

Comparisons were made of the experimentally measured doses in soft tissue near plane bone interfaces with calculated doses for monoenergetic photons. It was observed that at effective X-ray energies below approximately 60 keV the experimental results gave higher dose values than the theoretical values. However, at higher effective X-ray energies there was good agreement between the experimental and theoretical results. It was concluded that X-ray spectral considerations were mainly responsible for the deviations at lower effective energies. The results of this study indicated that when one wishes to calculate the dose absorbed in soft tissue adjacent to plane bone interfaces at low effective X-ray energies, the theory cannot be relied upon and experimentally determined doses such as those shown herein should be used instead.

The X-ray dose in soft tissue elements interposed between parallel slabs of bone was measured in the same manner as the dose in soft tissue adjacent to plane bone interfaces. However, due to experimental difficulties in simulating this bone-tissue orientation, it was not possible to make reasonable dose measurements.

ABSORBED DOSE FROM X RAYS IN SOFT TISSUE ADJACENT TO BONE MEASURED BY INDUCED CONDUCTIVITY IN POLYETHYLENE FILMS

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INTRODUCTION

It is relatively easy to calculate or measure the dose from X rays in a soft tissue element where conditions of electronic equilibrium exist; i.e., in a soft tissue element surrounded by a semi-infinite volume of soft tissue. Similarly, the dose in an element of bone can be calculated where electronic equilibrium exists. However, if an element of soft tissue lying very near a slab of bone or inside a bone cavity of finite size is considered, then the problem of calculating the dose in this soft tissue element becomes quite difficult. In these latter cases the secondary electron fluxes which originate in both materials due to interaction with the primary photon radiation must be evaluated. By considering where these electrons originate, the manner in which they lose their energy, and the locations where their energy is deposited, the dose can be calculated. Due to the complexity of these calculations certain simplifying assumptions must be made which may decrease the accuracy of the final results. In such cases it is desirable to measure the doses and compare these results with the calculated values. It was with this objective in mind that this research project was performed.

X-rays were chosen as the form of ionizing radiation to be

employed because the majority of the calculations that have been made on dose near bone have been carried out using X rays. Hence, comparisons with theory are possible. Many measurements of X-ray dose in soft tissue near bone have been made by other investigators. However, more data at different X-ray qualities and different bone-soft tissue orientations are needed to support theoretical computations, and to provide additional data for clinical application. One problem in comparing these experimental results with theory is that most of the available calculations have assumed a monoenergetic photon beam. Those calculations that have given consideration to X-ray spectra are both limited in number and accuracy by the complexity of correcting for the distribution of X-ray spectra.

For X rays produced by voltages of 100 to 300 kVp[1] the general effect on the dose to a soft tissue element near bone is to increase the energy absorption to a value higher than would be expected in similar soft tissue remote from bone. This effect is attributed to the different attenuation coefficients of bone and soft tissue in the photoelectric interaction energy range. In the Compton interaction energy range the attenuation coefficients of bone and soft tissue are very similar and little effect is noticed, thus permitting one to

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[1] Throughout this paper the term kVp will be used to denote the tube potential at which the X rays are generated. The term keV will be used consistently to indicate the effective photon energy of an X-ray spectra, unless it is otherwise specified as meaning a monoenergetic photon beam.

calculate the soft tissue dose or bone dose irrespective of their relative orientations. It is, therefore, the photoelectric process which gives rise to the increased dose phenomena.

This increased soft tissue dose is of importance since it is in the region within bone that the marrow cells of the hematopoetic system are found. The marrow cells are one of the most radiation sensitive cell types in the body. Since these cells play an important role in the synthesis of many elements of the blood it is desirable to know as accurately as possible how much energy they absorb during X-irradiation. The increased dose to the bone cells which are the living elements found within the cavities and canals of the inorganic bone matrix is also an important consideration. Damage to these cells may result in improper bone growth, repair, or maintenance processes. It is well known that excessive localized doses of X rays result in decreased bone density, bone malformation and osteogenic sarcomas. These effects are attributed to death or inactivation of the osteocytes, osteoblasts, and osteoclasts either from the direct action of the energy absorption or the indirect action of necrosis resulting from collapse of the Haversian canals and other canals and canalicules located in bone (McLean and Budy, 1964).

The increased absorption in soft tissue is limited spatially over a distance equal to the maximum range of electrons which have originated in the nearby bone. To measure the effect and describe

it physically an instrument capable of measuring dose must be used having a detecting volume considerably less than the range of the photoelectrons which are produced. A polyethylene-film dosimeter, similar to one used by Fowler (1957), was constructed which permitted measurements within the spatial limitation. The electrical conductivity of polyethylene changes in a manner which is dependent on the amount of energy deposited in it by the radiation. It is available in films down to thicknesses of ten microns. Using this dosimeter and simulating different bone-soft tissue orientations, soft tissue doses can be measured as a function of distance from bone and X-ray quality.

HISTORICAL REVIEW AND THEORETICAL CONSIDERATIONS

The problem of calculating or measuring the energy absorbed in soft tissue elements located within or adjacent to bone is one that has been the subject of thought and investigation by a number of workers (Aspin and Johns, 1963; Charlton and Cormack, 1962; Ellis, 1966; Epp, Woodard and Weiss, 1959; Fowler, 1957; Howarth, 1965; Spiers, 1949, 1951; Wingate, Gross and Failla, 1962). Semi-rigorous solutions and precise experimental measurements, however, have only been made in the past few years. Due to the complex nature of the calculations, and the inherent difficulties involved in experimental measurements, even these more recent results are not as exact as would be desired.

To introduce this subject, consideration will first be given to a description of the phenomena which give rise to the absorption of energy from a beam of X rays. The factors influencing this energy absorption will also be briefly described. With this background, the theory of bone dosimetry and the events leading to our present state of knowledge of this subject will be considered.

Energy Absorption from X Rays

X rays may deposit energy in the matter with which they interact by either photoelectric absorption, Compton scattering,

or pair production. The properties of the matter with which they interact and the energy which they possess determine to what degree these three processes will occur.

The pair production process was not a significant mechanism of energy absorption in this study since the X rays used were of energy less than 1.022 MVp.

In the photoelectric process a photon of energy less than approximately 100 keV may eject an orbital electron from an atom and in so doing the photon ceases to exist. It is, however, physically impossible for the electron to absorb all of the energy of the photon if momentum is to be conserved. Consequently the nucleus, or atom, must recoil for conservation of both energy and momentum in the process. The probability of photoelectric absorption by a particular atomic electron increases with decreasing photon energy since for low energy photons the electron is relatively more tightly bound to the nucleus and it is easier for the nucleus to recoil to conserve momentum. Similarly, electrons are more tightly bound to high atomic number nuclei than to nuclei of low atomic number. The probability of photoelectric absorption, therefore, increases with increasing atomic number of the matter with which they interact. The probability of this process occurring is usually expressed as the photoelectric cross-section, τ. The theory of quantum electrodynamics predicts the proportionality between the photoelectric

cross-section τ, the X-ray energy E, and the atomic number Z to be:

$$\tau \propto Z^5/E^{3.5} \qquad \text{(Eisberg, 1964).}$$

In the Compton scattering process, photons of energy greater than approximately 100 keV will suffer scattering collisions with atomic electrons which may be considered unbound to nuclei because of the relatively high photon energy. As a result of such collisions the photons transfer part of their energy to these electrons and are themselves degraded in energy. They do not completely disappear as in the photoelectric process. The probability of Compton scattering occurring is expressed as the Compton cross-section. This cross-section depends only on the electron content of the material with which the photons are interacting. Over a wide range of materials the number of electrons contained per unit mass remains fairly constant, and the Compton cross-section varies little for different materials. The well-known Klein-Nishina relations predict a slight dependence of the Compton cross-sections on photon energy in the energy range from 0.1 to 1.0 MeV. This is but a slight dependence as shown in Whyte (1959) and is considered insignificant in this study.

The high energy electrons which originate from the photoelectric and Compton interactions are responsible for transferring the energy of the X rays to the medium with which they are

interacting. These electrons cause further ionization and excitation of atoms by coulombic interactions. By considering the X-ray intensity incident on a small volume of the medium, the cross-sections for attenuation of these X rays, and the energy necessary to form an ion pair in the medium, the energy absorbed in the volume can be calculated if conditions of electronic equilibrium exist.

For electronic equilibrium to exist in a small volume of some material, this volume must be farther than the maximum range of the secondary electrons originating in the material away from all boundaries with different materials. If this condition is met then the ionization occurring in this volume may be considered as due only to those electrons which have originated within it. Since the density of ionization is known, direct conversion to energy absorbed in the material is possible.

If the volume of material in which the dose is to be calculated is within range of the electrons originating in a different material then electronic equilibrium does not necessarily exist in the defined volume. There may be more, or less, secondary electrons being scattered into the defined volume than are being scattered out. Therefore, the ionization within the volume cannot be considered as being due only to the secondary electrons which were produced in it from attenuation of the X rays. Since the density of ionization is not directly known, the energy absorbed in the material cannot

be calculated as easily as when electronic equilibrium exists.

It is this case of nonelectronic equilibrium that is of concern in determining the dose from X rays in soft tissue adjacent to a bone interface or within a bone cavity.

Dose in Soft Tissue Near Bone

Since the effective atomic number of bone, 13.8, is considerably higher than that of soft tissue, 7.40, the photoelectric cross-section for bone will be higher than that for soft tissue. Consequently, X rays with energies in the photoelectric range will be attenuated to a higher degree in bone than in soft tissue and the secondary electron flux arising in bone will exceed that arising in soft tissue. This results in bone absorbing considerably more energy at electronic equilibrium than soft tissue.

As the X-ray energy is increased, Compton scattering becomes the predominant attenuation mechanism. As previously mentioned, the Compton cross-sections are similar for materials of varying atomic number due to their similar atomic electron densities. Photons of energy greater than 100 keV and less than 1.022 MeV will, therefore, be attenuated about equally in both bone and soft tissue and the secondary electron fluxes arising in either bone or soft tissue will be approximately the same. The dose in either bone or soft tissue from X rays of effective energies greater than 100 keV

and less than 1.022 MeV will, therefore, be approximately the same.

If a layer of soft tissue lying near bone irradiated with X rays of energy less than 100 keV is considered, it is apparent that conditions of electronic equilibrium will not exist in it if it is located within a distance equal to or less than the maximum ranges of those electrons originating in the bone, or soft tissue. By calculating the number of electrons that arise in the bone and surrounding soft tissue, their energy and their ranges, the dose in the layer of soft tissue can be calculated if one knows the relative position of the tissue to the bone. A knowledge of the electron stopping powers of the bone and soft tissue is also necessary since the ranges of electrons in bone and soft tissue differ.

Methods of Calculating Dose in Soft Tissue Near Bone

The results of several investigations carried out during the 1930's have verified that different tissues of the body absorb energy from X rays in differing amounts depending on their chemical composition and densities. The expression relating the chemical composition and effective atomic number on which the photoelectric attenuation coefficient depends was derived by Mayneord (1937) and later experimentally verified by Spiers (1946) who measured the dose in materials of varying composition at different X-ray energies.

Spiers (1946) determined the effective atomic number of bone to be 13.8 and that of soft tissue to be 7.4. Using these values he determined that for 40 keV X rays the energy absorbed in bone is six times greater than that absorbed in soft tissue. Realizing that there is soft tissue contained in bone, Spiers (1949) became concerned with how this higher energy absorption in bone might possibly affect the dose in this soft tissue. If the soft tissue is located in bone cavities of dimensions approximately one to two microns in diameter, then the energy which it absorbs is shown by the Bragg-Gray principle to be approximately the same as the energy absorbed in the bone surrounding the cavity. There is a slight reduction in the soft tissue dose since the electron stopping power is less for soft tissue than for bone. Though the soft tissue contained in bone is usually not found in cavities of such small dimensions, there is a good possibility that it may lie sufficiently close to the bone that the energy absorbed in it will be considerably higher than the energy absorbed in that soft tissue which is remote from bone. Spiers (1949) was the first investigator to derive relations for the energy absorption in soft tissue lying near or within bone. Specifically, Spiers (1949) considered soft tissue interposed between parallel, semi-infinite slabs of bone, and soft tissue adjacent to plane bone interfaces. The parallel-slab cavity simulates the Haversian systems contained in bone, the cavities in which the osteocytes are found,

and the lamellar marrow layers found in trabecular bone. The plane interface simulates the endosteal and periosteal layers of bone cells and some types of trabecular bone.

To determine the energy absorbed in soft tissue at these two locations, Spiers (1949) made several assumptions to simplify the calculations. It was assumed that electrons are emitted isotropically from X-ray attenuation, and that these electrons travel in straight lines with constant specific ionization along their tracks. Also, the ranges of the electrons were taken to be 70 percent of their full rectified ranges. This assumption is in accordance with the theory of Lea (1962). A single energy was taken for the photoelectrons ejected from the K-shell, and a mean energy was taken for the spectra of Compton recoil electrons.

The energy absorbed in soft tissue located near bone was calculated by summing the contributions of electrons arising in both the bone and soft tissue to the ionization at the desired location in the soft tissue. These calculations were carried out for a variety of monoenergetic X-ray beams and the results plotted as ions/μ^3/ Roentgen versus distance of the soft tissue from the cavity wall, or distance away from the plane bone interface. For low energy X rays, less than approximately 60 keV, the energy absorbed in soft tissue very close to plane bone interfaces and cavity walls is quite high, approaching the high equilibrium dose absorbed in bone. This

increased dose falls off rapidly at greater depths in the soft tissue due to the limited range of the photoelectrons which are produced by low energy photons. For higher energy X rays photoelectric absorption diminishes and Compton scattering predominates. Since bone and soft tissue absorb approximately the same dose in the Compton energy region, the increased dose effect in soft tissue very close to bone is not as great. The few photoelectrons that are scattered into the soft tissue from bone do not have such limited ranges, and even though the increased dose to soft tissue is not as great, it extends farther into the soft tissue layer.

Spiers (1949) averaged the dose to soft tissue layers of varying thickness and plotted these results as soft tissue dose versus effective X-ray energy. The soft tissue dose was greatest for the smallest cavities and thinnest layers of soft tissue next to bone, and was maximized for X-ray exposures between 40 and 60 keV.

Munson (1950) corrected Spiers' (1949) calculations for a minor error in his approximation of an electron flux. Spiers (1950) acknowledged this correction but showed it to have little effect on his original results.

Wilson (1950) calculated the energy absorbed from X rays at the walls and in the center of the soft-tissue-filled, cylindrical Haversian canals. Since bone necrosis is a result of the collapse of the lumen of the Haversian canals, the energy absorbed in the

canal walls is of considerable importance. Wilson (1950), like Spiers (1949), assumed the range of electrons to be 70 percent of their full rectified range. By determining the volume of bone from which electrons could arise and reach either the wall or center of the canal, the increased dose to the soft tissue contained at these locations was estimated. This "contributing volume" method was a rough approximation to the actual increased dose effect. The results compare qualitatively with those of Spiers (1949) but are somewhat lower because this method underestimates the depth in soft tissue to which electrons arising in bone may contribute their energy. It was concluded from the results that since tumors are usually composed of soft tissue, in order to give the maximum dose to the tumor and minimum dose to the Haversian canals, one should employ therapeutic X rays of as high energy as possible.

In a continuation of his earlier work, Spiers (1951) presented tentative figures for the permissible dose in Roentgens to bone as a percentage of that which could be administered to soft tissue under similar treatment conditions. It was shown that for monoenergetic X rays of 50 keV energy the permissible exposure to bone is only 35 percent of that exposure which could be administered to soft tissue remote from bone. For 200 keV X rays the exposure has increased to 90 percent of the soft tissue exposure. The higher the X-ray energy, therefore, the larger the permissible exposure

to bone becomes. This is in agreement with those conclusions drawn by Wilson (1950). The results presented by Spiers (1951) are reproduced in the International Commission on Radiological Units and Measurements Report 10d (1963).

Spiers and Woodard (1953) found it necessary to extend the earlier calculations to include X-ray spectra in order to quantitate the results of a biological experiment on the energy absorbed in living mouse bone with theory. The energy spectra of 100, 185, and 1000 kVp X-ray beams were determined by Greening's method (1947), and from these the spectra of electrons which arise in bone were calculated. By summing the contributions of these electrons, the doses in different soft tissues contained within bone were determined. Spiers and Woodard (1953) reported that for the 1000 kVp radiation there was little increased dose to the soft tissue within bone. However, the 100 and 185 kVp radiations increased the dose to these tissues considerably over the dose to soft tissue remote from bone.

Another extension of the Spiers (1949) method of calculating dose in soft tissue near bone was made by Epp, Woodard, and Weiss (1959) who were interested in comparing the Relative Biological Effects of 250 kVp X rays and Cobalt 60 gamma rays on the bone marrow of the mouse. Adjusted values for the X-ray spectra as recommended by Spiers and Woodard (1953) were used. Unlike

Spiers and Woodard (1953), however, Epp, Woodard, and Weiss (1959) did not assume that the bone lying adjacent to soft tissue was necessarily of equilibrium thickness. This is actually the case in some trabecular bone types where very thin layers of bone and bone marrow are interleaved in a network. Using a physical model of mouse bone, the dose in various types of marrow cavities was determined and averaged to give the average dose absorbed by the mouse bone marrow during whole body irradiation. For the 250 kVp X rays the average dose to the marrow was 0.986 rads per Roentgen, while for the Cobalt 60 gamma rays the average dose was 0.94 rads per Roentgen. Had the marrow been remote from bone the doses would have been 0.967 and 0.980 rads per Roentgen respectively (Hine and Brownell, 1964). Thus the dose in marrow from 250 kVp X rays is slightly increased over the remote soft tissue value. It was pointed out by Epp, Woodard, and Weiss (1959) that this slight increase is somewhat misleading, since it is an average value. The marrow lying next to bone actually receives a much higher dose than this, but is reduced when an average effect over the total volume of marrow is computed. Therefore, many more marrow cells may be killed or inactivated in certain areas near bone than in other areas where the marrow may be some distance away from bone.

Charlton and Cormack (1962) revised Spiers' (1949) method of calculation to take into account the variation of electron Linear

Energy Transfer (LET) with energy. Monoenergetic X rays were considered to simplify the calculations. Doses in soft tissue near plane bone interfaces, in parallel-slab bone cavities, and in cylindrical bone cavities were determined. This was the first attempt to calculate the dose distribution across a cylindrical cavity since the previous calculations by Wilson (1950) had determined the dose only at the center or at the wall of the cylindrical cavity. It was shown that the dose to soft tissue elements contained in cylindrical bone cavities is generally higher than the dose to layers of soft tissue interposed between two bone slabs. Because more accurate X-ray attenuation coefficients were employed in these calculations, comparison with Spiers' (1949) results are not too meaningful. However, Charlton and Cormack's (1962) results are quite similar to Spiers' (1949) results for both the plane interface and parallel-slab geometries, being slightly higher at X-ray energies below 60 keV.

Aspin and Johns (1963) also calculated the absorbed dose in soft tissue-filled cylindrical cavities within bone. They did not consider the variation of electron LET with energy, though their method extended the theory to include actual spectra rather than monoenergetic X rays. These calculations were made to obtain theoretical results which could be compared with measurements of the cylindrical soft tissue cavity dose that were made concurrently.

A method of determining the dose in the bone marrow of

humans was given by Spiers (1963). Using a cadaver, the dimensions of the marrow interspaces in different types of trabecular bone were determined. Using the method of Charlton and Cormack (1962) the average marrow dose in the four major red marrow sites of the body were then determined. The results are presented in such a manner that if the depth dose in Roentgens to the marrow site being irradiated and the incident X-ray energy are known, then the dose factor in rads per Roentgen can be read directly from a graph. This is a very convenient method of determining bone marrow doses in both X-ray diagnosis and X-ray therapy.

Recently, Charlton and Cormack's (1962) calculations have been extended by Howarth (1965) to include the dose absorbed by spherical soft tissue cavities in bone. Account was taken of the variation of electron LET with energy. Rather than assume a mean energy for the Compton recoil electrons, their continuous energy spectra were used. Dose contributions from Auger electrons were considered and determined to have the effect of considerably increasing the dose to soft tissue lying very near to bone. Accurate tables of the geometrical functions and physical parameters necessary for calculating the dose in soft tissue at any point close to a plane, cylindrical, or spherical bone interface were presented. These tables and functions were obtained by evaluating numerical integrals for the different bone-soft tissue orientations. Though no attempt

was made to apply these calculations to X-ray spectra rather than monoenergetic beams of X rays, it was noted that by programming the required energy spectra into a computer along with the data which were presented, it is possible to extend the theory to X-ray spectra.

A review of some of the methods of calculating X-ray dose in soft tissue near or within bone is given by Spiers (1966). Comparisons of the results obtained by using these different methods are also presented. Spiers (1966) concluded that even though the results are not perfectly compatible, there is sufficient agreement with available experimental data to increase the acceptability of the theory.

Methods of Measuring Dose in Soft Tissue Near Bone

Attempts to measure the X-ray dose in soft tissue near bone began several years before the development of Spiers' (1949) theory. Stenstrom and Marvin (1946) used a combination of carbon-walled and bone-walled air ionization chambers to measure the dose in soft tissue within bone cavities of ten microns radius, this being the thickness of soft tissue which is the mass equivalent of the air that was used to fill the chamber. The ratio of the response of the bone chamber to that of the carbon chamber varied over the energy range from 30 keV to 200 keV in a manner closely following the

ratio of the dose calculated by Spiers (1949) in a tissue slab ten microns thick interposed between two bone slabs to the dose to similar tissue remote from bone. The maximum response ratio was 3.9 at an effective X-ray energy of 50 keV. This ratio is slightly less than the ratio calculated by Spiers (1949). Deviations from theory can be attributed to the non tissue-equivalence of both carbon and air. The electron stopping power of air is also different from soft tissue.

Spiers and Woodard (1953) made quantitative comparisons between the biochemical changes produced in bone by exposure to X rays of three differing qualities and the absorption of energy in bone which is expected from theoretical considerations. The alkaline phosphatase activity of living mouse bone was used as an index of radiation damage since this enzyme plays an essential role in bone metabolism and can be measured quantitatively. The average depressions of the phosphatase activity produced by 100 kVp and 185 kVp X rays were respectively 1.35 and 1.36 times greater than the average depression produced by the same exposure to 1000 kVp X rays. By considering the spatial distribution of phosphatase activity in and near bone, the average distance of the phosphatase producing cells from the bone was determined to be approximately 30 microns. With this distribution the doses from the 100 and 185 kVp radiations were calculated to be 1.2 to 1.3 times higher than

the dose from the 1000 kVp radiation. The Spiers method (1949) of dose calculation was used. This report which is based upon a biological indicator of radiation damage, rather than physical measurements, further supports the results previously discussed.

Fowler (1957) measured the dose from X rays in soft tissue adjacent to plane bone interfaces using a polyethylene-film dosimeter. Earlier work by Fowler and Farmer (1953, 1954, 1955, 1956) had confirmed that the electrical conductivity induced in polyethylene upon irradiation is related to the energy absorbed in it by a known power law. Since polyethylene is nearly soft tissue-equivalent, direct measurements of the dose in soft tissue were possible. By placing the thin film of polyethylene next to bone, which was simulated by Pyrex glass, the dose in soft tissue near bone was determined. The film was then placed next to soft tissue, simulated by polystyrene, to determine the equilibrium soft tissue dose. By taking the ratio of these two doses the increased dose in soft tissue near bone was determined for different quality X-ray beams. Since fairly thick graphite electrodes were deposited on the polyethylene to measure the induced current, it was not possible to get closer than 10-15 microns away from the bone interface. Therefore, the greatest increased dose effect near the interface could not be observed. Exact quantitative agreement with Spiers' calculations (1949) was not possible. However, there was general qualitative

agreement for effective X-ray energies greater than 50 keV. Fowler's (1957) results were high at low effective energies. This was attributed to the use of a spectrum of X-ray photons rather than a monoenergetic beam, as was assumed in Spiers' (1949) calculations. Those calculations made later by Spiers and Woodard (1953) which took into account the spectra of X rays are in closer agreement at the lower energies.

The first experimenters to perform precision measurements of the dose in soft tissue at plane bone interfaces and in parallel-slab bone cavities were Wingate, Gross, and Failla (1962). The measurements were made using a parallel-plate ionization chamber (extrapolation chamber) which was designed to measure X-ray dose in either bone or soft tissue. This was accomplished by employing electrodes of either bone-equivalent or soft tissue-equivalent plastics with soft tissue-equivalent or bone-equivalent gas contained between them. By measuring the ionization in this gas for different electrode spacings, the dose distributions across bone-soft tissue interfaces were determined with high precision. Measurements of the dose at distances approaching one micron from the bone interfaces could be made since the density of the gas was low due to reduced pressure, thus simulating very thin layers of either soft tissue or bone. The average dose to the soft tissue contained in very small parallel-slab bone cavities was also determined with

precision. Unlike the experiments mentioned previously, this experiment measured the absolute dose in soft tissue near bone. This meant that virtually all sources of experimental error had to be eliminated. Comparison with the calculations of Spiers (1949) was very good with most of the deviations arising from X-ray spectral considerations.

Aspin and Johns (1963) measured the X-ray dose in soft tissue contained in small cylindrical cavities within bone. A soft tissue-equivalent suspension of T4 Bacteriophage was employed as a biological dosimeter. By placing these cells in "bone-equivalent" Pyrex glass capillary tubes and determining their survival rates upon X-irradiation, the dose in the soft tissue cavity was determined. Measurements were made of this dose both as functions of X-ray energy and bone cavity radius. Comparisons of these experimental results with calculations which were also made in this same paper indicate agreement with theory is quantitative if statistical analyses of the measurements are made. It is to be noted that the actual X-ray spectra employed in the measurements were used in making the calculations.

Another measurement of the dose in soft tissue cavities within bone was performed recently by Ellis (1966). Pyrex sintered glass filters filled with aqueous ferrous sulfate were used to simulate soft tissue cavities within trabecular bone, aqueous ferrous sulfate

resembling soft tissue and Pyrex glass resembling bone. The dose in the ferrous sulfate solution upon X-irradiation was determined by a standard colorimetry method of chemical analysis. The ratio of the dose to the ferrous sulfate within the pores of the glass filters to the dose in free ferrous sulfate solution was taken to be a measure of the increased dose to soft tissue within cylindrical bone cavities. Comparison with theory was complicated by the fact that the pores of the glass filters used were of uneven dimensions. Thus, average pore sizes had to be used. The results were compatible with both Spiers' (1949) and Charlton and Cormack's (1962) theories at effective X-ray energies above 50 keV, but were not in agreement at lower energies.

MATERIALS AND METHODS

In this investigation the increased dose in soft tissue adjacent to plane bone interfaces and in parallel-slab bone cavities during X-irradiation was measured. The materials employed consisted of a polyethylene-film dosimeter and an apparatus designed to simulate the soft tissue contained at these two locations. The properties of the X rays employed in the measurements will be briefly described since consistent specification of X-ray quality was necessary to correlate the different results obtained.

The Polyethylene-Film Dosimeter

Theory

It has been verified by Fowler and Farmer (1953, 1954, 1955, 1956), Fowler (1957, 1959), Armistead, Pennock, and Mead (1949), and Mayburg and Lawrence (1952) that the electrical conductivity of dielectric materials is increased upon exposure to X-irradiation. These investigators were primarily concerned with examining dielectric materials suitable for use in radiological instruments. It was found in the course of their work that the magnitudes of the conductivity changes were directly related to the amount of energy deposited in the dielectric material by the radiation. This relation

can be stated as:

$$k \propto D^m$$

where k is the electrical conductivity, D is the absorbed dose rate, and m is a constant whose value depends on the dielectric material being used.

Fowler (1957), realizing that these conductivity changes were a good indication of the amount of energy absorbed in the dielectric material, designed a device capable of measuring dose rate by applying a potential difference across a thin film of polyethylene and measuring the current flow through it during X-irradiation. This electrical current, which is directly proportional to the conductivity change, was then related to the dose rate by the proportionality shown previously.

Fowler (1957) chose polyethylene as the dielectric material because the currents measured during irradiation were considerably larger than the associated dark currents (thermal noise, cosmic rays, etc.). Polyethylene is approximately soft tissue-equivalent with an effective atomic number of 5.44 (the effective atomic number of soft tissue is usually taken to be 7.40). This indicates that the photoelectric X-ray attenuation coefficients of polyethylene are similar but not identical to those of soft tissue. The stopping powers of polyethylene and soft tissue are the same (Whyte, 1959), however, and these materials are of approximately the same mass

density. These factors indicated to Fowler (1957) that the dose in polyethylene can be used as a measure of the dose in soft tissue.

Fowler (1957) presented a detailed description of the polyethylene-film dosimeter and its operating characteristics. Fowler (1959) also gave a thorough theoretical treatment of the radiation induced conductivity phenomena.

Construction

The polyethylene-film dosimeter employed in these measurements consisted of a film of polyethylene 12.7 microns thick, upon which were deposited two graphite electrodes (Figure 1). The polyethylene film was obtained from British Visqueen Ltd.[2] and was specified to be free of impurities and additives that are sometimes contained in polyethylene. The thickness of the polyethylene film was determined by making weight-density measurements and computing the thicknesses of several films of known area. The uncertainty in the determined 12.7 micron thickness was ±1.0 micron for five different samples. Care was exercised in the handling of the film since stretching it considerably altered its dimensions.

The polyethylene film was rinsed in ethanol to remove surface

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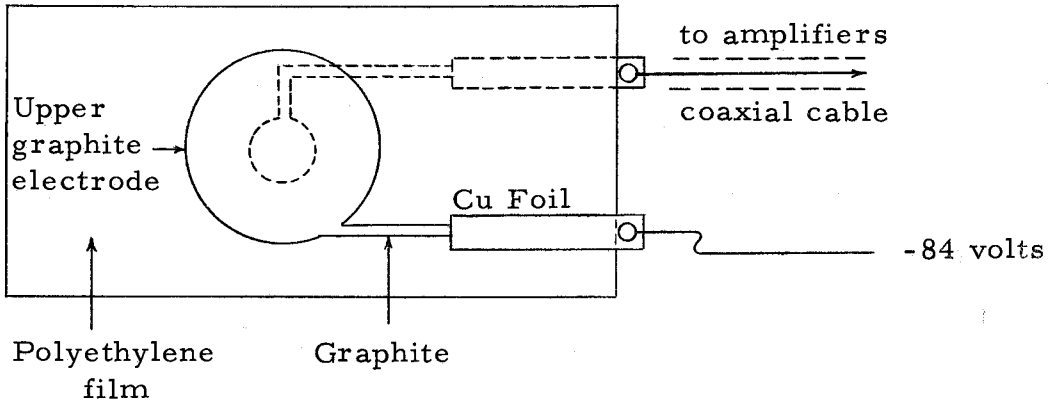
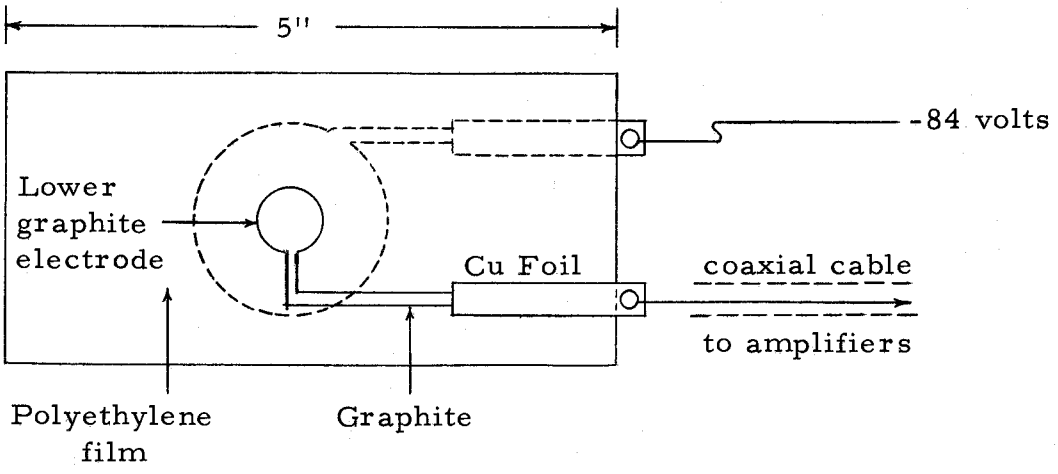
[2] British Visqueen Limited. Six Hills Way, Stevange, Hertfordshire, England.



Top view



Bottom view

Figure 1. Diagram of polyethylene-film dosimeter showing film, electrodes, and lead-in strips.

contamination which could affect its electrical properties. Immediately after cleansing, the graphite electrodes were painted onto the film, as shown in Figure 1, with a clean sable-hair artists brush. Acheson Colloids Dag Dispersion 154[3] which is an aromatic suspension of graphite was used. Since this suspension was quite viscous, it was diluted with ethanol to enable electrodes of the necessary dimensions to be deposited on the film. This thinner mixture flowed freely and deposited coherent layers of graphite, ensuring good electrical continuity and eliminating regions of high resistance. The upper voltage electrode was 4.5 centimeters in diameter and the lower collecting electrode 1.0 centimeter in diameter. The thickness of these electrodes was determined by weight-density measurements. A reproducible electrode thickness of 0.5 ± 0.2 microns was normally obtained.

The sensitive area of the film where the change in the conductivity was measured is that area between the two electrodes which is defined by the smaller collecting electrode. The X-ray beam was collimated with lead slits and projected onto this sensitive area. Since the measurements made were all relative, rather than absolute, it was concluded that there was little need for having a well-defined sensitive area such as is usually provided by guard-ring arrangements.

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[3] Acheson Colloids Company. Port Huron, Michigan.

However, a somewhat inefficient guard-ring was provided for this purpose and will be explained later.

The electrodes deposited on the polyethylene film were connected into the measuring circuit, Figure 2, by means of graphite lead-in strips which were painted directly onto the film. These graphite strips made contact with strips of copper foil which were secured to the film with epoxy cement and plastic tape. Solder connections at the ends of the copper strips made contact with the voltage side and collecting side of the measuring circuit.

The measuring circuit was a current loop containing a voltage source and a Victoreen Hi-Meg resistor. The voltage was supplied by two 42 volt mercury cells chosen for long-time stability. The current through the polyethylene film flowed through the Hi-Meg resistor and the resulting potential drop was fed into a preamplifier containing a vibrating-reed capacitor where it was changed to an alternating signal. This signal was further amplified by the main amplifier and read out on a meter calibrated in amperes. The preamplifier, Hi-Meg resistors, and main amplifier were incorporated in a Victoreen Model 475 vibrating-reed electrometer which was capable of measuring currents as low as 10^{-15} with high precision.

Since currents of less than 10^{-12} amperes were measured, provision had to be made to shield all of the collecting circuit components against electrostatic noise pickup and also against air

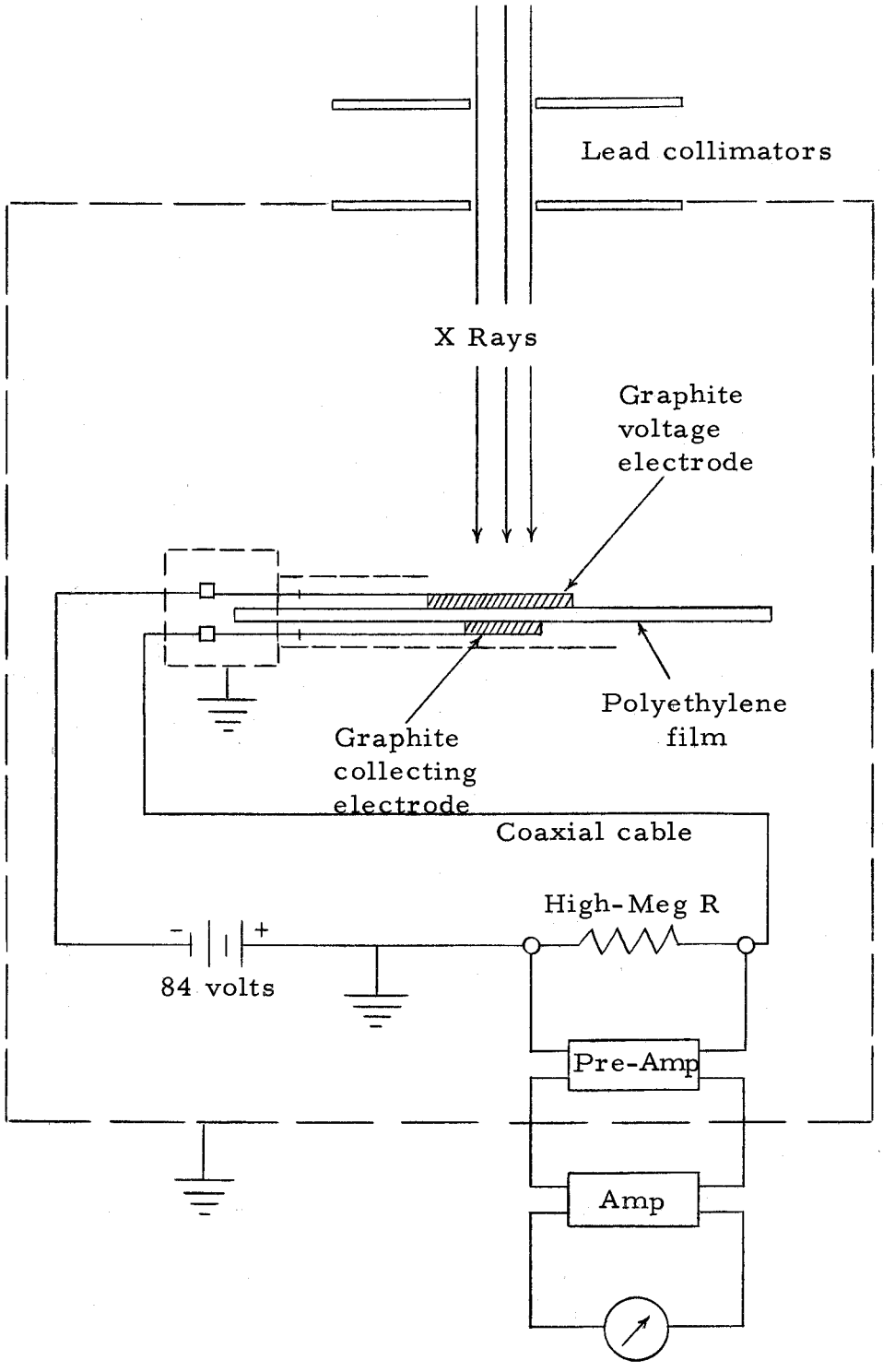

Figure 2. Diagram of circuit used to measure the radiation induced conductivity of polyethylene. Guard circuits are denoted by dotted lines.

ionization produced by the X rays. The circuits depicting these guard arrangements are shown in Figure 2. Short, grounded coaxial leads were employed to minimize air ionization and noise pickup as much as possible. The use of a low voltage (84 volts) reduced the strength of collecting fields. Grounded sheets of either copper or graphite-coated polyethylene provided guards against a collecting field existing between the collecting electrode and voltage electrode lead-in strips. These are shown in Figure 3. The lower sheet of copper can also be considered as a field shaping electrode (guard-ring) which serves to define the sensitive collecting area of the polyethylene film, as previously mentioned.

Figure 3 shows the apparatus used to hold the polyethylene film in place. This holder was made of Plexiglass to provide additional insulation.

The electrometer was connected to the preamplifier with 20 feet of coaxial cable provided with the electrometer. This enabled remote readings to be made without personnel exposure to radiation. All of the other components were housed in a lead-shielded, grounded Faraday cage made of lead, aluminum foil, and plywood (Figure 4).

Operating Characteristics

Before conductivity measurements were made using the polyethylene-film dosimeter, calculations were carried out to determine

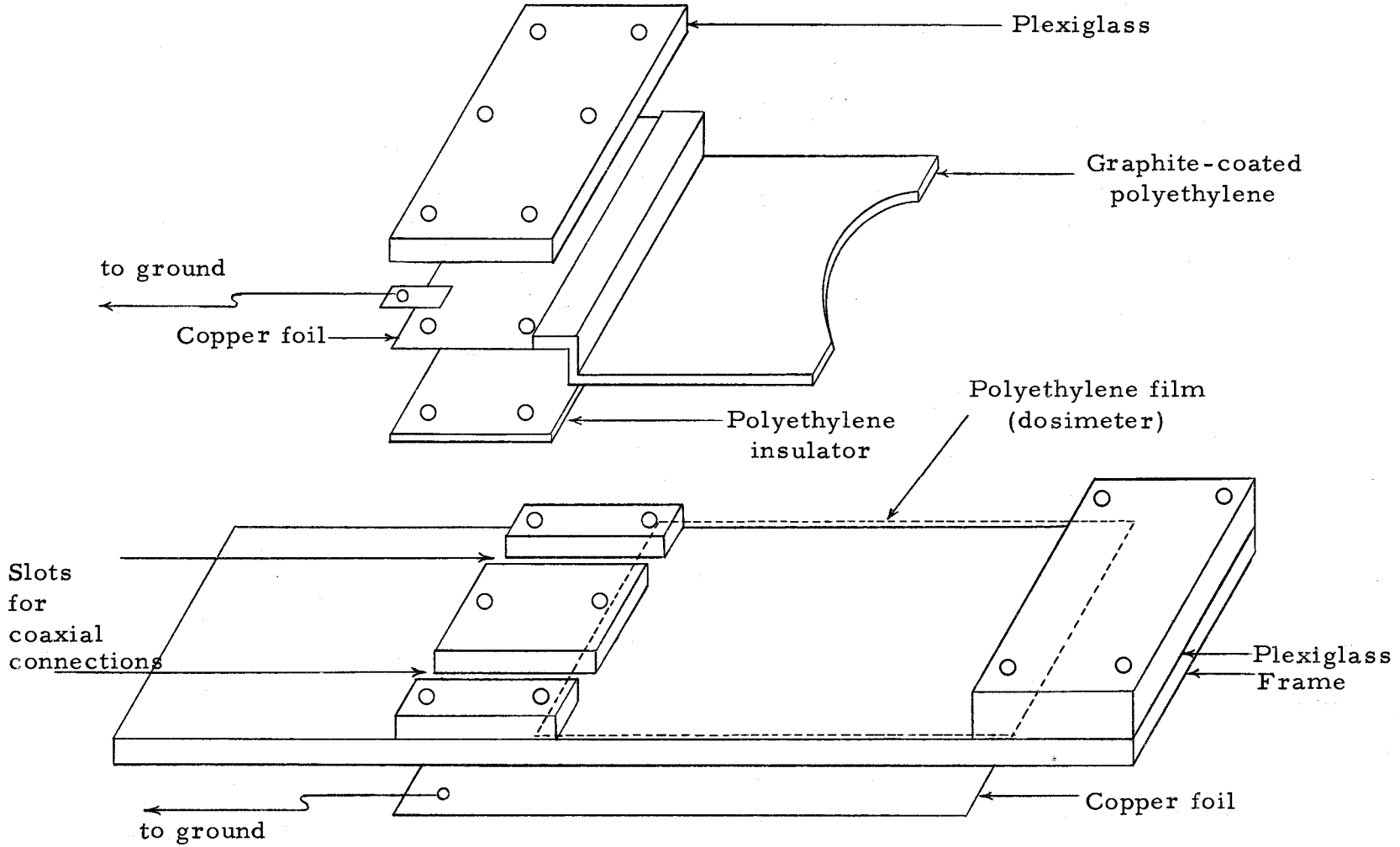

Figure 3. Apparatus for holding polyethylene-film dosimeter. Guard electrodes are shown.

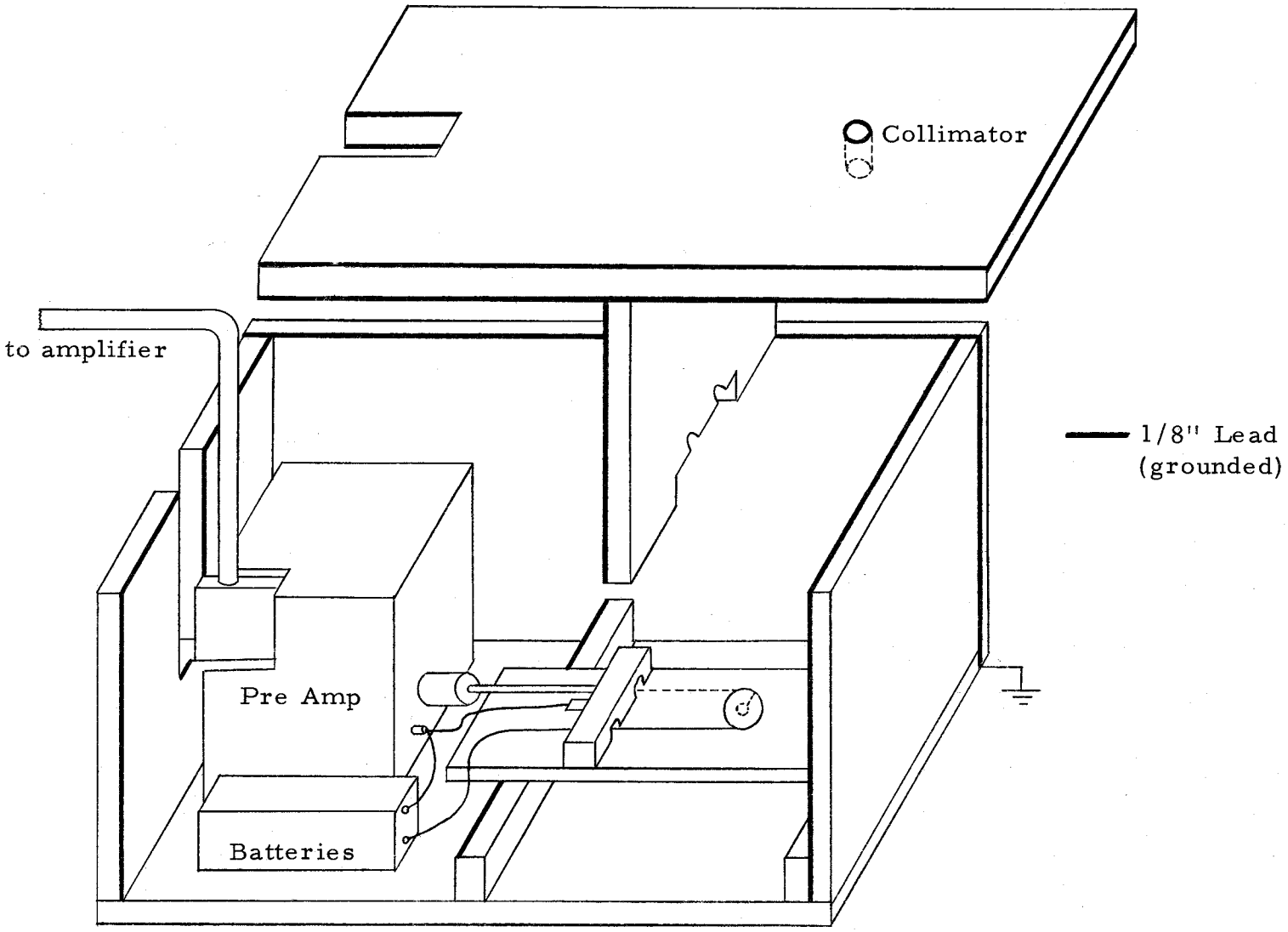

Figure 4. Complete apparatus shown housed in lead shielded Faraday cage.

the magnitude of the leakage current due to the natural conductivity of polyethylene. This was calculated with the formula:

$$I = \frac{E}{R} = \frac{E}{pL/A}$$

where E is the applied potential, L is the film thickness, A is the sensitive area between the graphite electrodes, p is the resistivity of polyethylene, and I is the electrical current. The following values were used:

$E = 84$ volts

$L = 1.4 \times 10^{-3}$ cm

$A = 0.8$ cm^2

$p = 10^{17}$ ohm-cm (manufacturers' specification)

Using these values the leakage current was computed to be 5×10^{-13} amperes. This value was in satisfactory agreement with the measured leakage current which was observed to be between 10^{-12} and 10^{-13} amperes. However, the measured value is dependent upon temperature, film tension, humidity, and equipment construction. Since the experimental and theoretical current values were in reasonable agreement, it was assumed that the conductivity of the film was being measured rather than other sources of leakage.

The polyethylene-film dosimeter was irradiated with 75 kVp X rays to measure the magnitudes of the induced currents[4] and the

[4] The current flow due to the radiation induced conductivity of the polyethylene will be hereinafter referred to as the induced current.

response time of the system. The results are shown in Figure 5, and indicate that the induced current reached an equilibrium value approximately eight to ten minutes after the X rays were turned on. When the X rays were turned off the current decreased rapidly at first, and then decayed away exponentially as predicted by Fowler's theory (1959). It was concluded from these results that to make the required measurements, the X rays should be left on for at least ten minutes for consistent measurements. The time lapse between successive exposures had little effect on the equilibrium induced current values. It was observed that increased exposure rates accelerated the approach to equilibrium, but the current decay after exposure was not as rapid. The results shown in Figure 5 correspond to an exposure rate of ten Roentgens per minute.

The results shown in Figure 5 indicated that the induced current at ten Roentgens per minute was significantly greater than the leakage current for satisfactory measurements to be made. At higher exposure rates the induced currents were increased by factors of 10 to 100 over the leakage current.

To determine the relationship between the dose rate in the polyethylene and the induced current, the induced current was measured as a function of exposure rate. The results of these experiments are shown in Figure 6. The exposure rate is directly proportional to the absorbed dose rate and can be used as a measure

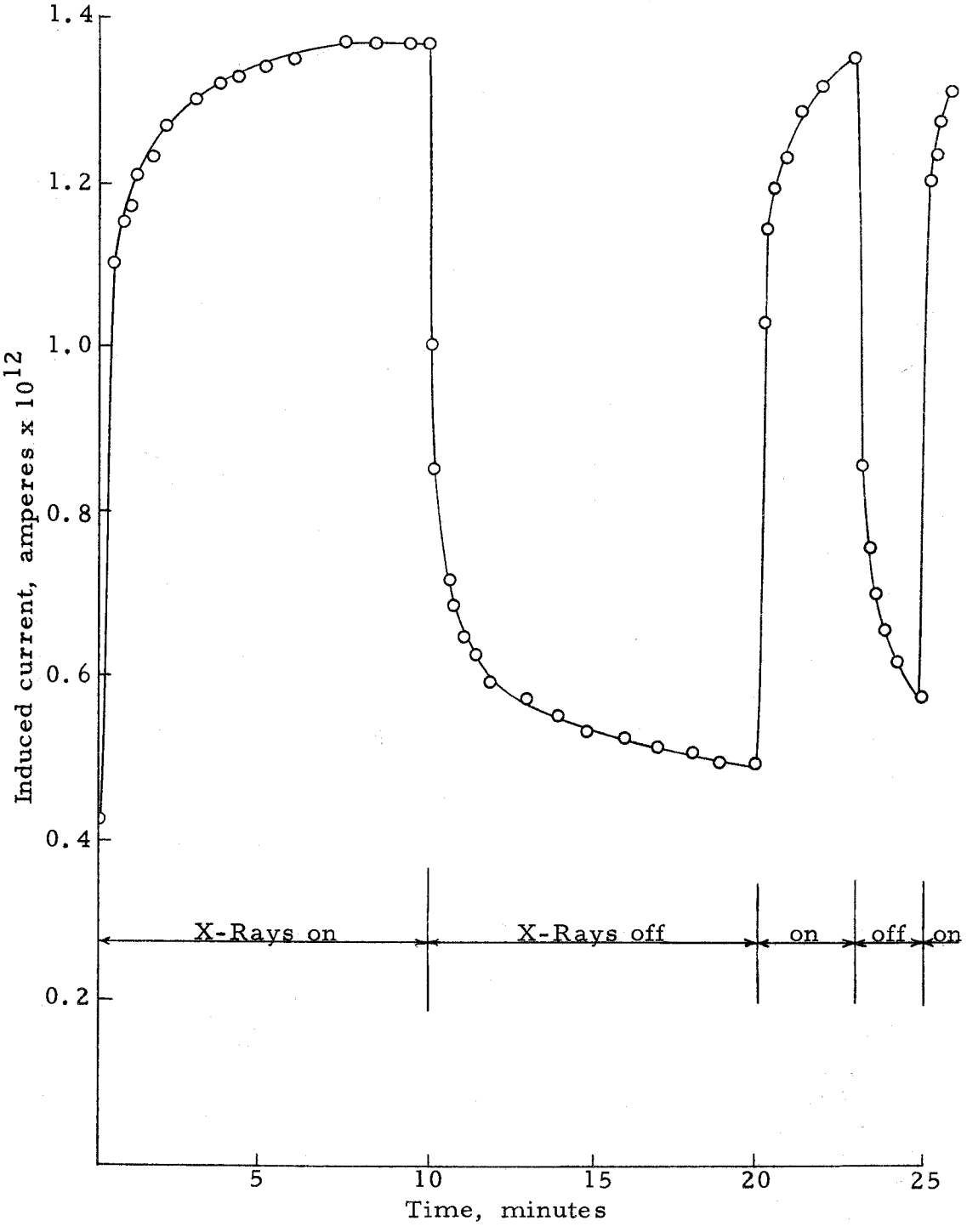

Figure 5. Relationship between induced current in polyethylene and time during which X rays are off and on. Exposure rate of 10 Roentgen/min.

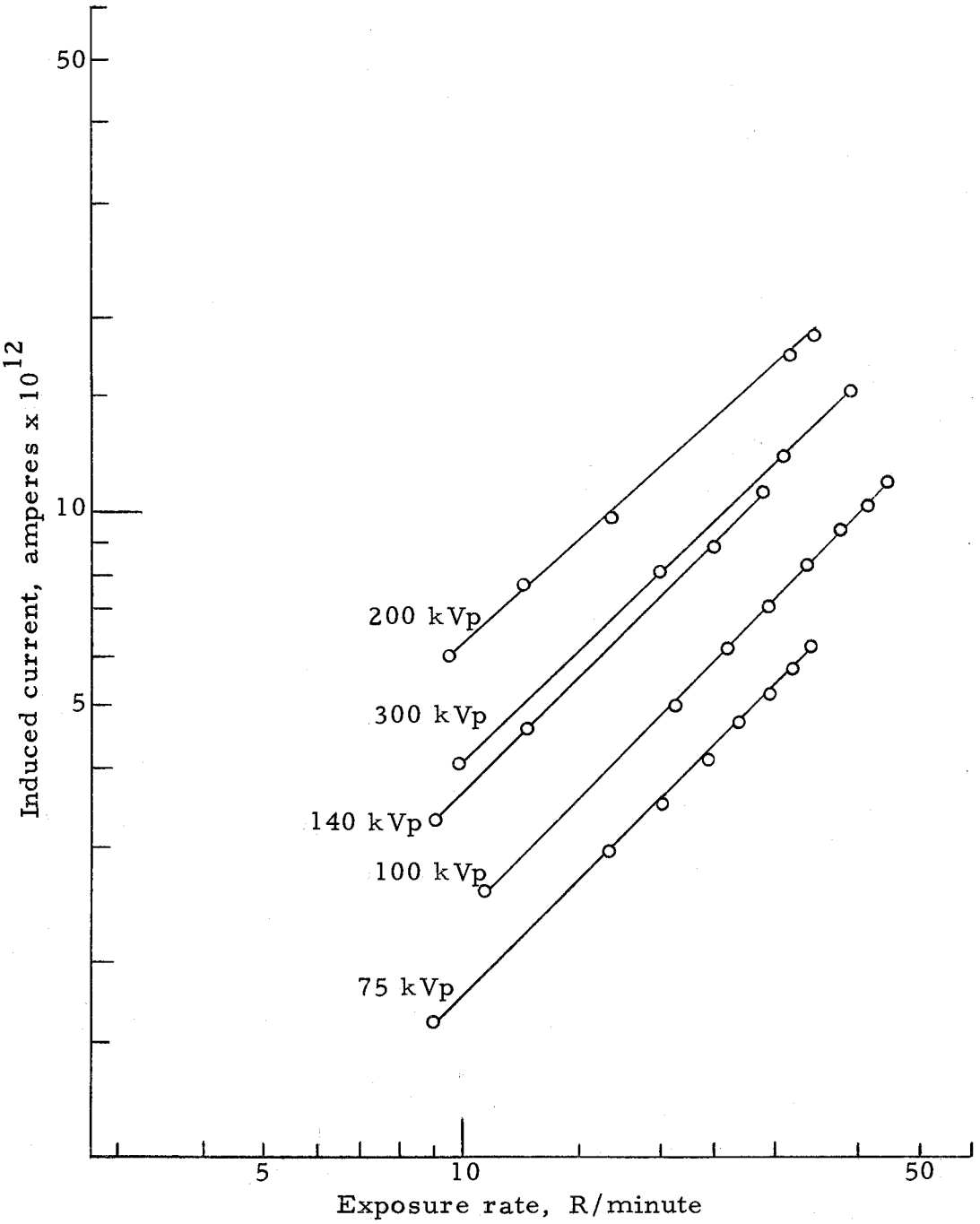

Figure 6. Induced current in polyethylene as a function of X-ray exposure and X-ray energy.

of the absorbed dose rate in the polyethylene. Since the log-log plot of exposure rate versus induced current was linear, it was verified that the results obeyed the relationship:

$$i \propto k \propto D^m$$

where i is the induced current, k is the induced conductivity, and D is the absorbed dose rate. The slope of the line, m, was found to vary slightly with X-ray energy. The values of m that were determined are listed below for the respective X-ray energies:

kVp	m
75	1.00
100	1.00
140	0.97
200	0.93
300	0.96

Fowler and Farmer (1956) found the value of m for polyethylene to be 0.80, which is lower than these values. This discrepancy may be attributable to a variety of different experimental methods and materials than those used by Fowler and Farmer (1956). The use of lower exposure rates in this study than those used by Fowler and Farmer (1956) may also affect the measurements.

Since the value of m was not constant over the X-ray energy range employed, it was necessary to compute doses from induced currents by using the value of m corresponding to the X-ray energy used.

Errors introduced in the measurements by air ionization were

determined by shielding the sensitive area of the polyethylene-film dosimeter with a disk of lead 0.5 inches thick. The current measured upon X-irradiation was increased eight to ten percent over the leakage current. The current measured with the lead disk removed was 1300-1400 percent greater than the leakage current. It was concluded that air ionization was negligible, and the guard electrodes were functioning in accordance with the design criteria.

A neon lamp was placed next to the apparatus to determine if electrostatic noise was being picked up in the measuring circuit. Measurements of the leakage current with the lamp both off and on were identical. This proved noise pickup from external sources to be negligible and showed that the grounded Faraday cage was functioning properly.

The results of these preliminary check-out procedures indicated that the polyethylene-film dosimeter was operating in an acceptable manner for the determination of dose rates in soft tissue. The only corrections necessary were those related to X-ray energy.

Dose Measurements

To measure the dose in soft tissue near bone with the polyethylene-film dosimeter it was necessary to employ an apparatus capable of simulating both soft tissue adjacent to plane bone interfaces and soft tissue contained in parallel-slab bone cavities. The

methods of measuring the soft tissue dose in these two locations differed somewhat and will, therefore, be presented separately.

Dose Measurements in Soft Tissue Adjacent to Plane Bone Interfaces

The increased dose to soft tissue adjacent to a plane, semi-infinite slab of bone was measured both as a function of depth into the soft tissue and X-ray quality. The apparatus employed in these measurements is shown in Figure 7.

A disk of bone-equivalent plastic, 0.50 centimeters thick and 4.5 centimeters in diameter, was used to simulate a slab of bone. This plastic was supplied by Dr. Francis Shonka of St. Procopius College, Lisle, Illinois and is specified by Shonka, Rose, and Failla (1958) to have the same chemical composition as human bone. Plastic material having the same composition as soft tissue was also supplied by Dr. Shonka but was employed in the cavity measurements only.

Disks of polyethylene were used throughout to simulate soft tissue since the dosimeter employed was made of polyethylene. Consistent use of this material avoided disturbance of the secondary electron fluxes. The small amount of graphite deposited on both sides of the polyethylene-film dosimeter was assumed to have a negligible effect on these electron fluxes since its atomic number

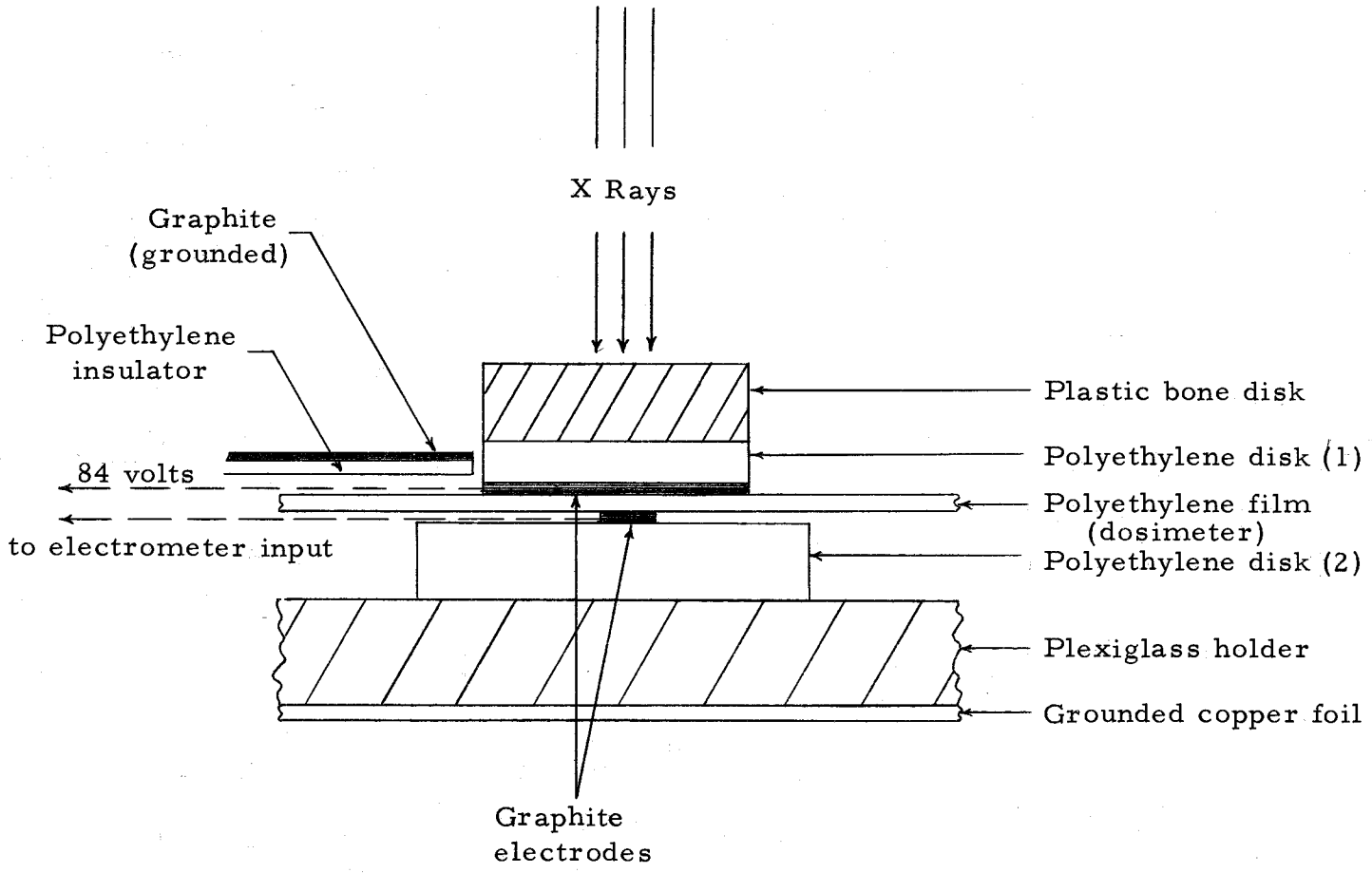

Figure 7. Apparatus used to simulate plane bone-soft tissue interfaces. Guard electrodes are also shown.

of 6.0 is not considerably different than the effective atomic number of polyethylene.

The distance from the bone interface to the polyethylene dosimeter was determined by the thickness of the disk of polyethylene (1), shown in Figure 7, between the dosimeter and the bone disk. Varying the thickness of this disk varied the distance of the dosimeter from the bone interface. To obtain absorbed doses in soft tissue directly adjacent to bone, the disk of polyethylene (1) was completely removed.

The depth into the soft tissue at which the measured dose was considered to occur was calculated using the following relationship:

$$d = x + y + z/2 \quad \text{microns}$$

where x is the variable thickness of polyethylene disk (1), y is the upper graphite electrode thickness of 0.5 microns, z is the thickness of the polyethylene dosimeter determined to be 12.7 microns, and d is the distance from the bone interface to the midpoint of the polyethylene-film dosimeter. Figure 8 shows this relationship schematically.

Dose measurements were made by assembling the apparatus as shown in Figure 7 with the desired thickness of polyethylene disk (1). Polyethylene disk (2), shown in Figure 7, was made 200 microns thick to simulate an infinite layer of soft tissue behind the dosimeter. The mechanical currents resulting from disturbance

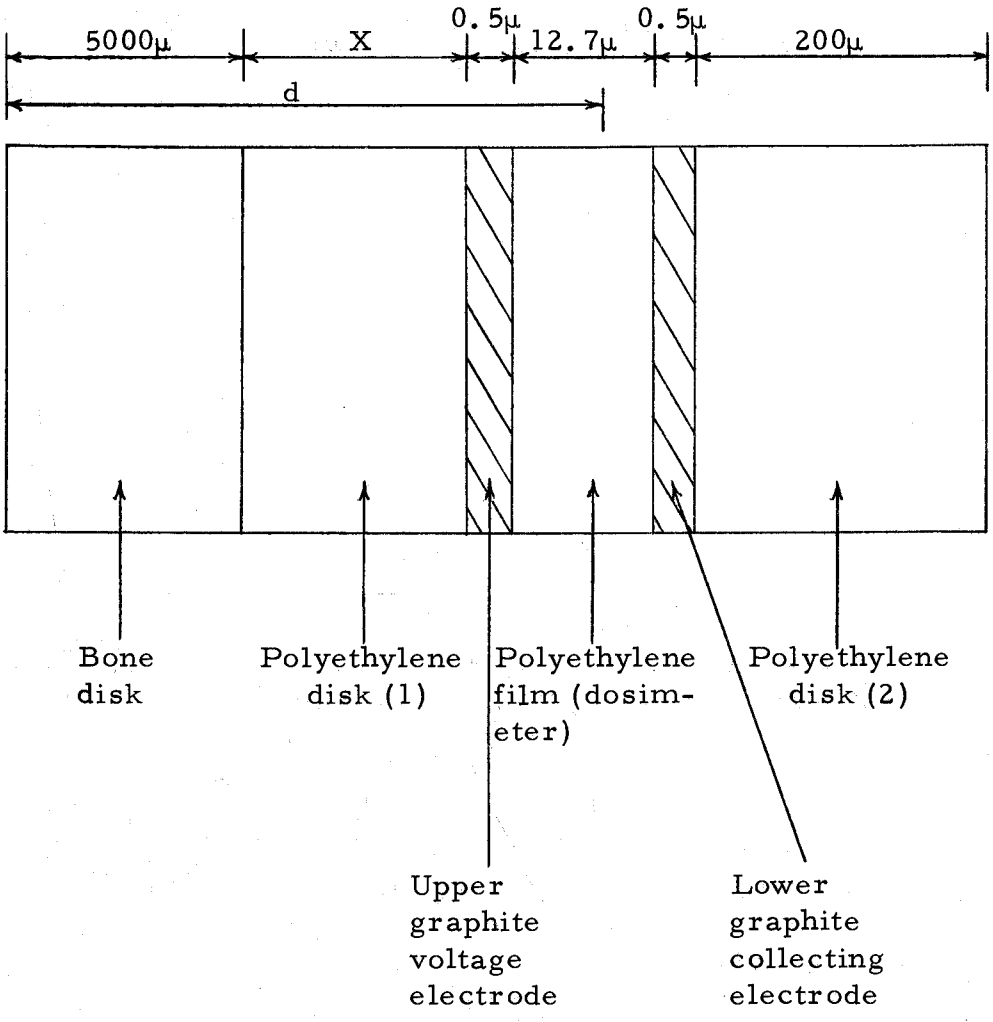

Figure 8. Diagram showing dimensions of the components of the simulated plane bone-soft tissue interface.

of the polyethylene dosimeter were allowed to subside for a period of approximately ten minutes. X-irradiation was begun and after ten minutes of continuous exposure the current due to the induced conductivity in the polyethylene was measured.

The relative energy absorption in soft tissue near plane bone interfaces was measured rather than the dose in rads to this soft tissue. This eliminated the necessity of calibrating the polyethylene dosimeter to read dose directly. The relative energy absorption was determined by first making polyethylene disk (1) of equilibrium thickness; i.e., of thickness greater than the maximum range of photoelectrons originating in bone. A thickness of 84 microns was sufficient for this purpose based upon the ranges of the maximum energy photoelectrons quoted in Lea (1962). The current, i_o, induced in the polyethylene film with this polyethylene disk in place was then taken to be related to the equilibrium dose rate in tissue, D_o. By then varying the thickness of polyethylene disk (1) from 0 to 84 microns, the induced currents, i, in the polyethylene film at distances closer to the bone were determined. By taking the ratio of i to i_o the increased induced currents in the polyethylene at these distances were determined. The values of i/i_o were converted to increased doses by the relationship:

$$\frac{D}{D_o} = \left(\frac{i}{i_o}\right)^{\frac{1}{m}}$$

where m is the constant determined earlier and D/D_o is the increased dose in soft tissue near bone. D/D_o is equal to unity for soft tissue remote from bone.

Dose Measurements in Soft Tissue Within Parallel-Slab Bone Cavities

The increased energy absorption in a layer of soft tissue interposed between two parallel, semi-infinite slabs of bone was measured as a function of X-ray quality. A soft tissue layer 12.7 microns in width was employed to simulate the Haversian canals found in bone which range in diameter from 10 to 100 microns. A 12.7 micron cavity was chosen since the dose to small cavities is greatest and is, therefore, of the most concern.

The apparatus employed in these measurements was similar to the apparatus used for measuring the dose in soft tissue adjacent to plane bone interfaces. A few of the components were changed, to better simulate a bone cavity. The apparatus is shown in Figure 9. A hole, 4.5 centimeters in diameter, was drilled in the Plexiglass dosimeter holder to enable a bone-equivalent plastic disk and soft tissue-equivalent plastic disk of Shonka's material to be inserted beneath the polyethylene film. A soft tissue disk and bone disk were also placed above the film. The bone disks were both 4.5 centimeters in diameter and 0.50 centimeters thick, while the soft tissue

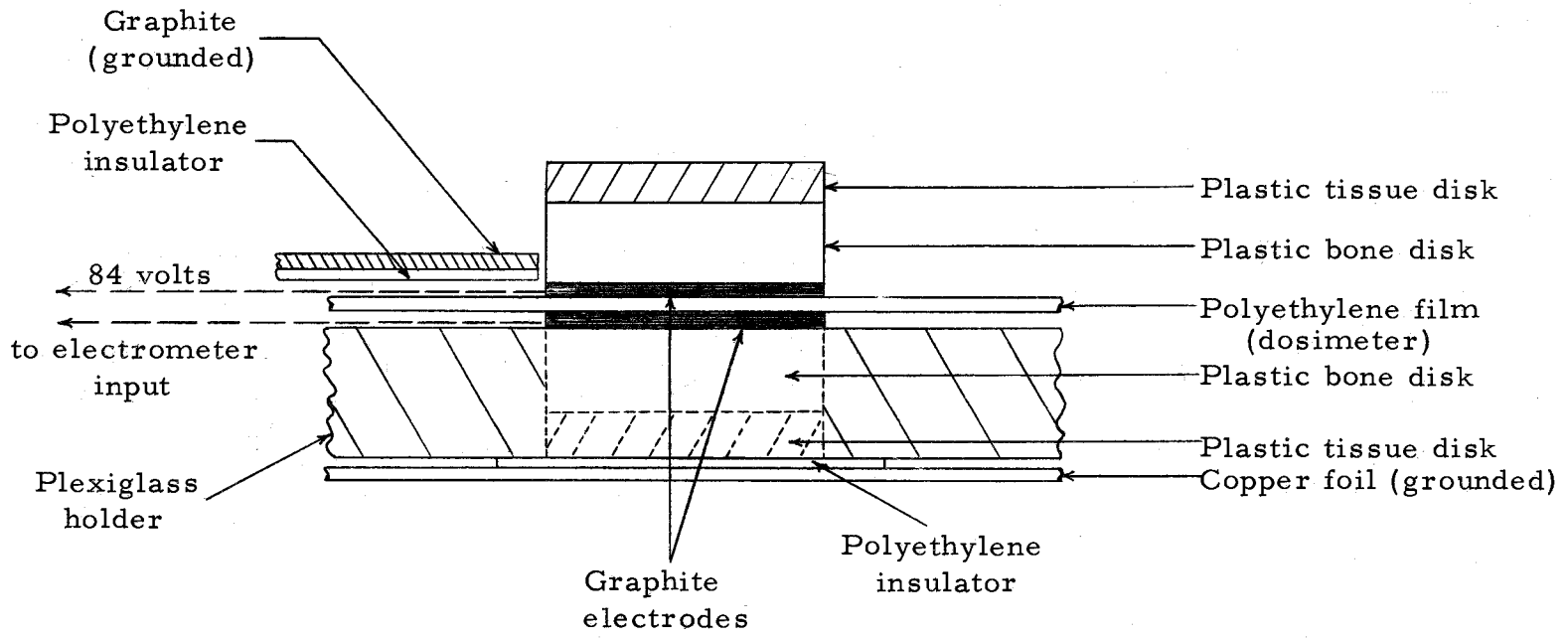

Figure 9. Apparatus used to simulate a parallel-slab soft tissue cavity in bone. Guard electrodes are also shown.

disks were both 4.5 centimeters in diameter and 0.20 centimeters thick.

Since the bone and soft tissue disks contained graphite and were conducting, it was necessary to enlarge the lower graphite collecting electrode to 4.5 centimeters in diameter, thus making it the same size as the upper graphite voltage electrode. The X-ray beam was not recollimated and the area of the polyethylene film upon which the X rays impinged remained the same as before. Air ionization might have become a problem had the X-ray beam been enlarged to the size of the graphite electrodes.

By arranging the soft tissue disks and bone disks as shown in Figure 9, a small soft tissue element interposed between two semi-infinite slabs of bone was simulated. The width of this soft tissue element was equal to the width of the polyethylene film and graphite electrodes. By measuring the current, i, induced in the polyethylene during irradiation, a measure of the dose rate, D, in a soft tissue cavity within bone was obtained.

By then rearranging the bone disks and soft tissue disks so that the soft tissue disks were adjacent to the upper and lower graphite electrodes, a true soft tissue orientation was simulated. This was possible due to the fact that the soft tissue disks were of equilibrium thickness. The current induced upon X-irradiation, i_o, was taken to be a measure of the equilibrium soft tissue dose, D_o.

By taking the ratio of i to i_o and converting to D/D_o, as previously done for the plane bone interface measurements, the increased dose in the small soft tissue cavity was obtained.

There were certain problems associated with switching the bone and soft tissue disks to obtain the respective D and D_o values. Physical re-orientation can cause changes in geometry and differing film tensions. Since the objective was to measure D/D_o as a function of X-ray energy for a constant soft tissue cavity thickness, this problem was solved by first making measurements of D with the bone disks adjacent to the polyethylene film for the range of X-ray energies studied. The cavity geometry was then changed so the tissue disks were adjacent to the polyethylene film, and measurements of D_o were made for the same range of X-ray energies. Theory predicts that the D/D_o value for high energy X-rays, approximately 300 kVp, is equal to unity. The D/D_o values at lower energies were normalized by dividing by the 300 kVp D/D_o value. Errors were reduced by making only one change in the geometry and remaining errors were cancelled by the normalizing process.

X-Ray Production, Measurement, and Energy Specification

To provide useful data for clinical dosimetry, two X-ray machines used for X-ray therapy were employed in this investigation. The machines used were a General Electric Maxitron 300, which is

a 300 kVp beryllium window unit operated at 100-300 kVp, and a General Electric Maximar 100, which is a 100 kVp beryllium window unit operated at 50-100 kVp.

For purposes of comparison with theory, it was necessary to specify an effective energy which roughly described each of the different X-ray spectra used in the measurements. This was calculated using the relationship:

$$\mu_a(E) = \frac{0.693}{HVL}$$

where HVL (half-value-layer) is the thickness of absorber which reduces the intensity of the X-ray beam to one half, and μ_a is the linear X-ray absorption coefficient which depends on the X-ray energy. The effective energy, E, was then taken from tables by Grodstein (1957) which present μ_a as a function of X-ray energy. Though this method of calculation is strictly applicable only to monoenergetic photons, it is the only method available for specifying the effective energy of an X-ray spectrum (Johns, 1961).

The half-value-layers of the X-ray spectra obtained with the 300 kVp X-ray unit were dialed on the control panel and the kVp and beam filtrations were set automatically. Reference to data for the 100 kVp X-ray unit gave the half-value-layers as a function of kVp and beam filtration which were adjusted manually.

A Victoreen Radocon, Model 575, employing a medium energy

chamber (30-400 keV) was used to measure the exposure rate response of the polyethylene dosimeter at the different X-ray energies. No attempt was made to correct the low energy measurements since the measurements were relative.

To ensure constant exposure conditions while using the 300 kVp X-ray unit, an ionization chamber permanently mounted in the X-ray beam was used for monitoring purposes. No such arrangement was available on the 100 kVp X-ray unit so the exposure was checked periodically for constancy using the Radocon.

EXPERIMENTAL RESULTS

Results of Dose Measurements in Soft Tissue Near Plane Bone Interfaces

The increased dose, D/D_o, in soft tissue adjacent to a plane bone interface was measured as a function of depth in the soft tissue and X-ray quality. The experimental results are presented in tabular form, Tables 1 and 2, and graphical form, Figures 10, 11, and 12. The graphical results are most useful for comparisons with theory since trends in deviations can be readily seen and interpreted. The tables show the experimental uncertainties in the measured values. The measured values shown are the mean of four individual determinations and the uncertainties are average deviations from these means.

The results obtained with the 100 kVp X-ray unit and the 300 kVp X-ray unit are presented separately since the spectral distributions obtained with these two units differed. Measurements were made using X-ray qualities which are commonly employed in X-ray therapy.

The results presented in Figures 10, 11, and 12, and Tables 1 and 2 are descriptive of the increased dose effect in soft tissue near bone. In clinical dosimetry, however, it is desirable to know the increased dose in finite layers of soft tissue rather than the dose

Table 1. Increased X-ray doses in soft tissue adjacent to plane, semi-infinite slabs of bone--300 kVp X-ray unit.

X-Ray Quality kVp / HVL	100 0.25 mm Al (14.4 keV)	100 0.5 mm Al (17.7 keV)	100 1.0 mm Al (22.2 keV)	140 3.0 mm Al (32.6 keV)	140 0.5 mm Cu (62 keV)	200 1.0 mm Cu (83 keV)	300 4.0 mm Cu (177 keV)
Distance from Interface to Tissue Element	$\frac{D}{D_o}$	$\frac{D}{D_o}$	$\frac{D}{D_o}$	$\frac{D}{D_o}$	$\frac{D}{D_o}$	$\frac{D}{D_o}$	$\frac{D}{D_o}$
6.9	2.09±.06	2.26±.03	2.27±.06	2.12±.04	1.74±.01	1.46±.05	1.00±.01
19.6	1.29±.01	------	1.39±.01	1.45±.01	1.39±.01	1.25±.01	1.00±.01
26.0	1.09±.02	------	1.11±.03	1.22±.02	1.26±.04	1.19±.02	1.00±.01
32.3	1.00±.01	------	1.02±.02	1.16±.01	1.16±.01	1.12±.01	1.00±.02
45.0	1.00±.01	------	1.00±.01	1.00±.01	1.04±.02	1.02±.01	1.00±.01
57.7	1.00±.01	1.00±.01	1.00±.01	1.00±.01	1.01±.01	1.01±.01	1.00±.01
83.0	1.00±.01	1.00±.01	1.00±.01	1.00±.01	1.00±.01	1.00±.01	1.00±.01

Table 2. Increased X-ray doses in soft tissue adjacent to plane, semi-infinite slabs of bone--100 kVp X-ray unit.

X-ray quality	50 kVp 0.75 mm Al HVL (20.2 keV)	75 kVp 1.85 mm Al HVL (27.6 keV)	100 kVp 2.95 mm Al HVL (32.5 keV)
Distance from interface to tissue element	$\frac{D}{D_o}$	$\frac{D}{D_o}$	$\frac{D}{D_o}$
6.9	1.39 ± .05	2.05 ± .14	2.24 ± .05
19.6	1.01 ± .01	1.24 ± .07	1.37 ± .06
26.0	1.03 ± .02	1.16 ± .06	1.24 ± .03
32.3	0.95 ± .03	1.05 ± .02	1.16 ± .03
45.0	1.00 ± .01	0.99 ± .06	1.08 ± .03
57.7	-------	1.02 ± .04	1.05 ± .03
83.0	-------	0.96 ± .04	0.99 ± .02

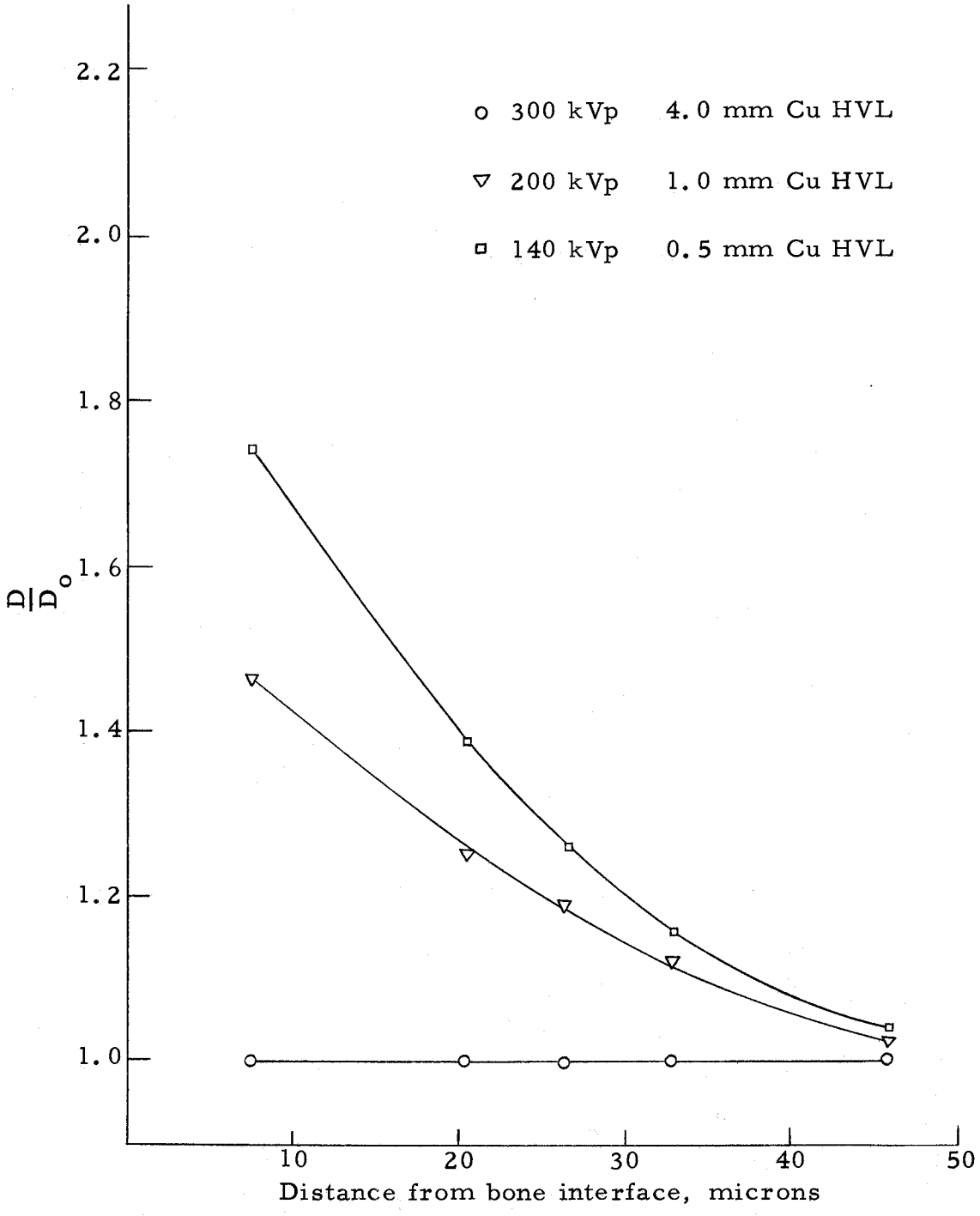

Figure 10. Increased X-ray doses in soft tissue adjacent to plane, semi-infinite slabs of bone--300 kVp X-ray unit.

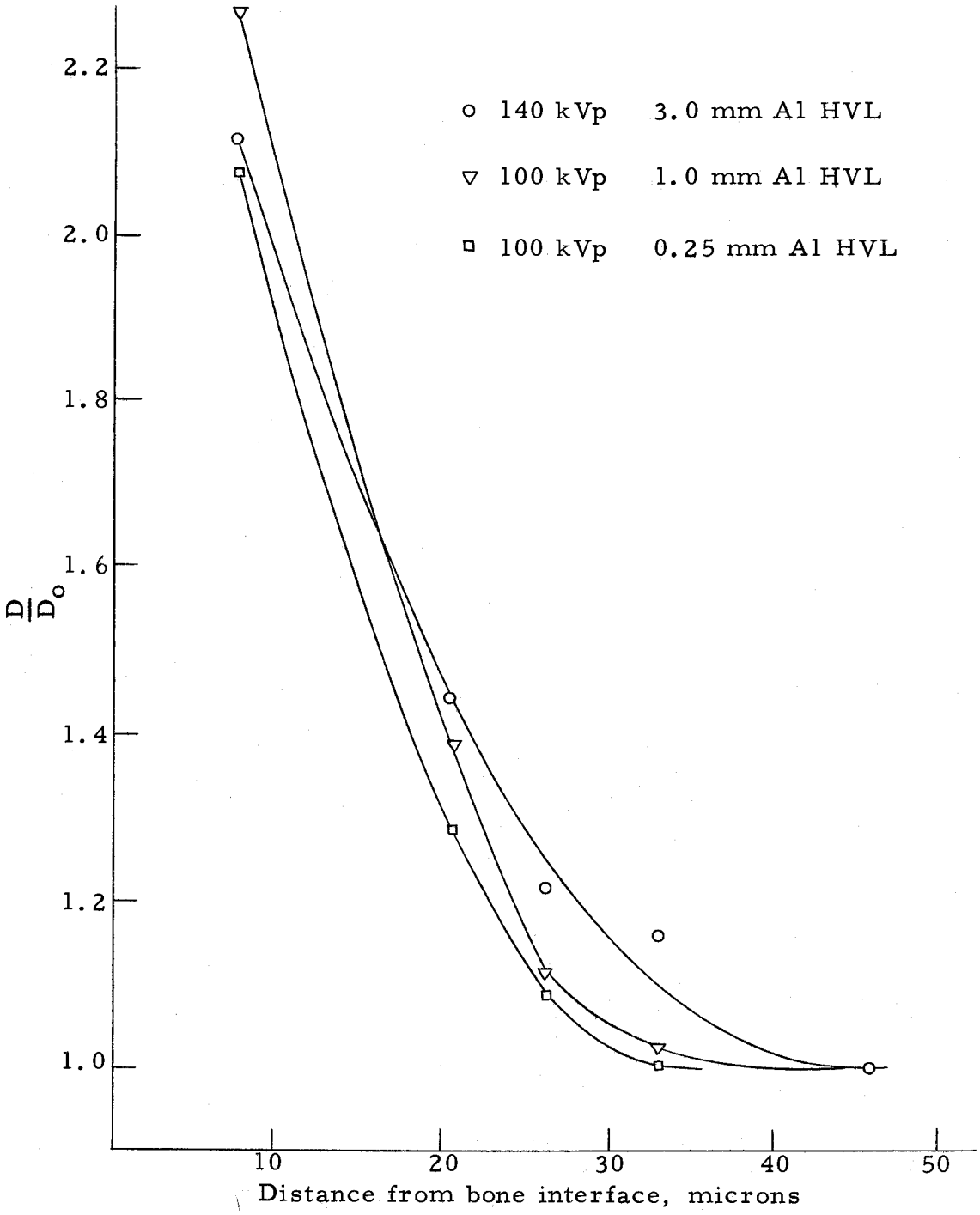

Figure 11. Increased X-ray doses in soft tissue adjacent to plane, semi-infinite slabs of bone--300 kVp X-ray unit.

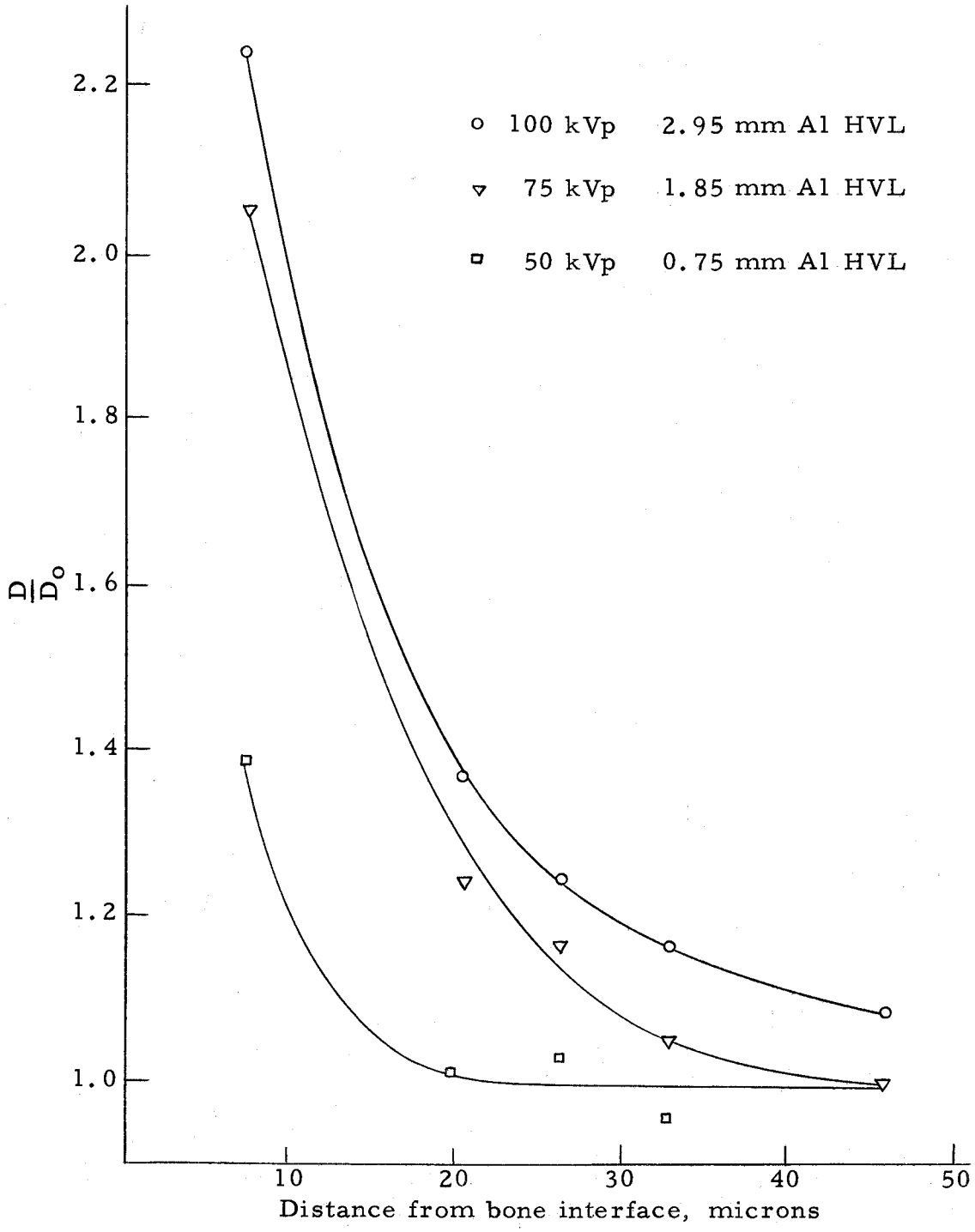

Figure 12. Increased X-ray doses in soft tissue adjacent to plane, semi-infinite slabs of bone--100 kVp X-ray unit.

at various depths in this soft tissue. The increased dose in finite soft tissue layers was easily determined since the polyethylene dosimeter integrated the dose over its finite thickness of 12.7 microns. By using polyethylene disks (1), shown in Figure 7, that were 0, 12.7, and 25.4 microns thick, and measuring the dose at each of these three positions, it was possible to average the measured doses and calculate the dose to soft tissue layers of 13.2, 25.9, and 38.6 micron thicknesses located adjacent to bone.[5] The results of these measurements and calculations are shown in Figure 13 and Tables 3 and 4. The results obtained with the 300 kVp X-ray unit and 100 kVp X-ray unit are presented separately.

To calculate the dose in the soft tissue of interest, the dimensions of the soft tissue and its position relative to bone must be determined. Knowing the X-ray quality, the increased dose factor, D/D_o, is taken from Table 3 or Table 4. This value is then multiplied by the equilibrium dose, D_o, to the soft tissue to obtain the corrected dose, D. If the thickness of the soft tissue layer is not found in either Table 3 or 4, or if it is not located directly adjacent

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[5] More precisely, these doses apply to 12.7, 25.4, and 38.1 micron soft tissue layers located 0.5 microns away from bone. For simplicity of presentation, the 0.5 micron graphite electrode thickness was added to each layer of soft tissue in which the dose was actually measured. Errors introduced by doing this are negligible compared to experimental uncertainties since 0.5 microns is a small percentage of the actual soft tissue layer thickness.

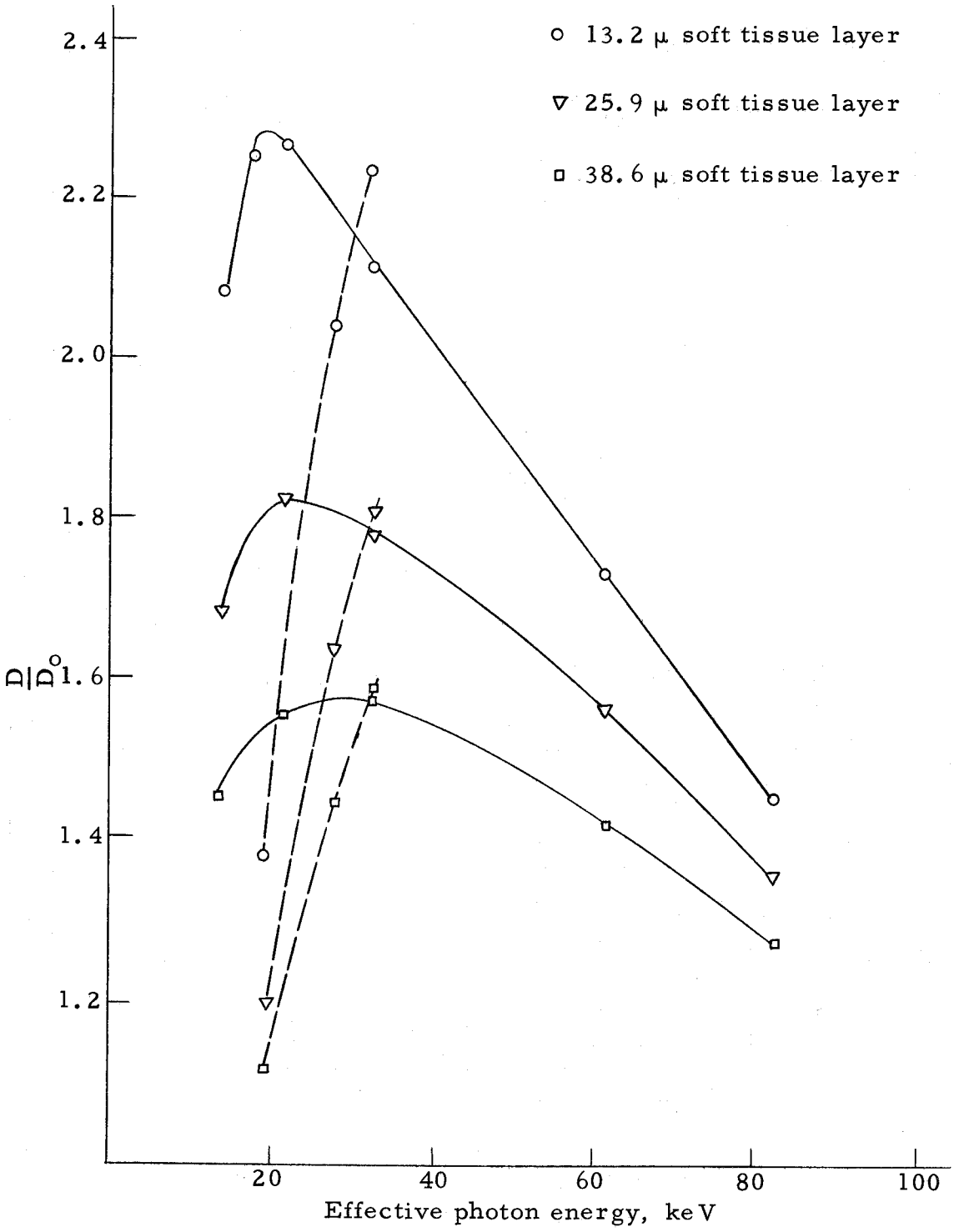

Figure 13. Increased X-ray doses in soft tissue layers of finite dimensions located adjacent to plane, semi-infinite slabs of bone shown as a function of effective photon energy--100 kVp X-ray unit, dotted curves; and 300 kVp X-ray unit, solid curves.

Table 3. Increased X-ray doses in soft tissue layers of finite dimensions located adjacent to plane, semi-infinite slabs of bone--300 kVp X-ray unit.

X-ray quality		Effective photon energy (keV)	D/D_o		
kVp	HVL		13.2 μ soft tissue layer adjacent to bone	25.9 μ soft tissue layer adjacent to bone	38.6 μ soft tissue layer adjacent to bone
100	0.25 mm Al	14.4	2.09 ± .06	1.69 ± .05	1.46 ± .05
100	0.5 mm Al	17.7	2.26 ± .03	-------	-------
100	1.0 mm Al	22.2	2.27 ± .06	1.83 ± .05	1.56 ± .05
140	3.0 mm Al	32.6	2.12 ± .04	1.78 ± .05	1.58 ± .05
140	0.5 mm Al	62	1.74 ± .01	1.57 ± .01	1.43 ± .02
200	1.0 mm Al	83	1.46 ± .05	1.36 ± .05	1.28 ± .05
300	4.0 mm Al	177	1.00 ± .01	1.00 ± .01	1.00 ± .01

Table 4. Increased X-ray doses in soft tissue layers of finite dimensions located adjacent to plane, semi-infinite slabs of bone--100 kVp X-ray unit.

X-ray quality	Effective photon energy (keV)	D/D_o 13.2 μ soft tissue layer adjacent to bone	D/D_o 25.9 μ soft tissue layer adjacent to bone	D/D_o 38.6 μ soft tissue layer adjacent to bone
50 kVp 0.75 mm Al HVL	20.2	1.39 ± 0.05	1.20 ± 0.05	1.12 ± 0.05
75 kVp 1.85 mm Al HVL	27.6	2.05 ± 0.14	1.64 ± 0.15	1.45 ± 0.13
100 kVp 2.95 mm Al HVL	32.5	2.24 ± 0.05	1.81 ± 0.09	1.59 ± 0.09

to bone, the increased dose factor may be found by graphical integration of the appropriate area under the curves of Figures 10, 11, or 12.

Results of Dose Measurements in Soft Tissue Contained Within Parallel-Slab Bone Cavities

Due to experimental difficulties which will be described later, it was not possible to measure the dose in soft tissue contained in parallel-slab bone cavities. The results of these measurements shown in Table V indicate a marked discrepancy with theory. For example, these results show that there is a decreased dose in soft tissue cavities for certain of the lower X-ray energies. Even where there is an increased dose effect, it is smaller than predicted by theory. A much larger increased dose effect would be expected in soft tissue within bone cavities than in soft tissue adjacent to plane bone interfaces.

In Table V are shown the results of measurements made using the 300 kVp X-ray unit. Measurements made using the 100 kVp X-ray unit were equally incompatible and for this reason are not presented.

Table 5. Increased X-ray dose in a 12.7 micron soft tissue layer interposed between two parallel, semi-infinite slabs of bone--300 kVp X-ray unit.

X-ray quality	Effective X-ray energy	D_b/D_t[6]		D/D_o[7]	
		Run 1	Run 2	Run 1	Run 2
100 kVp .25 mm Al HVL	14.4 keV	1.43	1.86	0.86	0.81
100 kVp 1.0 mm Al HVL	22.2 keV	1.20	1.93	0.72	0.84
140 kVp 3.0 mm Al HVL	32.6 keV	1.52	2.19	0.92	0.95
140 kVp .50 mm Cu HVL	62.0 keV	1.70	2.52	1.02	1.10
200 kVp 1.0 mm Cu HVL	82.7 keV	1.70	2.41	1.02	1.05
200 kVp 1.5 mm Cu HVL	99.3 keV	1.79	2.46	1.08	1.07
300 kVp 2.0 mm Cu HVL	113 kev	1.66	2.30	1.00	1.00

[6] D_b represents the dose measured with bone disks adjacent to the polyethylene film. D_t represents the dose measured with tissue disks adjacent to the polyethylene film.

[7] D/D_o is obtained by normalizing the D_b/D_t values such that D_b/D_t for the 300 kVp--2.0 mm Cu HVL measurement is unity.

COMPARISON OF EXPERIMENTAL RESULTS WITH THEORY

To confirm the validity of the doses which were measured in this experiment, comparison was made with the doses expected from theoretical considerations. Though various investigators have used different methods to calculate these doses, the calculations are essentially the same. Howarth's calculations (1965) were adapted for this investigation. This investigator used the least number of assumptions and the predicted doses should be more exact. The main disadvantage of the Howarth theory (1965) is that the calculations are feasible only if monoenergetic X-rays are assumed. To take the spectral distributions into account a computer is required, and an elaborate analysis program must be used.

Comparisons between the experimental results and Howarth theory (1965) will be made only for the plane interface situation. Incompatible results were obtained in the parallel-slab cavity measurements, and comparisons with the theory were not possible.

Though Howarth's theory (1965) is quite complex, the use of simplified equations and accurate tables which were provided simplified the calculations for the increased dose in soft tissue adjacent to plane bone interfaces. By determining the maximum energy, T, of the photoelectrons produced by the monoenergetic X-ray beam of interest, the effective range of the electrons, R_o, could be

calculated using the formula:

$$R_o = AT^{1.75}$$

where A is a constant given by Howarth (1965). Dividing the depth into the soft tissue at which the increased dose was to be calculated by R_o gave a number determining a geometrical factor, tabulated by Howarth (1965), for computing the increased dose at that depth. The increased dose at both the interface, $D/D_o(0)$, and at different depths in the soft tissue, $D/D_o(x)$, were then calculated using the equations:

$$\frac{D}{D_o}(x) = 1 + \hat{k}_T G_T$$

$$\frac{D}{D_o}(0) = 1 + \hat{k}_T G_T(0) + 0.5 k_a$$

where G_T is the geometrical factor for the photoelectrons, $\hat{k}_T$ is a parameter associated with the photoelectric X-ray attenuation coefficient, and k_a is a parameter associated with the production of Auger electrons. The values of $\hat{k}_T$ and k_a were tabulated in Howarth (1965) as a function of X-ray energy. These calculations were carried out for monoenergetic X-ray beams of approximately the same energy as the effective energies of the X-ray spectra employed in this experiment.

Comparisons of the results of these calculations with several of the experimentally measured values are shown in Figures 14 and 15.

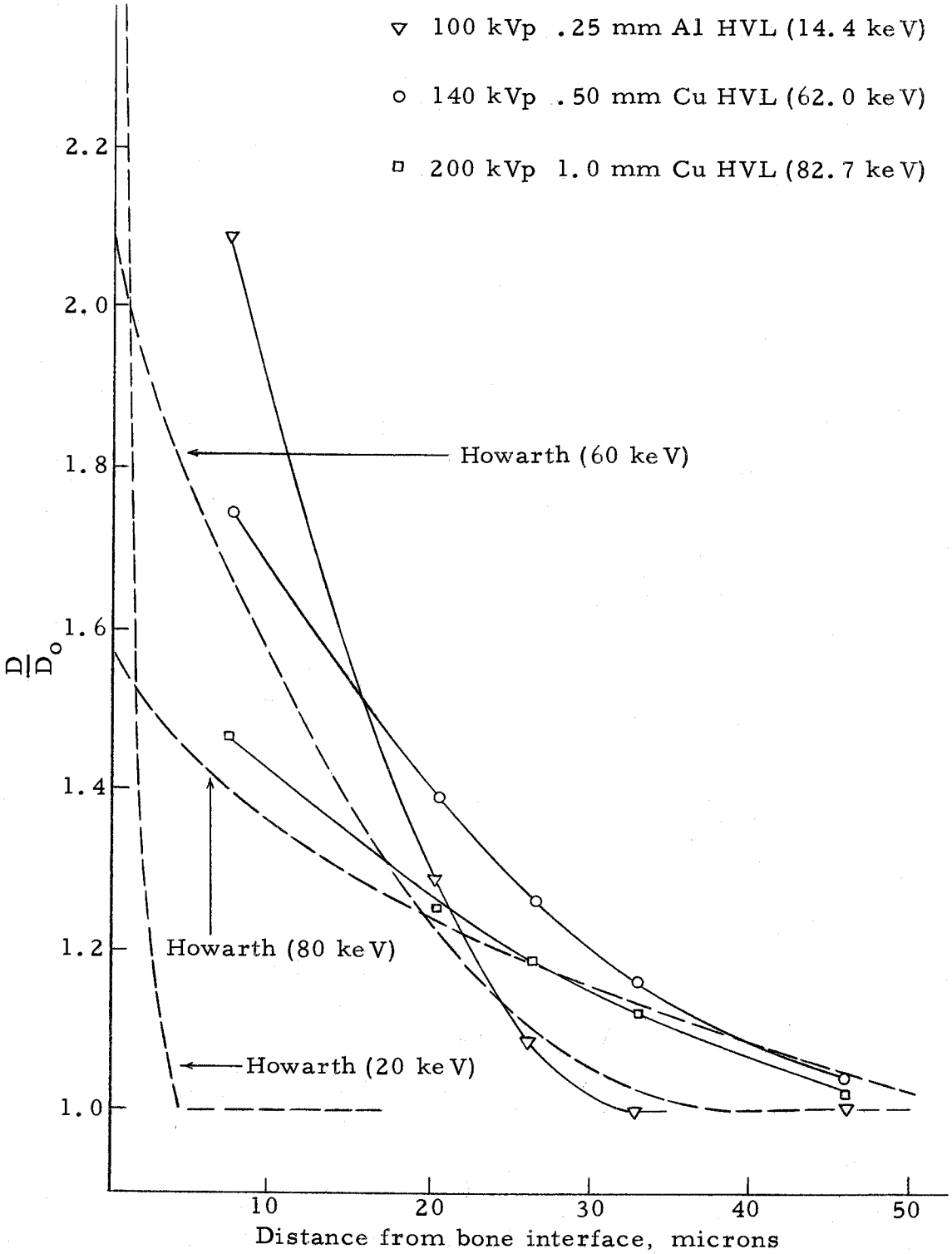

Figure 14. Increased X-ray doses in soft tissue adjacent to plane, semi-infinite slabs of bone. Comparison of results obtained with 300 kVp X-ray unit, solid curves, with Howarth (1965) data, dotted curves.

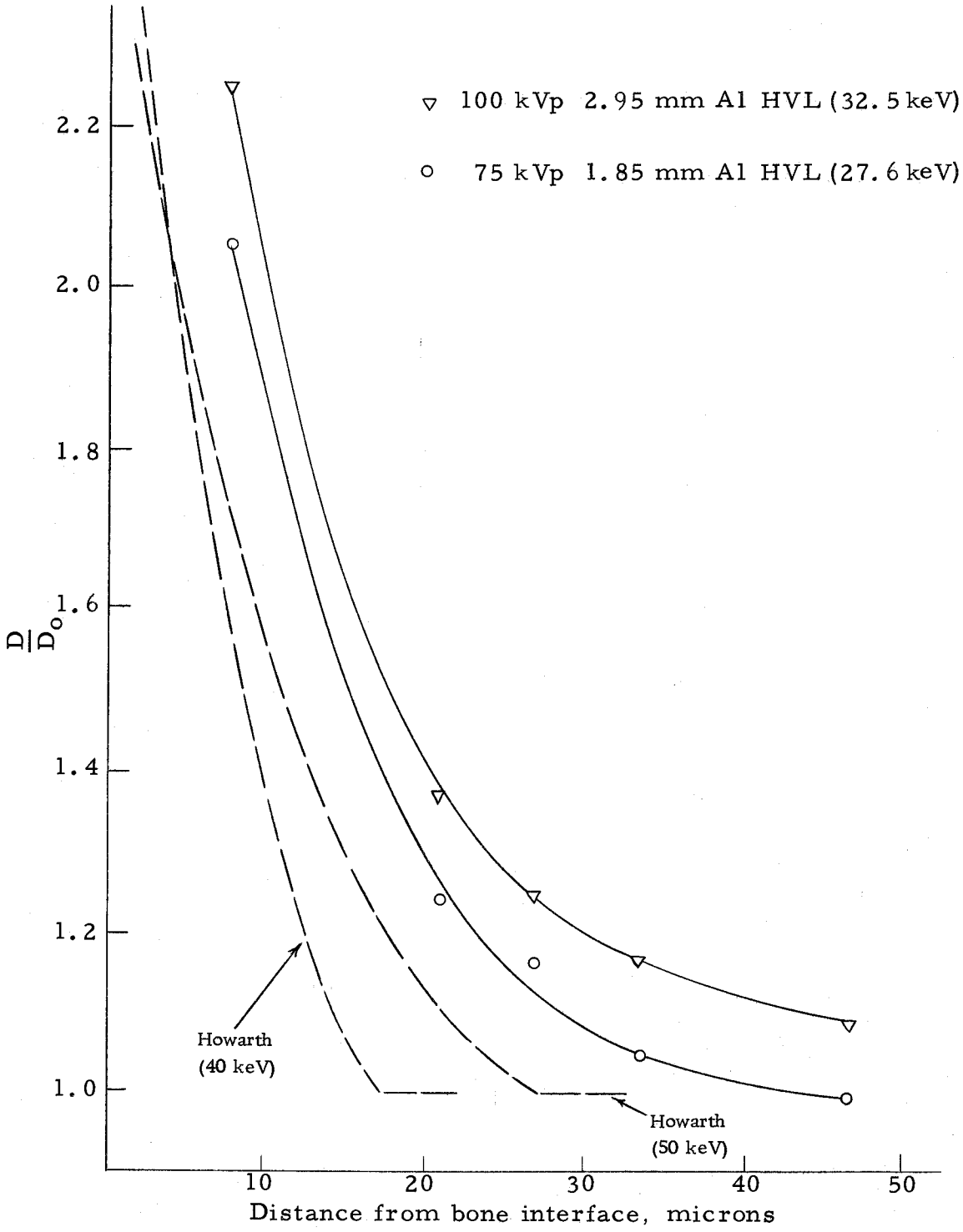

Figure 15. Increased X-ray doses in soft tissue adjacent to plane, semi-infinite slabs of bone. Comparison of results obtained with 100 kVp X-ray unit, solid curves, with Howarth (1965) data, dotted curves.

Only a few of the experimental results are used for comparative purposes to avoid complication of these diagrams. The results using the 300 kVp X-ray unit are shown in Figure 14 and the results using the 100 kVp X-ray unit are shown in Figure 15. The experimental results which are shown were selected in order that comparisons at both high and low X-ray energies could be made.

Comparisons are also made of the Howarth calculations (1965) with the results which were presented in Tables 3 and 4, and Figure 13. This comparison is shown in Figure 16 for the 13.2 micron tissue layer only. To obtain the desired theoretical values it was necessary to graphically integrate the results of Howarth (1965), such as are shown in Figures 14 and 15, for a range of different monoenergetic photon beams. In performing these integrations it was necessary to subtract the increased doses occurring in the area adjacent to the bone interface where the graphite electrode of the polyethylene dosimeter was located, since the dose was not measured at this location.

Figure 16 shows comparisons between theoretical and experimental doses for a 13.2 micron soft tissue layer only. Comparisons between experiment and theory for soft tissue layers of different dimensions showed the same general trends.

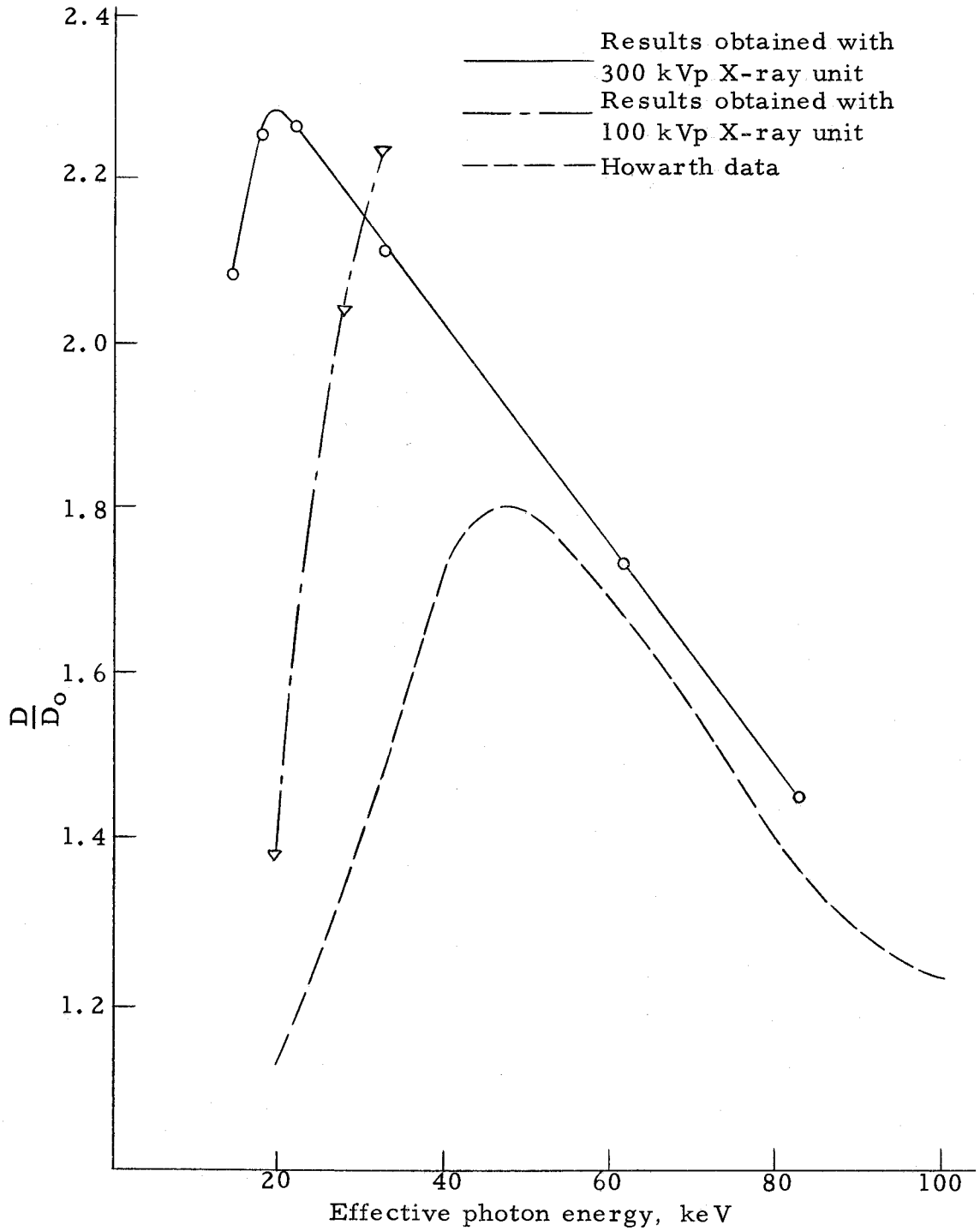

Figure 16. Increased X-ray dose in a 13.2 micron soft tissue layer located adjacent to a plane, semi-infinite slab of bone, as a function of effective X-ray energy. Comparison of results obtained with both the 100 kVp and 300 kVp X-ray units with theoretical data of Howarth (1965).

DISCUSSION OF RESULTS

Examination of the comparisons shown in the preceding section would lead to the conclusion that experiment and theory are not in good agreement. However, X-ray spectra of broad energy distributions were employed in this study and the results compared in Figures 14, 15, and 16 will not be in close agreement with theory which is based upon a monoenergetic radiation source. One of the main purposes of this experiment was to see what effect the spectra of X-ray energies does have on the dose in soft tissue near bone.

As can be seen from Figures 14 and 15, the experimentally determined dose distribution in soft tissue adjacent to a plane bone interface has the same general shape as the theoretical dose distribution. Relations between the experimental values are similar to relations between the theoretical values, but the experimental values differ considerably in magnitude from theory. Also, the results obtained with the 300 kVp X-ray unit differ from the results obtained with the 100 kVp X-ray unit. In order to validate the experimental results it is necessary to give plausible explanations for these differences.

The differences in magnitudes between the theoretical and experimental results is attributable to two major factors, the X-ray spectra and the nontissue-equivalence of polyethylene in the

photoelectric energy region.

In an X-ray spectrum there are photons of both higher and lower energies than the effective energy upon which the comparisons are based. The lower energy photons will considerably increase the dose at points in soft tissue close to the bone interface, while the higher energy photons will extend the increased dose effect further into the soft tissue layer. This is what was observed experimentally in Figures 14 and 15. Using a lightly filtered 100 kVp X-ray beam with an effective energy of 14 keV, both of the above effects are important, since most of the X rays are being attenuated by the photoelectric process. For a heavily filtered 200 kVp X-ray beam with an effective energy of 83 keV the photons with energy greater than 100 keV will have little effect on the increased dose since they are attenuated predominantly by the Compton process. There will be few low energy photons since heavy filtration is employed, and the dose in soft tissue very close to bone will not be appreciably increased above what would be expected for the 83 keV photons. In the latter case the effective energy is a good indication of the energy of the photons which are increasing the dose in the soft tissue, and comparison between experiment and theory should be good. Examination of the experimental results confirms this hypothesis.

It would have been desirable to have used heavier beam filtration to produce X-ray spectra of narrower energy distribution.

However, X-ray outputs were too low at high filtration for accurate measurements to be made.

Discrepancies between experiment and theory also arise from the nontissue-equivalence of polyethylene. The increased dose in polyethylene would be expected to be greater than that in soft tissue since there will be a greater difference between the equilibrium doses in bone and polyethylene than bone and soft tissue. This expected deviation is shown by the experimental results.

Differences in the results obtained with the 300 kVp X-ray unit and 100 kVp X-ray unit are attributed to the different spectral distributions produced by these units.

In Figure 16 it can be seen that measurements made at high effective energies on the 300 kVp X-ray unit and low effective energies on the 100 kVp X-ray unit are in reasonable agreement with theory. This seems to indicate that spectral considerations account for the major discrepancy between experiment and theory.

In the preceding paragraphs the terms discrepancy, difference, and deviation should not be construed as meaning experimental errors and uncertainties. Rather, they refer only to discrepancies between the experimental results and the theoretical results of Howarth (1965). The magnitudes of the experimental uncertainties were always sufficiently small to ensure differentiation between the different experimental results which were obtained.

Experimental errors were kept to a minimum in this experiment by making relative measurements only, rather than absolute measurements. So long as determinations of the dose distributions next to a bone interface were carried out under identical experimental and environmental conditions, most of the experimental errors were assured of cancelling. Four individual determinations of these dose distributions were made for each X-ray quality on different days to determine the uncertainties. The types of errors which will not cancel are those due to environment such as temperature or humidity changes during the individual runs. Errors in the determination of the constant m in the formula relating induced current and absorbed dose rate could introduce further uncertainty.

Since incompatible results were obtained for the dose measurements in soft tissue cavities within bone, it is necessary to define what was at fault if the previous results for the plane bone-soft tissue interface are to remain valid. The most plausible explanation for the results can be attributed to the nature of the polyethylene dosimeter used in the measurements. Unlike the dosimeter used in the plane interface measurements, the lower graphite collecting electrode was the same size as the upper graphite voltage electrode. Thus the area of the film defined by these electrodes was larger and the leakage currents measured without X-irradiation were approximately an order of magnitude larger. Since the X-ray beam was

not recollimated, its projected area on the sensitive area of the film was the same as for the plane interface measurements. As a result the current induced by X rays was not sufficiently greater than the leakage current to enable accurate measurements to be made. Air ionization was also greater with the larger collecting electrode.

Another problem encountered in the cavity measurements was attributable to the soft tissue disks and bone disks which were used. Though a wide variety of these disks were available in different thicknesses, the only ones which could be used were those with the dimensions mentioned earlier. The thinner disks were badly warped resulting in undesirable air gaps. The amount of both bone and soft tissue plastic interposed in the X-ray beam was, therefore, greater than desirable and considerable X-ray attenuation resulted. This reduced the induced currents, and introduced uncertainty in defining the effective energy of the radiation reaching the material surrounding the cavity. This problem did not present itself in the interface measurements since only a bone disk was interposed in the X-ray beam.

The above explanations for the poor cavity results are but hypotheses, and there is no direct experimental evidence that they are the sources of error. However, these two factors were the only experimental conditions which were different from the plane interface

measurements, and since reasonable results were obtained for the plane interface it is reasonable to assume the poor results to be due to these factors.

SUMMARY AND CONCLUSIONS

In this investigation a polyethylene dosimeter in conjunction with an apparatus for simulating soft tissue at locations near bone was used to measure the increased X-ray dose in soft tissue near bone. Both plane bone-soft tissue interfaces and parallel-slab bone-soft tissue cavities were considered, and the dose in the soft tissue at each of these locations measured.

Specifically, the increased dose in soft tissue adjacent to plane, semi-infinite slabs of bone was measured as a function of depth in the soft tissue and effective X-ray energy. The increased dose in soft tissue layers of finite dimensions located adjacent to bone was measured as a function of X-ray energy. It was found that agreement between experiment and the theory of Howarth (1965) was generally qualitative, with the magnitudes of the experimental doses being considerably higher than predicted by Howarth (1965). This difference was attributed to X-ray spectra and the nontissue-equivalence of polyethylene. From these results it can be concluded that the increased dose in soft tissue near plane bone interfaces is greater than predicted by theory when X-ray spectra are considered, especially for broad X-ray spectra specified by low effective energies.

An attempt was made to measure the increased dose in soft

tissue elements between two parallel, semi-infinite slabs of bone. However, these measurements could not be made due to experimental difficulties. The theoretical values for the dose in the Haversian canals, canalicules, or lacunae found within the inorganic matrix of bone requires further support due to the incomplete experimental results currently available.

The data presented in this investigation further confirm the theoretical calculations of the X-ray dose in soft tissue near bone. X-ray spectra as compared with monoenergetic radiation sources were shown to have a considerable effect on the derived doses, and should be considered when the theoretical calculation method is used.

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