Time-Specific Measurements of Energy Deposition from Radiation
Fields in Simulated Sub-Micron Tissue Volumes

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MASTERS THESIS

**TIME-SPECIFIC MEASUREMENTS OF ENERGY DEPOSITION FROM RADIATION
FIELDS IN SIMULATED SUB-MICRON TISSUE VOLUMES**

by

Michael A. Famiano

A Thesis Submitted in Partial Fulfillment
of the Requirements for the Degree of Master of Science
(Rackham School of Graduate Studies) at the University of Michigan.

Approved: April 29, 1996

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Acknowledgment

The fallacy of human memory renders it impossible to fully acknowledge those who have assisted in a work without creating invidious distinctions by failing to remember someone who has helped. However, a few individuals out of the many who have contributed to this paper are listed here.

Dr. David Hamby has patiently worked with me throughout the course of the research leading to this paper. His deep understanding of the subject and immense patience with inane questions has made the work progress quite efficiently. As well, his sense of humor has kept me out of trouble on more than one occasion. In addition, the tickets to sporting events have provided a great amount of comfort to a cluttered mind. Hopefully, this compliment will not sound condescending, as I consider Dr. Hamby a giant in his field.

Dan Hague and Jim Miller, two very good friends of mine, have helped out tremendously. Dan, by providing me with equipment to complete the project, helped me progress through the stages of this research quite expeditiously. Jim, by allowing me to use his workshop and providing helpful suggestions, as well as allowing me to keep him awake while others slept, helped make a great deal of this research possible.

Finally, I mention my gracious, loving wife, Nikki, to whom this work (as all my work) is dedicated. By surviving many nights without me, evenings of little conversation because of my preoccupation with other things, complaints, and tedious tasks to help me complete this paper, she has shown me a new level of patience, understanding, and compassion. To compile the help she has been to me through the written word would require volumes. Her beauty shall endure long after this paper - or its memory - ceases to exist.

The research was performed under appointment to the Applied Health Physics Fellowship Program administered by Oak Ridge Institute for Science and Education under contract number DE-AC05-76OR00033 between the U.S. Department of Energy and Oak Ridge Associated Universities.

"Where were you when I laid the earth's foundation?....
.... Can you bind the beautiful Pleiades?
Can you loose the cords of Orion?
Can you bring forth the constellations in their seasons
or lead out the Bear with its cubs?
Do you know the laws of the heavens?...."

- God to Job
(Job 38:4, 31-33)

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ABSTRACT: A tissue-equivalent spherical proportional counter is used with a modified amplifier system to measure specific energy deposited from a uniform radiation field for short periods of time ($\sim 1\mu\text{s}$ to seconds) in order to extrapolate to dose in sub-micron tissue volumes. The energy deposited during these time intervals is compared to biological repair processes occurring within the same intervals after the initial energy deposition. The signal is integrated over a variable collection time which is adjusted with a square-wave pulse. Charge from particle passages is collected on the anode during the period in which the integrator is triggered, and the signal decays quickly to zero after the integrator feedback switch resets; the process repeats for every "triggering" pulse. Measurements of energy deposited from x rays, ^{137}Cs gamma rays, and electrons from a $^{90}\text{Sr}/^{90}\text{Y}$ source for various time intervals are taken. Spectral characteristics as a function of charge collection time are observed and frequency plots of specific energy and collection time-interval are presented. In addition, a threshold energy flux is selected for each radiation type at which the formation of radicals (based on current measurements) in mammalian cells equals the rate at which radicals are repaired.

I. Introduction

Microdosimetry (the study of energy deposition by ionizing radiation in sub-cellular volumes) in the spatial domain has been widely explored during the past 30 to 40 years. Many papers have been written, and at least 11 symposia have been attended by eminent scientists who have devoted their lives to the study of energy deposition in microscopic volumes. However, despite the numerous studies conducted which concern the deposition of energy in microscopic volumes in space, no experimental studies have been found which take into account the deposition of energy in the time and spatial domain. The primary reason for this is that many of the studies deal with the energy characteristics of single particles. However, few studies deal with the energy deposition from radiation fields or fluxes. Indeed, simulations have been written to determine the space and time evolution of ionization in matter^[1]. Yet experimental evidence of energy deposition over short periods of time is lacking. There is a need, therefore, to gain an understanding of energy deposition in space and time by measuring energy deposition frequency distributions as a function of charge collecting time, $f(z, \Delta t)$. This understanding will be gained by simulating tissue equivalent spherical volumes down to diameters of 500 nm for collection times on the order of nanoseconds. These measurements will be conducted in a uniform field as opposed to fluxes of single particles. The investigator will examine microdosimetric parameters as a function of time using a gas-filled detector of variable charge falloff time and a gated signal.

II. Objectives

The three-fold objective of this experiment is to modify an existing microdosimeter to operate at variable charge collection times, use this device to measure energy deposition from uniform radiation fields in a spherical volume as a function of deposition time, and relate these data to time-specific biological repair processes in cells.

III. Significance

The relevance of microdosimetric measurements is clear when considered from a biological standpoint. In a uniform radiation field, the probability of radiation interaction in a small tissue volume, like a cell nucleus or chromosome, is very important in considerations of DNA damage, carcinogenesis, and cell death. However, the

time during which this energy is deposited has received little attention in microdosimetric measurements. The importance of this becomes evident when one considers the time scale of cellular repair, carcinogenesis, and death. The dual-radiation action theory involved in cellular DNA damage would dictate that two passages of radiation can occur to cause a double strand break (DSB) in DNA, while in some cases, a single passage will cause a single strand break (SSB). Higher LET radiation can cause a DSB with a single particle passage.^[2] However, for lower fluxes (dose rates) DNA damage may be less severe due to repair of SSBs by cells. Should a single strand break occur, if repair of the break takes place before another radiation passage and additional energy deposition, then a DSB will not occur. In this case, the time scale of DNA repair becomes extremely important.

The initial deposition of energy during a traversal of DNA by a particle will take place in roughly 10^{-18} s.^[2,3] Radical products are produced in approximately 10^{-12} s.^[2,3] Indirect damage of DNA is complete in the range of about 10^{-8} to 10^{-6} s.^[3]

Biological repair of SSBs in mammalian cells has been shown to occur at a rate of several minutes to several hours. This repair rate depends highly on the number of SSBs in a cell and breaks are repaired with a half-life on the order of hours.^[4] Simpler SSBs are repaired through the excision repair process, controlled through the action of the DNA ligase enzyme.^[5] In *E. coli* bacteria, the repair of a SSB takes approximately 2 seconds per enzyme molecule in the cell.^[6] At these repair rates, it would seem pointless to consider time dependence of energy deposition in a microscopic volume since many uniform fields will result in an energy deposition rate causing a rate of endpoint production much greater than the biological repair rate in cells. However, chemical repair rates of SSBs are much faster.

In many cases, up to 65% of all radiation induced strand breaks in a cell are the result of indirect action.^[7] Repair, or neutralization, of free radical sites on or near the DNA takes place before a break occurs, preventing the actual formation of breaks, but still counteracting the action of radiation near the DNA.^[8] Therefore, the indirect action of radiation is counteracted. This is of concern in considering the importance of indirect action from low LET radiation. The $\cdot\text{OH}$ radical, the primary product inducing strand breaks by indirect action, is also ablated quickly through chemical processes. In V79 cells (hamster), the repair of radical sites happens on the order of 10 ms^[8], while in bacterial cells, the reaction rate of radicals and scavengers occurs during a time span on the order of microseconds or even hundreds of nanoseconds.^[7] For lower LET radiation, radical repair can occur on the order of 100 ns.^[9]

Therefore, any energy deposited within about 100 ns of the initial traversal can result in cell damage from double strand breaks due to the indirect effect. Critical parameters, therefore, are the energy necessary to induce radical formation and the time during which this energy is deposited in the volume of interest.

IV. Background

IV.I. Microdosimetric Quantities

The fundamental goal of experimental microdosimetry is to simulate tissue equivalent volumes on the order of cellular and subcellular sizes in a detector or a series of detectors to measure energy deposition in these volumes. Detector geometries are typically cylindrical or spherical. For a spherical detector, the energy deposited is heavily dependent on the path length of the particle through the detector, therefore, one would expect the energy distribution to be similar in shape to the chord length distribution of the detector. For a spherical detector, the chord length distribution for a particle emitted from a point source and traveling a straight line through the detector is:

$$P(x) = \frac{x}{\sqrt{1 - \left(\frac{r}{a}\right)^2} \left[1 - \left(\frac{x}{2r}\right)^2 \right]} \int_0^{2r} \frac{x dx}{\sqrt{1 - \left(\frac{r}{a}\right)^2} \left[1 - \left(\frac{x}{2r}\right)^2 \right]} \quad (1)$$

where x is the chord length, r is radius of the sphere, and a is the source distance from the center of the sphere.^[10] If $a \gg r$ then

$$P(x) = \frac{x}{\int_0^{2r} x dx} = \frac{x}{2r^2} \quad (2)$$

and for $a=r$

$$P(x) = \frac{x}{\frac{x}{2r} \int_0^{2r} 2r dx} = \frac{1}{2r} \quad (3)$$

So for the common case in which $a \gg r$, the chord length distribution is triangular. In the special case of $a=r$, the chord length distribution is constant. The distribution for $a \gg r$ is shown in Figure 1.^[11]

It has been shown that the energy deposition spectrum of single particles will also appear triangular (see Figure 1), with some higher-energy events due to scattering within the detector and more low-energy events due to partial energy deposition in the detector from events occurring outside the detector.^[11] This distribution is useful in determining the probability of

energy deposited by a particle passage in a tissue sphere of the same total interaction cross section (directly related to the total mass) as the fill gas in the detector active volume.

When charge collected on the anode is the experimental measurement taken, the primary quantity of interest in microdosimetric research is the specific energy, z (microscopic dose). This is defined as the sum of all energy deposits ϵ_i to a microscopic volume S with a mass M ^[11]:

$$z = \frac{\sum \epsilon_i}{M} \quad (4)$$

Energy deposition in a given experimental measurement is calculated by taking the total charge collected on the anode, converting to charge deposited in the detector by taking into account the charge collection efficiency and the charge multiplication of the detector, and multiplying by the electron charge to obtain the total ion pairs liberated in the detector by the incident particle. The energy deposited is then the number of ion pairs in the detector multiplied by the energy necessary to create a single ion pair for the specific detector gas used (the W -value, typically around 34 eV per ion pair). W -values for some gases are given in Table 1. The mass of the detector is easily calculated by taking the volume of the detector and multiplying by the corrected fill gas density. For an individual event, that is, the passage of a single particle through the detector volume, the specific energy is expected to vary with the track length of the particle through the detector volume. Many recorded events will then result in a specific energy distribution similar to that shown in Figure 1.

Table 1. W -Values for Various Gases^[12]

<u>Gas</u>	<u>W-value</u> (ev/ion pair)
Ne	36.2
Ar	26.2
Xe	21.5
Ne + .5% Ar	25.3
Ar + .5% C ₂ H ₂	20.3
Ar + .8% CH ₄	26
Ar + 10% CH ₄ (P-10)	26

Another important value in microdosimetric measurements is the lineal energy, y (microscopic LET). This is defined to be the energy deposited for a single event ϵ_i in a volume divided by the mean chord length in that volume^[14]:

$$y = \frac{\epsilon_i}{\bar{x}} \quad (5)$$

for a sphere, the mean chord length is $4r/3$ as calculated by taking the frequency average (first moment) of $P(x)$:

$$\bar{x} = \int_0^{2r} xP(x) dx \quad (6)$$

For the common case in which $a \gg r$:

$$\bar{x} = \int_0^{2r} x \left(\frac{x}{2r^2} \right) dx = \frac{4r}{3} \quad (7)$$

This results in a lineal energy for a particle passage of $3\epsilon/4r$.

Detectors used for microdosimetric detection can be of the same type used in "macroscopic" radiation detection, except for the differences in size and fill gas pressure. Since the actual detector is typically much smaller, the multiplication region is also smaller. This makes it necessary to use a smaller anode, and sometimes a lower voltage. In the past, detection methods used have been primarily gas-filled detectors. Although solid state detectors have been used,^[15,16,17] as well as optical imaging of particle tracks,^[18] cloud chamber techniques, and even electron spin resonance (ESR) processes,^[19] the detector of choice is the gas-filled detector.^[20] Gas-filled detectors are used for their ability to simulate tissue equivalence with the choice of fill gas. Organic fill gases work well as a proportional gas since the incident radiation interacts in a similar manner as with tissue. For gas-filled detectors, a typical fill gas would be P-10, a mixture of methane and argon.^[12] In addition to P-10, experimenters also use mixtures containing methane, nitrogen, and carbon dioxide in order to create a tissue equivalent active volume.^[21] Proportional counters, although limited by larger volumes, can be made "wall-less" to avoid effects of the non-tissue equivalent wall.^[20] Other gas-filled detectors include the ionization chamber, Long detectors, and Rossi counters. For this experiment, the fill gas used was a tissue equivalent (T.E.) mixture of 64.4% methane (by molar volume), 32.5% carbon dioxide, and 3.1% nitrogen gas.

It would be extremely difficult to construct a counter of small enough physical dimensions to measure radiation deposition in the microscopic volumes of interest. The small tissue equivalent sizes in physically large detectors are simulated by constructing the detection volume as small as possible, using low gas pressure in the counter, using a low density gas, and simply by virtue of using a gas (of lower density than tissue). For example, if the chamber's active volume has a diameter d_c , then the average energy deposited in the chamber from the passage of a charged particle is^[20]:

$$E_c = \frac{2}{3} d_c L_w \left(\frac{P}{760} \right) \left(\frac{\rho_g}{\rho_w} \right) \quad (8)$$

where L_w is the lineal energy of the particle in water, P is the gas pressure in the chamber, and ρ_g/ρ_w is the ratio of the density of the fill gas to that of water at one atmosphere. ($2/3d_c$ is the average chord length of the sphere.) For tissue equivalence, the tissue is assumed to be mostly water at one atmosphere.

It is possible to calculate the specific energy in an equivalent tissue sphere based on the total energy deposited in the detection volume and the mass of the tissue equivalent sphere. The ratio of the specific energy deposited in the detector to that deposited in an equivalent volume of tissue is:

$$\frac{z_c}{z_t} = \frac{M_t}{M_c} \quad (9)$$

or:

$$z_t = \frac{z_c M_c}{M_t} \quad (10)$$

and, substituting with Equation 4:

$$z_t = \frac{E_c}{M_t} \quad (11)$$

where E_c is the total energy deposited in the chamber. M_c and M_t are the masses of the chamber and tissue equivalent sphere respectively. Using Equation 8, the simulated size of a tissue equivalent sphere is calculated by the following:

$$E_t = E_c \quad (12)$$

$$\frac{2}{3}d_t L_w = \frac{2}{3}d_c L_w \left(\frac{P}{760} \right) \left(\frac{\rho_g}{\rho_w} \right) \quad (13)$$

$$d_t = d_c \left(\frac{P}{760} \right) \left(\frac{\rho_g}{\rho_w} \right) \quad (14)$$

IV.II. Uncertainty in Measurements

Variance is defined to be the square of the standard deviation of a distribution. Variance is a simple index of the width of a distribution. The relative variance of a distribution is given by the ratio of the variance to the average value of the distribution (first moment):

$$V = \frac{\sigma^2}{\int x f(x) dx} \quad (15)$$

Rossi and Zaider have enumerated six factors that contribute to the variance of energy distributions in microdosimetric measurements.^[14] These are:

- Distribution in the number of (energy deposition) events
- LET distribution of individual particles
- Distribution of path lengths of particles through a site (discussed above)
- Distribution of number of collisions
- Distribution of energy imparted in individual collisions
- Distribution of fraction of energy retained in the site (not escaping as secondary events)

While the first three are properties of all particles interacting in matter, the last three are properties of individual particles depositing their energy.

The distribution of the number of events can follow a Poisson distribution based on the total incident particles in a site and the average number of energy deposition events per incident particle or radiation. It must be kept in mind that y , the lineal energy of an event, is only due to a single energy deposition event from a particle, and therefore, the relative variance in the number of energy deposition events, V_n , is equal to zero. This is not true in a radiation field, however, since energy deposition is being measured, and this is based on the total energy deposited for incident particles. The distribution in the number of collisions V_c also follows a Poisson pattern.^[14]

For monoenergetic particles, the distribution of particle LET V_L is zero. However, for a spectrum of energy, such as that created by the photons from an x-ray machine, the relative variance depends on the incident particles.

The last two items in the above list follow complex physical mechanisms depending on the particle velocity, mass, charge, and many other interaction parameters.^[14] These phenomena are not well understood, and values are taken empirically.

Given the above values, the total relative variance of a spectrum is then^[14]

$$V_T = V_v + V_L + V_t + V_L V_t + V_c + pV_\varepsilon \quad (16)$$

where V_ε is the relative variance of the variance of the fraction of deposited energy not escaping the site, and p is a factor accounting for δ -ray escape. Despite the potential utility of Equation 16, the inadequate information of the above quantities dictates that the total variance is determined by spectral measurements.^[14]

V. Approach and Methods

Since the primary purpose of this work is to determine energy deposition in microscopic volumes for various time spans, it is necessary that the detector signal be toggled on and off precisely. This is done with a digital signal and will be discussed shortly. Energy deposition is measured for time spans ranging from a microsecond to several milliseconds. Several measurements have been taken, and the energy deposited is registered on a spectrum. As well, several radiation fields have been selected. These include the Maxitron 300 x-ray machine owned by the University of Michigan (UM) Department of Environmental and Industrial Health (EIH). Measurements were taken at a kVp setting of 250 and a current setting of .5 mA with an MFI filter packet (1 mm of aluminum, followed by .25 mm of copper, followed by 4 mm of aluminum) and at a distance of 2 m. In addition to the x-ray source, the ^{137}Cs source located at the EIH department was also used with measurements taken at a distance of 2 cm. This source has an activity of 26.4 mCi (as of March 1996). Measurements have also been taken with the $^{90}\text{Sr}/^{90}\text{Y}$ source located at the UM Willow Run facility; measurements were taken at a distance of 13.6 cm. The ^{90}Sr activity of the source is 21 mCi as of 1996; the ^{90}Y activity, which is in a state of secular equilibrium, has the same activity.

The radiation detector utilized for these experiments must be able to collect charge for various time spans. These time spans must be intrinsic to the detector and not the measuring system. In addition, proportionality must be preserved. That is, the output voltage amplitude must be proportional to the total charge collected by the detector. Because measurements are conducted in a uniform energy field, it is quite possible for multiple particles to deposit energy in the detector during the charge collection time (and this is expected in many cases). In this case, the total energy will be collected, rather than energy values from individual particle passages. In this situation, pulse pileup is desirable. Furthermore, it is necessary to be able to gate the detector anode signal for time spans varying by many orders of magnitude.

VI. Equipment

VI.I. Microdosimeter

The detector used for this experiment is the LET-1/2 spherical tissue equivalent proportional counter made by Far West Technology, Inc. The detector consists of a tissue equivalent (TE) plastic sphere housed in a grounded aluminum shell. The aluminum shell serves as a vacuum tight housing and is capable of being submersed. The anode of the spherical detection volume is surrounded by a helix held at a slightly lower potential to create a more uniform electric field in the proportional region of the detector. This detector is described in detail in Appendix A.

The active volume of the detector is defined by a Shonka Type A-150 sphere of thickness 0.127 cm and an inside diameter of 1.27 cm. The plastic employed to define the active volume consists of elemental compositions listed in Table 2. Compared to ICRU tissue, much of the oxygen has been replaced with carbon while maintaining roughly the same amount of hydrogen and nitrogen. Given a density of 1.12 g/cm³, the density thickness is calculated to be 142 mg/cm². This thickness would be of little concern for higher energy photons, and the tissue equivalence of the plastic will also enable a good simulation of tissue with the counter.

Table 2. Elemental Compositions in Shonka A-150 Plastic and ICRU Tissue

	<u>Percent by Weight^[14]</u>					
<u>Material</u>	<u>H</u>	<u>C</u>	<u>N</u>	<u>O</u>	<u>F</u>	<u>Ca</u>
ICRU Tissue	10.1	11.6	2.6	76.2	-	-
A-150 Plastic	10.2	76.8	3.6	5.9	1.7	1.8

It is possible that particles interacting in the detector gas outside the active volume may generate unwanted electrons in the active volume, thus generating unwanted signal (dose). For an ideal detector, the Bragg-Gray principle must be satisfied. That is, the active volume of the detector must be small compared to the range of secondary charged particles produced in the medium so that the particle flux is not greatly affected. In addition, the surrounding medium must be large compared to the range of secondary charged particles so that electronic equilibrium is established.^[12]

In the case of Compton scattered electrons from the wall into the detector, the maximum electron energy occurs from backscattered photons:^[12]

$$E_{e-}|_{\theta=\pi} = h\nu \left(\frac{2h\nu/m_0c^2}{1+2h\nu/m_0c^2} \right) \quad (17)$$

For 250 kVp x-ray photons, the maximum energy of a recoil electron is calculated to be 0.124 MeV. Since the range of electrons in matter depends only slightly on the material,^[22] a general value can be obtained for a range from the empirical formulas found in the *Radiological Health Handbook*.^[23]

$$R = 412E^{1.265-0.0954 \ln E} \quad (18)$$

and:

$$\ln E = 6.63 - 3.2376 \sqrt{10.2146 - \ln R} \quad (19)$$

where R is the range in mg/cm^2 and E is the electron energy in MeV. Electrons with energy of 0.124 MeV have a range of roughly $19.4 \text{ mg}/\text{cm}^2$. Compton electrons produced on the inside of the aluminum detector housing do not penetrate to the active volume of the detector. The density thickness of the detector cavity is calculated using the average chord length of the detection volume multiplied by the density of the gas:

$$t = \frac{2}{3}d\rho = \frac{2}{3}d\rho_{atm}\left(\frac{P}{760}\right) \quad (20)$$

where the gas is corrected for pressure change. In order to satisfy the first requirement of the Bragg-Gray principle, pressure cannot exceed a maximum value given by the following:

$$P \ll \frac{3}{2}\left(\frac{760}{d\rho_{atm}}\right)R_{e^-} \quad (21)$$

Given the detection volume with a diameter of 1.27 cm filled with a tissue equivalent gas with a density of $1.064 \times 10^{-3} \text{ g}/\text{cm}^3$ (at one atmosphere), and the above calculated maximum Compton electron range, then the chamber pressure must be less than $1.64 \times 10^4 \text{ mm}$ (or 17 atmospheres). Since the detector will be operated at pressures on the order of 10s of mm, then the detector volume most certainly satisfies the first condition of the Bragg-Gray Principle.

The second condition requires the thickness of the surrounding medium to be larger than the range of the secondary electrons, so that electronic equilibrium is established. The surrounding medium is taken to be the wall surrounding the active volume and the fill gas between the wall and the aluminum housing. However, the total thickness can be taken to be equal to the thickness of the plastic wall surrounding the active volume. Because the gas-filled region between the outer housing and the plastic wall is 0.270 cm thick, and since the detector is operated at a gas pressure of 58.75 torr, then the density thickness of this region is calculated to be $0.22 \text{ mg}/\text{cm}^2$ - negligible in comparison to the density thickness of the plastic wall. The plastic wall of the active volume has a thickness of 0.127 cm, or $142 \text{ mg}/\text{cm}^2$. The minimum pressure necessary in the chamber such that it can be treated as a Bragg-Gray cavity with 0.124 MeV electrons is then calculated by:

$$142 + t_g\rho_{atm}\left(\frac{P}{760}\right) < R_{e^-} \quad (22)$$

However, if gas pressure is neglected, then the inequality simplifies to:

$$R_{e^-} > 142 \quad (23)$$

Compton scattered electrons of 0.124 MeV have a range of $19.4 \text{ mg}/\text{cm}^2$, much less than the thickness of the plastic shell. The plastic shell is quite important in maintaining electronic equilibrium in this case.

However, the situation is slightly different for ^{137}Cs gamma rays (0.662 MeV). The maximum energy of Compton electrons is calculated to be 0.478 MeV. Using Equation 21, the range of these electrons is then calculated to be $153.7 \text{ mg}/\text{cm}^2$. Obviously, the TE shell would not satisfy the Bragg-Gray Principle as the

secondary electrons are able to penetrate to the active volume of the detector. However, in order to maintain electronic equilibrium, electrons produced on the inside wall of the aluminum detector housing must have a range less than the thickness of the chamber wall and the gas between the housing and the wall.

This problem is dealt with by examining the magnitude of secondary electrons penetrating to the active volume of the detector. It is assumed that the contribution to density thickness from the gas surrounding the active volume is negligible. For Compton electrons produced outside of the plastic sphere, the minimum energy needed to penetrate the sphere is 0.452 MeV.^[23] Using the general equation for Compton electron energy:

$$E_{e^-} = h\nu \left(\frac{(h\nu/m_0c^2)(1-\cos\theta)}{1+(h\nu/m_0c^2)(1-\cos\theta)} \right) \quad (24)$$

a minimum scattering angle of 131.5° is necessary for electrons to penetrate the wall of the sphere with an energy of zero MeV upon entering the active volume.

The differential cross section for Compton scattering is given by the *Klein-Nishina Formula*.^[22]

$$\frac{d\sigma_c}{d\Omega} = r_0^2 \left[\frac{1}{1+\alpha(1-\cos\theta)} \right]^2 \left[\frac{1+\cos^2\theta}{2} \right] \left[1 + \frac{\alpha^2(1-\cos\theta)^2}{(1+\cos^2\theta)[1+\alpha(1-\cos\theta)]} \right] \quad (25)$$

where r_0 is the classical electron radius:

$$r_0 = \frac{e^2}{4\pi\epsilon_0 mc^2} = 2.818 \text{ fm} \quad (26)$$

and α is the photon energy in units of electron rest energy:

$$\alpha = \frac{h\nu}{m_0c^2} \quad (27)$$

Integration of Equation 25 over all solid angles yields the total Compton scattering cross section for one electron:

$$\sigma_c = \frac{\pi r_0^2}{\alpha} \left\{ \left[1 - \frac{2(\alpha+1)}{\alpha^2} \right] \ln(2\alpha+1) + \frac{1}{2} + \frac{4}{\alpha} - \frac{1}{2(2\alpha+1)^2} \right\} \quad (28)$$

However, in the case of energy deposited in the active volume of the detector from electrons produced outside the active volume of the detector, one is not interested in Compton electrons produced by photons which scatter at angles less than 131.5° (2.295 radians), since these electrons will not have a range greater than the wall thickness. Rather, it is necessary to integrate the *Klein-Nishina Formula* over only the solid angles in which θ lies between 131.5° and π :

$$\sigma_c = \int_0^{2\pi} \int_{\theta_{\min}}^{\pi} \frac{d\sigma_c}{d\Omega} d\Omega \quad (29)$$

where $\theta_{\min}$ is the minimum scattering angle necessary to introduce Compton electrons into the active volume of the detector. Using the substitution $u=\cos\theta$ yields:

$$\sigma_c = \int_0^{2\pi} \int_{-1}^b r_0^2 \left[\frac{1}{1+\alpha(1-u)} \right]^2 \left[\frac{1+u^2}{2} \right] \left[1 + \frac{\alpha^2(1-u)^2}{(1+u^2)[1+\alpha(1-u)]} \right] du d\phi \quad (30)$$

where $b=\cos(\theta_{\min})$. Integrating Equation 30 yields the Compton scattering cross section for only those photons that produce electrons with enough energy to penetrate the wall of the detector from any point of production outside the detector wall:

$$\sigma_c = \frac{\pi r_o^2}{\alpha} \left\{ \left[1 - \frac{2(\alpha+1)}{\alpha^2} \right] \ln \left(\frac{2\alpha+1}{\alpha+1-\alpha b} \right) + \frac{2+5\alpha^2-4\alpha^3 b^2+2\alpha^3 b+2\alpha^3 b^3+6\alpha-4\alpha^2 b^2}{2\alpha^2(\alpha b-\alpha-1)^2} + \frac{\alpha-1}{\alpha^2} - \frac{1}{2(2\alpha+1)^2} \right\} \quad (31)$$

The integrated cross section from $\theta_{\min}$ to 180° is illustrated in Figure 2. It is seen that the total cross section for photons scattered at a minimum angle of 131.5° is quite small compared to the total interaction cross section for Compton interactions (that is, photons scattered at a minimum angle of 0°). Of course, the interaction cross section for photons scattered at 180° approaches zero. For Compton electrons produced at photon scattering angles greater than 131.5° , the scattering cross section for each individual electron produced by .662 MeV gamma rays is calculated to be $0.3139r_o^2$ or 0.0249 b. The Compton cross section for scattering at all angles is $3.226r_o^2$ or $.256$ b.

The total Compton absorption coefficient is calculated using the calculated cross section for each individual electron according to the following^[22]:

$$\sigma = \sigma_c N Z \quad (32)$$

where Z and N represent the atomic number of the atoms in the medium and the number of atoms per unit volume.

It is seen that σ_c has dimensions of $[\text{length}]^2$ and N has dimensions of $[\text{length}]^{-3}$, resulting in dimension of $[\text{length}]^{-1}$ for σ . For a mixed material, the cross sections for each individual element are calculated and summed. For P-10 gas at one atmosphere and room temperature, $N_C=2.485 \times 10^{24} \text{m}^{-3}$, $N_H=9.941 \times 10^{24} \text{m}^{-3}$, and $N_A=2.236 \times 10^{25} \text{m}^{-3}$. For a chamber pressure of 25 mm, the values for N_C , N_H , and N_A become $8.174 \times 10^{22} \text{m}^{-3}$, $3.270 \times 10^{23} \text{m}^{-3}$, and $7.355 \times 10^{23} \text{m}^{-3}$ respectively. The tissue equivalent gas used in this experiment has $N_C=2.408 \times 10^{25} \text{m}^{-3}$, $N_H=6.401 \times 10^{25} \text{m}^{-3}$, $N_O=1.615 \times 10^{25} \text{m}^{-3}$, and $N_N=1.540 \times 10^{24} \text{m}^{-3}$. In order to simulate a one micron tissue sphere, a pressure of 58.75 torr is calculated (using equation 14). For this pressure $N_C=1.853 \times 10^{24} \text{m}^{-3}$, $N_H=4.925 \times 10^{24} \text{m}^{-3}$, $N_O=1.242 \times 10^{24} \text{m}^{-3}$, and $N_N=1.185 \times 10^{23} \text{m}^{-3}$. An effective Compton cross section can be calculated using:

$$\sigma = \sum_i \sigma_c N_i Z_i = \sigma_c \sum_i N_i Z_i \quad (33)$$

For the aforementioned T.E. gas at low pressure, the total Compton cross section, σ_T , is calculated to be $6.863 \times 10^{-4} \text{m}^{-1}$, and the cross section for Compton electrons that penetrate the detector wall, σ_w , is calculated to be $6.675 \times 10^{-5} \text{m}^{-1}$.

The differential loss in photon intensity as it traverses a distance dx is calculated by:

$$\frac{dI}{I} = -\sigma dx \quad (34)$$

Integrating the equation, and setting $I(x=0)=I_o$, the initial intensity of the incident photon beam results in:

$$I = I_0 e^{-\sigma x} \quad (35)$$

(Buildup was not considered as the amount of material between the source and the active volume was small.) The distance between the aluminum housing and the plastic sphere surrounding the active volume of the detector is 0.270 cm. Therefore, for a flux of ^{137}Cs photons incident on the detector filled with T.E. gas to a pressure of 58.75 torr, 0.97% are converted to Compton electrons. 0.095% interact to produce Compton electrons that penetrate the active volume wall. It is seen, then, that less than 10% of all Compton electrons produced in the chamber penetrate the wall of the detector into the active volume. This means that the detector is not quite in electronic equilibrium due to the few Compton electrons which penetrate the active volume from outside the plastic wall of the detector. So, an offset in the anode signal will occur, causing a positive shift in the spectrum. However, the total number of Compton electrons produced in the active volume will also cause a signal. The average number of Compton electrons produced in the active volume can be calculated using Equation 35. Again, for the conditions of 0.662 MeV gamma rays incident on the detector filled with T.E. gas to a pressure of 58.75 mm, 3% interact to produce Compton electrons. It is seen that only 3% of the Compton electrons which are formed (or enter) the active volume of the detector are produced outside the active volume. Since this is such a small value, the actual signal is not expected to be degraded much from a signal produced by the same detector at electronic equilibrium.

The kinetic energy of electrons produced via the photoelectric effect is calculated using the following equation:^[12]

$$E_{e-} = h\nu - E_b \quad (36)$$

Photon Compton Cross Section as a Function of Scattering Angle 0.662 MeV Photons in P-10 Gas

Figure 2. Photon interaction cross sections for interactions that produce Compton electrons which penetrate the detector active volume. Cross Section (Barns) vs. Minimum Scattering Angle.

where E_b is the binding energy of the electron in the material. For 250 kVp x rays, it is not possible to produce a photoelectron anywhere in the detector such that the range of the electrons exceeds the detector wall thickness. This is because E_b and $h\nu$ are always positive, so the photoelectron energy will never exceed 250 keV, the maximum photon energy. Electrons of this energy have a range of much less than the thickness of the wall surrounding the detector active volume. The cross section for photoelectric absorption of gamma rays is difficult to calculate,^[22] but can be roughly approximated by:

$$\tau \propto \frac{Z^n}{(h\nu)^{3.5}} \quad (37)$$

where n varies between 4 and 5.^[12] For aluminum, with an atomic number of 13, the total photoelectric cross section for 0.663 MeV gamma rays is much less than 0.001 cm²/g. Knowing this, then the cross section for the fill gas, of lower Z , and much lower density thickness is expected to be quite low, and very few photoelectrons are produced in either the aluminum casing or the fill gas of the detector, so the production of photoelectrons outside the active volume of the detector is neglected.

Detector calibration and count curves will be discussed in a following section.

VI.II. Gas Filling System

The detector can be filled with a gas of choice using the accompanying gas filling system, model GFS-1 - also made by Far West Technology, Inc. The system consists of a mechanical vacuum gauge, vacuum control valves to control gas flow into the detector, and connections to facilitate the filling of the detector. A rubber hose connection is made to the vacuum pump to evacuate the detector, gauge, and filling tubes. The detector and gas tubing is connected to the filling system via a Swagelok connection (model B-QC4-D-400). The gauge measures absolute air pressure from 0 to 200 torr, and can accurately be read to the nearest 0.2 torr. The gas filling system is diagrammed in Figure 3 and described in detail in Appendix B.

VI.III. Base Electronics

The detector base is configured as shown in Figure 4. C_1 is equal to 0.01μf; C_2 is 0.1μf. R_1 , R_2 , and R_3 are equal to 3.9MΩ, 30MΩ, and 6.8MΩ, respectively. The purpose of the coil is to create a more uniform

cylindrical electric field around the collecting anode. The voltage on the coil is calculated using a simple voltage divider relationship:

$$V_{coil} = H.V. \left(\frac{R3}{R3+R2} \right) \quad (38)$$

Given the values of R2 and R3, then the coil voltage is equal to 0.185H.V. The anode voltage is equal to the high voltage H.V.

VI.IV. Signal Integrator

The heart of the signal collection system is the signal integrator. The device acts as a continuous summing circuit for all detector signals. The integrator unit consists of two primary parts, the feedback amplifier which is the actual integration unit of the device, and the timing logic circuits. These will be discussed separately, and explanation of their concurrent function will follow.

The core of the feedback amplifier circuit is an ACF2101 Dual Switched Integrator Chip (BP model, DIP). This is a 24 pin integrated circuit produced by Burr-Brown. Data sheets for this chip are presented in Appendix C. The precise function of this chip (low charge transfer, low drift, low noise, and high speed) make it ideal as a signal integrator. The input signal is fed into the inverting input of the circuit (pin 6). Integration takes place during the period of time in which the *reset* switch is in the open position. The *hold* and *select* switches can also be used to control the output signal of the chip. Functioning of the switches is accomplished through TTL signals. In addition, the switches operate on an "inverse logic" basis. That is, a switch is open during logic high and closed during logic low. The chip operates at -12V and +5V. The ACF2101 chip schematic and diagram is shown in Figure 5.

Three characteristics of the integration chip that must be considered are the charge transfer, the signal droop, and the switching time. For all solid state switches in the integrator, the charge transfer is listed to be in the range of 0.1 to 1 pC. This transfer is most vital for the reset switch in parallel with the feedback capacitor (see Figure 5). The rated typical charge transfer for this switch is 0.1 pC. Since the feedback capacitor is 100 pf, then the voltage offset from the charge transfer is expected to be equal to the quotient of the charge transfer and the feedback capacitance, or one mV. This is very low compared to a typical pre-amplifier output voltage level, so the

charge transfer is not a problem.

The signal droop, the rate at which the held output voltage returns to baseline is listed in the data sheets to be typically 1 nV/μs. In order for this droop to be insignificant, the feedback capacitor voltage (or output voltage) must be much greater than the product of the droop and the integration time. Since the output signal is expected to droop 5 μV in 5 ms (the longest used integration time), the charge collected on the capacitor

Figure 4. LET-1/2 Base Electronics.

must exceed 5×10^{-4} pC. This is an extremely low value, and the charge is expected to easily exceed this value, as will be discussed shortly.

The switching time is the time necessary for a switch to completely change state. For all switches, this time is 200 ns. This value places a limit on the speed with which the chip is able to be operated. It is impossible to operate the device for integration times less than the switching time, as the switch will not change state fast enough. Rather, the minimum integration time must be greater than twice the switching time to allow for a complete change of state for the falling edge and the rising edge of the triggering signal.

Only one integrator of the chip is needed. In order to create an integrator, the *Cap A* and *In A* pins (pins 4 and 5) are connected together. The input signal enters the *Sw In A* pin (pin 6). *Com A* (pin 3) is grounded, and the output signal leaves through *Out A*. In order for the input (Hold) switch to maintain a constant resistance, as is

shown in the data sheets, the input current must be maintained between $-100 \mu\text{A}$ and $100 \mu\text{A}$. To accomplish this, a suitable input resistance is connected in series with the input signal. All input signals are positive, as will be discussed shortly, so the input current must be less than $100 \mu\text{A}$. Observation of these signals from the sources of choice for this experiment reveal that the maximum signal amplitude entering the integrator is $\sim 2.5\text{V}$. Therefore, the input resistance was selected to be $56 \text{ k}\Omega$, which results in a maximum current of nearly $50 \mu\text{A}$.

The timing circuit to drive the signal integrator switches is shown in Figure 6. It consists simply of three bistable monovibrators (one-shots) constructed from timer ICs. The voltage converter in the circuit is placed such

that the device can be operated from a NIM power module. Triggering of the circuit is done through the use of a TTL square wave. The *select* and *hold* signals are directly triggered by the input square wave, while the *reset* signal is triggered by the inverse of the *select* signal. The values of the components in the circuit are listed in Table 3.

The time that the output pulses stay high is given as a function of the product of the output resistance and capacitance. In the case of the Texas Instruments timers used in this experiment, the output pulse stays high for a duration equal to $1.1RC$.

Table 3. Component Values for the TTL Timing Circuit

<u>Component</u>	<u>Value</u>	<u>Units</u>
R_s	10	Ω
R_H	1	$M\Omega$
R_R	100	$k\Omega$
C_s	22	nf
C_H	22	nf
C_R	22	nf
C_{OUT}	22	nf
R_{OUT}	2.2	$k\Omega$

Figure 6. TTL Timing Circuit for Signal Integrator.

The combined circuit including the timing circuit and the integration circuit were mounted to perf-board

and encased in a steel box. Four connections to the box, in addition to the power connections were made with BNC connectors. These connections are the input signal connection, the integrated output signal connection, the trigger signal connection, and the hold signal connection, so that the width of the integration cycle can be easily determined

Figure 7. Basic Circuit Diagram of Timing and Integration Unit.

by examining the hold signal with an oscilloscope. The basic circuit layout is shown in Figure 7. In addition to the signal, trigger, and hold connections, three power connections were made to the circuit and crimped to plugs so that power could be drawn from the NIM bin being used at the time. The entire unit uses +12V, -12V, and ground connections in the NIM bin.

The circuit logic is shown in Figure 8. Since the solid state switches of the integrator chip operate on inverse logic, a switch closes when the timing signal drops low. The duration of the integration time is determined by the difference between the period of the trigger signal and the time that the hold circuit stays high, determined by the RC constant of the timing one-shot. The signal collection sequence is as follows:

- The *hold* switch is closed, while the *select* and *reset* switches are both open, causing the integrator to integrate signal.
- The *hold* signal is triggered high by the square wave pulse, while *select* is triggered low. This opens up the output, and a pulse is read from the integrator.
- The *reset* switch closes after about 8 μ s, causing the feedback capacitor to discharge, and the output pulse drops.
- The *select* switch goes high after a time determined by the RC constant of the one-shot, triggering the *reset* switch to open, which stays high for a period of time also determined by the RC constant of the one-shot.
- The *hold* signal then goes low (after a time also determined by the RC constant of the circuit) and the integration sequence begins again.

The duty cycle for the hold signal varies with integration time. This is due to the fact that the time during which the signal is high has a minimum value proportional to the RC constant of the circuit. If the period of the triggering signal is larger than the time the signal is high, then the signal will stay high for its minimum value, then

drop (and begin integrating), until it is triggered high again. If the period of the triggering signal is smaller than the time the signal is high, then the signal will stay high regardless of the triggering input. This has the effect of causing the integration signal to be triggered on every second

(or third, etc.) triggering pulse, rather than with every triggering pulse. The duty cycle is plotted as a function of time in Figure 9. As the integration time increases, the duty cycle approaches 50%, since the time that the signal is low is very close to the time that the triggering wave is high. As the integration time approaches zero, the duty cycle also approaches zero. This is of little consequence for the system except for the purpose of mentioning.

Testing of the circuit was done using a constant current source - in this case, a stable voltage source of .707V through a 22kΩ resistance. The input resistance of the integrator chip brings the total input impedance to 23.5kΩ. This results in a current of 30 μA. By setting the hold time (and hence, the integration time) to an integration time Δt, then the amplitude of the output pulse, which is expected to be a square pulse of width equal to 8 μs (the time lag between the select switch closing and the reset switch closing) is expected to be given by the following:

$$A = \frac{i \cdot \Delta t}{C_f} = \frac{i \cdot \Delta t}{100 \times 10^{-12}} \quad (39)$$

where C_f is the value of the feedback capacitance in the integrator. For an integration time of 5 μs, an output pulse of 1.5V is expected to result. Indeed, this is the case. The linearity of the integration can also be tested by disabling the *hold* and *select* switches such that they always remains closed. In this case, the integration time is the time during which the reset switch is open, and the output should resemble a ramp pulse which drops off at a rate equal to the maximum slew rate of the integrator op-amp. Using Equation 39 and the constant current source described, for an integration time set at 25 μs, the maximum amplitude of the triangle wave pulse is expected to be 7.5V. Again, the measured value corresponds to the calculated value.

Incorporation of the integrator unit into the counting system requires little more equipment than what is found in a basic counting system. The experimental setup is shown in Figure 10. Since the integrator integrates only in the negative direction, the second amplifier is necessary for pulse inversion. In addition, a series resistor is placed between the first amplifier and the integrator so that the current falls within accepted values. The

oscilloscope is necessary to read the hold signal so that the integration time can be easily determined. The first amplifier is placed in the system in order to eliminate the very long tail of the pre-amplifier. This tail is undesirable as it has a signal decay time of roughly 50 μ s, which is much longer than most signal integration periods. This would have the effect of collecting only part of the total signal from an event which - although takes place during the integration time interval - results in a pre-amplifier signal that is not totally collected during this interval. By having a much shorter pulse, events which take place during the integration time

interval can be totally collected, and the error resulting in collecting partial events is reduced. The output from the first amplifier is a Gaussian shaped pulse of the approximate form:

$$A(t) = A_0 e^{-\left(\frac{t-t_0}{\tau}\right)} \quad (40)$$

where τ is the time constant physically corresponding to the rise and fall time of the shaped pulses. In this case, τ is set to be 500 ns, which is low enough so that any input to the integrator is fully integrated for all integration times, unless the input occurs near a switching point, in which case it is partially integrated. This is expected, however, as the signal pulse train is of arbitrary frequency with time. It is seen that integration of Equation 40 results in a pulse whose amplitude is only dependent on A_0 , again a desirable affect.

VI.V. Square Wave Generator

The square wave is produced using a Wavetek 130 function generator. This device is capable of producing a sine wave, triangle wave, and square wave output of varying frequency, amplitude, and voltage offset. The frequency range is from 0 to 2.86 MHz. The maximum frequency corresponds to a minimum triggering pulse period of 349 ns. However, since triggering times of one microsecond will be used, and since the duty cycle of the timing circuit is 2.6% at one microsecond, the minimum triggering pulse period need only be 38.5 μ s, much larger than the minimum attainable frequency with the wave generator.

The amplitude of the generator is variable between 0 and 20V peak-to-peak. The square wave amplitude used for this experiment will be 5V_{pp}. In addition, the offset will be 2.5V, in order to produce a TTL signal with the generator.

VI.VI. Other Equipment

Aside from the irradiation sources used, the remainder of the equipment consisted of the NIM module equipment, the oscilloscope, and the MCA equipment for data analysis.

The NIM equipment used was as follows:

- Pre-Amplifier: EG&G Ortec 142A, S.N. 1673
- Amplifier: EG&G Ortec 571, S.N. 1889
- Inverting Amplifier: EG&G Ortec 575, S.N. 32
- NIM Bin: EG&G Ortec 4001C, S.N. 4351
- MCA: EG&G Ortec 916A ADC Card with EG&G *Maestro* Software (Model A65-BI)
WinMCA v. 2.08, *MCB.DLL* v. 2.22, *MCBShare.SYS* v. 2.00
- Oscilloscope: Tektronix 434, S.N. B540859
- High Voltage Source: EG&G Ortec 478, S.N. 51357

One important feature of the NIM bin is its power plug access at the front of the module. This allowed easy access to power for the integrator circuit and timing module, which were connected using the aforementioned plugs.

VII. Detector Counting, Calibration, and Spectra

Although the detector used is a proportional counter, its functioning is much different as compared to other counters of larger size and higher pressure. The interaction of radiation in the detector is, therefore, somewhat different than in larger detectors. An example is the calibration of the detector. In the case of many proportional counters, a photon source can be used (e.g. ⁶⁰Co). However, this relies on the fact that photons deposit all their energy in the active volume through scattering and collisional processes. In a smaller detector, such as the LET-1/2, photons do not deposit all their energy during a single traversal of the active volume. This calls for a somewhat different method of calibration, which will be described shortly. In this section, detector and spectral characteristics are discussed.

VII.I. Detector Counting Curve

Since the counting curve is intrinsic to the detector, it was not necessary to generate a counting curve with the entire counting system shown in Figure 10.

Rather, a typical counting system was utilized which incorporated the equipment shown in Figure 10 and outlined in Section VI.VI without the integrator, the second inverting amplifier, or the square wave generator. A 1 Ci PuBe neutron source was used as the counting source. The neutron fluence from this source was given to be

$1.75 \times 10^6 \text{ n/s}$. With the

detector at a distance of 10 cm, the neutron flux on the active volume is calculated to be 1393 n/s-cm^2 . Since the active volume cross section is 1.267 cm^2 , the rate at which neutrons are incident on the detector active volume is 1765 n/s .

The detector was filled to a simulated tissue volume of $1 \mu\text{m}$ ($P=58.75 \text{ torr}$), the simulated volume used for all measurements throughout the duration of the project. Using the previously described setup, with an amplifier gain of 100, one minute count times were taken, and the total counts measured with the MCA system used were recorded. The resulting count curve is shown in Figure 11. The plateau region was chosen to be in the region of 400 to 550 V, and an operating voltage of 550 V was used for the duration of the experiment.

Figure 11. Count Curve for LET-1/2 (Simulated Tissue Diameter = 1 micron).

VII.II. Detector Calibration

It is a slight disadvantage that the LET-1/2 cannot be calibrated using a convenient source of photons. Rather, a method of calibration using the previously mentioned neutron source is employed.^[14] It is assumed that recoil protons produced by the neutrons have a roughly constant LET as they traverse the detector active volume. Therefore, the maximum energy deposited by these protons is equal to the product of the maximum LET of the protons and the largest chord length of the detector, in this case, the diameter of the active volume. The linear stopping power of a material on an incident detector is given by:^[22]

$$S = \frac{dE}{dx} = \left(\frac{e^2}{4\pi\epsilon_0} \right)^2 \frac{4\pi z^2 N_o Z \rho}{m_o c^2 \beta^2 A} \left[\ln \left(\frac{2m_o c^2 \beta^2}{I} \right) - \ln(1 - \beta^2) - \beta^2 \right] \quad (41)$$

where z is the charge (in units of e) of the incident particle, N_o is Avagadro's number, Z is atomic number of the medium, ρ is the density of the medium, $m_o c^2$ is the rest energy of an electron, A is the atomic weight of the medium, β is equal to v/c where v is the particle speed, and I is the mean excitation energy of the medium (usually determined experimentally). For the case where $v \ll c$, the stopping power can be calculated by:

$$S = \left(\frac{e^2}{4\pi\epsilon_0} \right)^2 \frac{4\pi z^2 N_o Z \rho}{m_o c^2 \beta^2 A} \left[\ln \left(\frac{2m_o c^2 \beta^2}{I} \right) \right] \quad (42)$$

Since the stopping power varies with the particle energy, then the maximum stopping power occurs where:

$$\frac{dS}{dE} = 0 \Rightarrow \frac{dS}{d\beta} = 0 \quad (43)$$

Knowing this, the maximum value of S occurs at

$$\beta = \frac{e^1 I}{2m_o c^2} \quad (44)$$

so that the maximum stopping power is calculated as:

$$S_{\max} = \frac{8\pi z^2 N_o Z \rho}{e^1 I A} \left(\frac{e^2}{4\pi\epsilon_0} \right)^2 \quad (45)$$

Using this relation, and values of Z , A , ρ , A , and I corresponding to the tissue equivalent gas at the pressure used^[24], then the maximum LET of protons in the detector sphere filled with the T.E. gas at a pressure of 58.75 torr is calculated to be 108.35 keV. Since there is very little matter in the detector volume (i.e., the simulated volume is quite small), then this value of LET can be assume to be constant across the diameter of the detector. One can also make a corresponding statement about the LET of protons in tissue by converting the T.E. detector volume to an equivalent volume of tissue. In this case, since the equivalent tissue volume is calculated to be 1 μm , then one can say that the maximum LET of the recoil protons is 108.35 keV/ μm .

The neutron spectrum is then expected to have no counts (except for occasional heavy recoil nuclei) above this value. This "proton edge" presents a good calibration point for the entire spectroscopic system. A calibration spectrum taken with the detector is shown in Figure 12. The proton edge is quite apparent, as well as some heavy recoil counts. It is quite apparent that a larger contribution of counts occur at lower energies. This can be explained by considering the energy spectrum of the recoil protons and examining Equation 42. Regardless of the energy spectrum, the LET of the recoil protons reaches a peak at only one particular energy, after which is drops off for all other energies. That is, a fraction of incident protons have an LET equal to the maximum LET in the detector. This

effect is enhanced by the fact that the maximum energy can only occur when three things happen simultaneously. The proton must have a maximal LET; it must have this LET when it crosses the active volume of the detector, and it must cross the detector at its diameter.

Figure 12 is normalized such that the area under the spectrum is unity. As well it has been calibrated for units of specific energy in tissue using Equation 11.

This method works quite well for calibration of a typical microdosimetric spectroscopy system. However, calibration of the system with an integrator inserted becomes more difficult as it becomes necessary to consider the fact that in order for the integrator to produce a finite signal, it must integrate over a period of time. Due to the stochastic nature of the events in the detector, the event frequency (and hence, the period between adjacent events) is random. The integrator circuit must necessarily be triggered, meaning that it sums all signals occurring within the integrating time period. To calibrate the system correctly, it is necessary to read only one event at a time. However, this is not necessarily true with the integrator inserted in the system. The situation is complicated by the fact that

one does not necessarily know how many events are summed during the integration period. Two possible solutions are presented for this problem.

The first solution involves calculating the average fluence, or rate, of protons across the active volume of the detector and finding the average period of time that elapses between events. During this period, an average of one event takes place in the detector. The probability of x events taking place in the detector during this period is then calculated to be:

$$P(x) = \frac{e^{-1}}{x!} \quad (46)$$

By setting the integration period to the average time between proton passages, the probability that more than one event takes place can be calculated. If the integration period is set such that an average of one event takes place in the detector during this period, then the probability that more than one event takes place is:

$$P(x > 1) = \sum_{x=2} \frac{e^{-1}}{x!} = e^{-1} (e^1 - 2) \quad (47)$$

For a large number of events, then roughly 26.4% of all pulses from the integrator are the sum of more than one event. This is quite large and is expected to distort the neutron calibration spectrum beyond the point of being able to detect the proton edge easily. If the integration period is set lower than the average period between events in the detector, then one can say that the average number of events per integration period is equal to the integration period divided by the average period of time between individual events in the detector. By doing this, the probability of more than one event taking place during an integration period is lowered and is calculated by:

$$P(x > 1) = \sum_{x=2} \frac{\left(\frac{\tau}{T}\right)^x e^{-\left(\frac{\tau}{T}\right)}}{x!} = e^{-\left(\frac{\tau}{T}\right)} \left(e^{\left(\frac{\tau}{T}\right)} - \left[1 + \frac{\tau}{T} \right] \right) \quad (48)$$

Figure 13. Count Probability and Ratio as a Function of Fractional Integration Period.

where τ is the integration time period, and T is the average period between events. The probabilities of a single count, more than one count, and the probability ratio of a single count to multiple counts during an integration period is shown in Figure 13. As expected the probability of obtaining multiple counts during an integration period decreases as the size of the integration period decreases, as does the probability of obtaining a single event during the integration period. The ratio of the probability of obtaining a single event to that of obtaining multiple events increases as the integration period decreases, having no optimum point.

This method also requires one to know the rate at which recoil protons interact in the medium, as well, the probability of obtaining multiple pulses, although potentially small in cases of relatively low integration times, is still present, and the probability of counting one event is also quite small.

However, there is a more precise and simple way of calibrating the counting system with the integrator in place. Rather than use a source of random period between events, one is used which has a precise period of time between individual events - a pulse generator.

Initially, the neutron spectrum in Figure 12 was taken with an ordinary counting system (i.e., without the integrator or second amplifier). The pulse generator was used in the test input of the pre-amplifier to generate an artificial "event" of the same energy as the neutron edge. An EG&G Ortec 480 pulse generator was used (S.N. 2189). A pulse amplitude of 154 mV into the test input of the pre-amplifier was sufficient to create a spike in the neutron spectrum at the exact same channel as the proton edge. The same pulse was introduced into the test input of the pre-amplifier of the complete counting system shown in Figure 10. The integration time was set to be equal to the input pulse period of 16.6 ms. A single event during the integration period is assured. This raw calibration

Figure 14. Calibration Spectrum for LET-1/2 Signal Integration System.

spectrum is shown in Figure 14. The primary peak is the peak corresponding to a pulse from the pre-amplifier equal to a pulse from a proton of maximum LET crossing a diameter of the active volume of the detector.

The smaller, secondary peak is the noise spectrum from charge transfer and line noise. This spectrum was produced simply by turning the pulse generator off. Therefore, the only output was the integration of noise in the spectrum. This method has the advantage of presenting an average value of noise in the spectrum. The noise spectrum, expected to approximate a Gaussian distribution about an average value, is dealt with slightly differently than an ordinary background spectrum. Since the signal pulses output from the integrator are the sum of signal and noise, any pulse is expected to be offset by an average noise value. In addition, since this noise value is distributed about an average value, a peak in the spectrum will be the convolution of the noise spectrum and the original peak. This has the effect of making peaks wider, as is shown in the calibration peak in Figure 14. Since the integration period was quite long and the calibration pulse of low amplitude, then one expects a noise pulse to be similar in amplitude to the signal pulse. In the case of the integration spectrum, the calibration peak, taken to be at channel number 565, is actually offset by the average value of the noise, or 306 channels, so the actual value of the calibration peak is in channel number 259. This channel can be calibrated to an energy value of 108.35 keV, and a specific energy value of 32.56 Gy.

This calibration method has the advantage that the linearity of the system can be easily checked by adjusting the amplitude of the input pulse. Using the previous input pulse producing a peak at channel number 565, if its amplitude is reduced by a factor of 2, then the channel number that the peak is expected to occur in would be in the channel of the net peak divided by 2 and added to the noise offset of the system. Therefore, a pulse of one-half of the original amplitude is expected to be in channel number 436 with an error equal to the FWHM of the calibration peak. The actual peak fell quite close to channel number 436, well within the error value, so the MCA was taken to be linear with input energy. The linearity of the system becomes evident later as spectra are examined and discussed.

For a spectrum collected with an inverting amplifier gain of 10 and a primary amplifier gain of 250, the conversion factor from channel number to specific energy for a 32 bit MCA was 0.0128 Gy/channel.

VII.III. Microdosimetric Spectra

Before proceeding with measurements of integrated energy over time, ordinary spectra were taken with the counting system in Figure 10 without the integrator or inverting amplifier. These spectra are the results of single events in the detection system. Since events in the spectra are typically small and close to zero, the spectra are displayed on a plot of $z^2f(z)$ as a function of z with the horizontal scale being logarithmic. This has the affect of highlighting characteristics in the spectra as well as making the horizontal scale appear wider. In typical microdosimetric applications, the standard form of spectra is $y^2f(y)$ as a function of y on the same type of plot. However, since the concept of lineal energy becomes ambiguous for more than one event, specific energy was used

as the quantity of choice since it can be treated as a cumulative dose. The use of specific energy instead of lineal energy should have little effect on the spectral shape as both are related only by a constant:^[14]

$$z = \frac{y}{\pi r^2 \rho} \quad (49)$$

where r is the radius of the simulated spherical volume, and ρ is the density (typically taken to be 1 g/cm³). The relationship between a "typical" energy spectrum and the "standard" microdosimetric distribution can be explained by examining the frequency distributions of energies, and their moments. At a particular specific energy, the fractional contribution to dose $d(z)$ from that energy is given to be the product of the specific energy and the frequency of that deposition. This is known as the first moment of the spectrum, and applies to many probability distributions. If the frequency distribution is normalized, then the product is the fraction of events having that specific energy. The frequency average dose is given by the sum of products of all events and their frequencies:^[14]

$$z_F = \int_0^{\infty} z f(z) dz \quad (50)$$

Equation 50 can be used to determine the fraction of events with specific energies between two specific energies by changing the limits on the integral.

The dose average distribution is defined as the contribution to total *dose* from events of particular specific energies. This is equal to the second moment of the distribution divided by the first moment. In the case of a single discrete specific energy, this dose average is equal to the frequency average. For all specific energies, the dose average is defined by:^[14]

$$z_D = \frac{1}{z_F} \int_0^{\infty} z^2 f(z) dz \quad (51)$$

The physical significance of this is that z_D is defined to be the total contribution to the fraction of dose from two specific energies, while z_F is the contribution to the fraction of events from two specific energies. Again, by changing the limits of integration in Equation 51, the contribution to total dose from two specific energies can be determined.

These concepts can be applied to microdosimetric spectra and adjusting the integrals to fit the spectra. As an example, the ¹³⁷Cs spectrum taken with the counter is examined. Again, the counter was filled with the previously mentioned T.E. gas to a pressure simulating 1 μ m. The normalized spectrum was taken with an amplifier gain of 250, and the calibration used was that gained from the neutron calibration spectrum in Figure 12. The frequency spectrum is shown in Figure 15. Background for this spectrum was negligible. Details in this representation are very difficult to discern. Figure 16 is the same spectrum, but the vertical axis is of $zf(z)$, while the horizontal is on a logarithmic scale. This plot of frequency average as a function of specific energy can be explained by the following equation:^[14]

$$\int_{z_1}^{z_2} f(z) dz = \int_{z_1}^{z_2} [zf(z)] d\log(z) \quad (52)$$

The physical explanation is that the contribution to the fraction of events delineated by two specific energies is equal to the integral under the curve in Figure 16 with respect to the logarithm of the specific energy. Details of the spectrum in Figure 16 are more prominent as the horizontal scale is no longer linear.

In addition, one can also discern total contribution to the dose deposited in a similar manner. Since the total dose deposited by events of specific energy z is equal to $zf(z)$, then a plot of $z^2f(z)$, as in Figure 17, would also yield important information. This spectrum will allow one to determine the total contribution to dose from events between two specific energies and can be explained with an equation similar to Equation 52:

$$d(z) = \frac{zf(z)}{z_F} \quad (53)$$

so that the first moment of this distribution is the second moment of the frequency distribution:

$$zd(z) = \frac{z^2f(z)}{z_F} \quad (54)$$

and the contribution to dose between two specific energies can be given as:

$$\int_{z_1}^{z_2} d(z)dz = \int_{z_1}^{z_2} zd(z)d\log(z) = \frac{1}{z_F} \int_{z_1}^{z_2} z^2f(z)d\log(z) \quad (55)$$

so the dose from events between two specific energies is proportional to the integral under the curve in Figure 17 with respect to $\log(z)$. An examination of the spectrum in Figure 17 allows one to determine that the largest contribution to dose is from events of lower specific energy (less than one Gy).

Spectra of frequency distributions ($f(z)$ vs. z), dose distributions ($zf(z)$ vs. z), and distributions of the first moment of the dose distribution ($z^2f(z)$ vs. z) for x rays and the $^{90}\text{Sr}/^{90}\text{Y}$ β source are displayed in Appendix D. The explanations of these spectra follow the same logic as that of the ^{137}Cs spectra.

VIII. Procedures

The actual collection of data was virtually a self-sustaining process, requiring little human intervention. Using the experimental setup described in Figure 10 with the equipment outlined in Section VI.VI, data was collected for various integration times.

The procedures governing the data collection for various integration times were identical. The integration times for which data was collected were 1, 10, 50, 100, and 500 μ s as well as 1, 2, and 5 ms. For each collection sequence corresponding to a specific integration time, the integration time was recorded, as well as the duty cycle corresponding to that particular integration time. The count time was also recorded. In the case of the ^{137}Cs spectra, the count time was 60 s while that of the x-ray and β spectra were 180 s. In order to set the integration time properly, an oscilloscope was connected to the *hold* connection on the integration circuit which was powered by the NIM bin. In order to avoid saturation of the solid state switches in the integration circuit, power was turned on before the triggering signal was connected. After powering up the circuit, the square wave trigger signal was provided to the *trigger* input of the circuit and was adjusted until the width of the *hold* signal was equal to the desired integration period. It was not necessary to provide an input signal to the integration circuit during the reading of the *hold* signal as this is not affected by an input signal, nor was it necessary to provide power to the detector or the pre-amp.

The amplifier signal was then connected in series with the input resistor to the integration circuit. The input resistance of the integration circuit was provided externally with a BNC connection and alligator clips. Providing the input resistance externally had the advantage that it could easily be varied to compensate for a large input signal if necessary. However, it had the disadvantage of being easily overlooked, meaning that the only input resistance to the integrator circuit was provided by the integrator chip itself. This input value of $1.5\text{k}\Omega$ would have meant that the input current would have been much larger, and the chip would have approached saturation earlier than desired.

The integrator chip, being very sensitive to minute changes in charge across the switches was encased in a steel box with a wall thickness of roughly 0.32 cm. For irradiations using ^{137}Cs and x rays, the box was also shielded with lead bricks.

Having set the integration cycle, the detector power was turned on, and spectra were collected. For all spectra collected, the gain on the first amplifier was held at 250. For most spectra, the gain on the inverting amplifier was held at 10. In cases where dose rate was large, the MCA recorded pulses in overflow channels, and the gain was reduced. In some cases, the gain was increased due to the small amount of charge collected during short integration times. Since the system was calibrated to a gain of 10, all spectra collected were calibrated to a gain of 10 before conversion to energy values. For all integration times, background spectra were taken. These background spectra were used to determine the offset in the signal spectra using the method outlined in Section VII.II. All energy spectra were placed into an MCA spectrum of 32 bit resolution (1024 channels). Table 4 gives a list of spectra collected, integration times, and the gain of the inverting amplifier.

Analysis of the spectra was done by first converting the spectral data to a text tabular format using the *chconv* routine in the *Maestro* software package. The file was then analyzed using *Lotus 1-2-3* (release 5). Any positive offsets in the spectra were first subtracted. The horizontal axes of the spectra were then converted to specific energy in units of Gy by first multiplying by the quotient of 10 and the gain of the inverting amplifier (to obtain spectra all normalized to the same energy values) and then multiplying the resulting values on the horizontal axis by the calibration factor (such that channel 259 becomes 32.56 Gy; see Section VII.II). Spectra were then converted to the standard form of $z^2f(z)$ as a function of $\log(z)$. In addition, spectra were converted to the dose distribution $zf(z)$ as a function of $\log(z)$ for comparison. The frequency average dose was also calculated using Equation 50 and plotted as a function of integration time.

It is expected that the average dose deposited in the detector increases linearly with integration time. This should appear on the plot of frequency average dose as a function of integration time. As well, the slope of this plot should give some indication of the dose rate from the irradiation source.

For larger integration times, the signal noise offset peak did not appear in the signal spectrum. This is due to the fact that the output of the integrator was a single pulse whose amplitude corresponded to the sum of the signal pulse and any noise occurring during the integration period. Therefore, the spectrum would appear without the signal pulse, because all events would include at least one signal pulse. However, as the integration period decreases, the probability of an event occurring during the integration period decreases, and many output pulses from the integrator are only a sum of noise pulses, creating a low energy contribution to the spectrum. In the case where the signal spectrum overlapped the noise spectrum, the background peak in the noise spectrum was also evident as a peak in the signal spectrum. This was easily identified, as one expects the signal spectrum to be a distribution of energies about an average value.

Table 4. Integration Times and Amplifier Gain for Various Irradiation Sources

¹³⁷ Cs		⁹⁰ Sr/ ⁹⁰ Y		X Ray	
t_{int}	Gain	t_{int}	Gain	t_{int}	Gain
1 μ s	25	1 μ s	25	1 μ s	10
10 μ s	25	10 μ s	25	10 μ s	10
50 μ s	25	50 μ s	25	50 μ s	10
100 μ s	25	100 μ s	25	100 μ s	10
500 μ s	10	500 μ s	10	500 μ s	10
1 ms	10	1 ms	10	1 ms	10
2 ms	10	2 ms	10	2 ms	10
5 ms	10	5 ms	5	5 ms	5

IX. Results

Data are presented separately for each irradiation source. These data consist of frequency distributions of specific energy, dose distributions, and the first moment of dose distributions. As well, average dose is plotted as a function of integration time.

IX.I. ^{137}Cs Spectra

For ^{137}Cs , the background noise proved to be negligible in many cases. The only spectrum for which there was an offset in the peak value due to noise was the spectrum for the 5 ms integration time. This offset was 24 channels. For the remainder of the spectra, the offset was less than one channel, although some low energy noise pulses were present in the background noise spectra. Spectra of $z^2f(z)$ as a function of z are presented in Figure 18 for integrations times from .5 ms to 5 ms. As expected, the dose distribution forms a Gaussian curve about an average value, and this value increases with integration time. Figure 19 shows the same spectra for integration times below 500 μs . These spectra are quite compacted, even on a logarithmic scale. However, the average dose contribution continues to follow the same relationship as for the spectra collected at higher integration time. One interesting spectral characteristic that appears for spectra collected at lower integration times that does not appear in the spectra collected at higher integration times is the low energy spike. This is quite apparent in the spectrum collected for an integration time of 10 μs (0.01 ms). This spectrum is displayed in Figure 20. This spectrum does resemble the typical Gaussian distribution that is expected. However, at the lower energy section of the distribution (dotted line drawn in), rather than the typical "tail," an increase in the dose from lower energy events is evident.

This is not believed to be due to actual events, thus no dose from these low energy events is expected. Rather, this increased number of events is hypothesized to be due to an artifact of the integration. The integrator used does not discriminate between noise and signal due to actual events in the detector. Because this is so, noise is integrated

into this spectrum also, creating an offset at large integration times. At lower integration times, the probability of a single event taking place decreases as is expected to be calculated by a Poisson distribution. However, noise is always present in the system, so this is integrated into the signal. So, for the 10 μ s integration time displayed in Figure 20, the probability of a single event occurring in the detector is low, as is shown in Figure 13, and for each integration period an actual signal event may not necessarily be collected by the counting system.

However, at integration times lower than 10 μ s, the noise spike is not nearly as pronounced. This can be further explained by the setting of the detection limit of the MCA. While the probability of a single event occurring in the detector during the integration period continues to decrease quite rapidly with integration time, and while noise is still expected to exist, the lower integration time will mean that less noise is integrated into the signal, so the noise contribution is lower than the detection limit of the MCA.

Spectra of $f(z)$ and $zf(z)$ as a function of z for ^{137}Cs are shown in Appendix D. These also follow the Gaussian relationship of the spectra shown in Figures 19, and 20.

Of particular interest is the frequency average dose as a function of integration time. As stated previously, this is expected to be linear with integration time. Using a commercially available plotting program, the frequency average dose (calculated using Equation 50) was plotted against the integration time. For ^{137}Cs , this is displayed in Figure 21. It is seen that the linear fit is fairly good. In fact, the R^2 value for this fit was calculated using the plotting program to be 0.989. The frequency average dose for the 2 ms count time is slightly higher than expected, but well within the absolute variance of the dose spectrum for this integration time. This error is most likely due to a slightly larger amount of statistical noise that was not detected while integrating the background noise yet was integrated into the system during the integration time. However, the remainder of the doses seem to fit quite well.

The characteristics of this line present some useful information. The slope of the line is representative of the frequency average dose rate

from the source into the active volume. It is also noted that a positive y-intercept exists. This could very well represent a background dose, however, this would be quite a high background, and one expects the y-intercept to be zero, corresponding to zero dose at zero integration time. The presence of this offset can be due to several things.

One explanation comes from the fact the detection limit of

the MCA is finite but not zero. Because of this, zero energy and low energy events would not be averaged into the total frequency average dose, giving it a finite offset. This would seem quite possible, as the MCA has a manually adjustable LLD, which is typically set at around 10 channels in order to discriminate against low energy noise in typical applications. However, this explanation suffers from the fact that low energy events would be less frequent, if they even exist at higher integration times, creating a higher offset at lower integration times and a lower offset at higher integration times, giving the curve a more logarithmic appearance. In addition, the LLD of the MCA was purposely set as close to zero as possible for this experiment with the knowledge that signal events would be low energy events. However, a finite LLD would raise the average dose of lower energy events, creating an offset proportional to the dose rate.

The offset could also be due to charge transfer from the solid state switches in the system. Because the calibration was done with a large integration time of 16.6 ms, the offset due to actual charge transfer (not noise) would have been negligible. However, at integration times less than this, such as are used, the offset due to charge transfer would be more prominent. This offset would be constant regardless of dose rate. The sensitivity of the system is quite apparent here. Any offset is most likely due to a combination of the above two explanations.

The variance in the spectra are also interesting to note, and this will prove to be a fascinating point in consideration of x rays and β particles incident on the detector. Using the *Maestro* software package, the variance σ was taken to be the half-width at half-maximum (HWHM) of the frequency distribution. The relative variance, V , was then calculated using Equation 15. For the ^{137}Cs source, the relative variance is tabulated with respect to integration time in Table 5.

Table 5. Variance and Relative Variance of Dose Distributions vs. Integration Times for

t_{int}	$^{137}\text{Cs}_s$	
	σ (Gy)	V
5 ms	0.401	0.059
2 ms	0.513	0.183
1 ms	0.319	0.138
0.5 ms	0.013	0
100 μs	0.058	0.015
50 μs	0.047	0.011
10 μs	0.001	0.001
1 μs	0.034	0.001

Perhaps the most interesting point to make concerning the variance of the ^{137}Cs measurements is the small variance of in the spectrum corresponding to the 10 μs integration time. Remembering that this spectrum has a large contribution of dose close to zero in the form of a "spike," one can say that the HWHM of this peak is quite small, even though it is not an actual signal peak, this would result in a small relative variance. Although no real

trend can be noted at this point, the small relative variance for the spectrum corresponding to an integration time of 10 μ s is rather interesting; as well, relative variances seem to decrease with decreasing integration time.

IX.II. X-Ray Spectra

As with ^{137}Cs , x-ray spectra were also taken for the previously mentioned integration times. While the ^{137}Cs spectra were expected to be narrower (have a lower variance), those of the x-rays were expected to have a higher variance due to the wide spectrum of energies incident on the detector. Although a wide range of effects will influence the spectra, particularly in the microdosimetric domain, a collected spectrum is expected to be a convolution of the actual energy spectrum and the effects that a detection system will have on a monoenergetic source. Since the x-ray source emitted a continuous spectrum of energies, then the relative variance in the spectra were expected to be larger.

The standard dose spectra for integration times of 500 μ s and higher are shown in Figure 22. Again, these spectra follow a Gaussian distribution, with average dose contribution increasing with integration time. Quite noticeable is the extremely high peak from the spectrum obtained with an integration time of 5 ms. Since the peak height increases by a factor of z^2 , one certainly expects the 5 ms peak to be the highest. However, other contributions to this height include the lack of lower energy events in this peak. As the integration time increases, the average number of events during an integration period increases. Obviously, the reverse is true as the integration time decreases. With fewer events, the signal pulses are of lower amplitude and a larger number of

Figure 22. X-Ray Dose Spectra for Various Integration Times.

Figure 23. X-Ray Dose Spectra for Integration Times Less than 0.5 ms.

lower energy events are averaged into the dose average. If many events are clustered about zero, on a plot which increases as z^2 , the zero energy events will not be as visible.

The dose spectra for lower integration times are displayed in Figure 23. One notes that as the integration time decreases, the average dose contribution also decreases. Although the low energy contribution is still present, these spectra seem to have a shape more closely resembling a Gaussian distribution than those of the ^{137}Cs source. This is an important consideration in that the noise contribution is minimal in these spectra. Closer examination will reveal that the major dose contribution at any particular integration time is higher from the x-ray source than from the ^{137}Cs source. The flux of photons from the x-ray source is also higher than from the ^{137}Cs source, so that during an integration period, the probability of a single event taking place in the detector is higher if the detector is exposed to the x-ray source than if it is exposed to the ^{137}Cs source, therefore, the noise makes a lower contribution to the total signal.

The dose rate from the x-ray source as compared to the ^{137}Cs source can best be compared by examination of a plot of the frequency average dose as a function of integration time. This plot is shown in Figure 24. As before, there is a slight offset due to previously mentioned factors. The plot is also very linear, with an R^2 fit of 0.994. As well, the slope of this plot is greater than that of the same plot corresponding to the ^{137}Cs source. For this type of measurement, comparison of slopes would be a not only a good indicator of dose rate, but also a tool for comparison of various sources.

Table 6 gives a listing of absolute and relative variances for the frequency average doses from the x-ray source. As expected, the relative variance is larger in the x-ray source spectra than in those of the ^{137}Cs source. This

would most likely be attributed to the broad spectrum of x-ray energies incident on the detector causing a broader range of secondary electron energies.

Table 6. Variance and Relative Variance of Dose Distributions vs. Integration Times for 250 kVp X Rays.

t_{int}	σ (Gy)	$\bar{V}$
5 ms	2.717	0.891
2 ms	1.817	1.009
1 ms	1.105	0.519
.5 ms	0.186	0.021
100 μ s	0.056	0.003
50 μ s	0.04	0.002
10 μ s	0.018	0.004
1 μ s	0.012	0.002

For x rays, the absolute variance increases with integration time. This would be expected as a broader range of integrated noise is possible, and the width of the noise spectrum would increase with time. As well, the relative variance seems to increase to an asymptotic level. Typically, the absolute variance is expected to be roughly constant (accounting for the detection efficiency of the system with respect to

energy), however, since the low energy spectra are not complete Gaussian distributions (i.e., there is no low energy tail because most events are clustered around zero), the HWHM of the distribution is distorted, giving a value lower than expected, and, hence, a lower variance. The relative and absolute variances are plotted in Figure 25.

Spectra of $zf(z)$ and $f(z)$ as a function of z are also plotted in Appendix D.

IX.III. $^{90}\text{Sr}/^{90}\text{Y}$ Spectra

The spectra from the $^{90}\text{Sr}/^{90}\text{Y}$ source proved to be rather interesting, if not slightly anomalous. Because the aluminum housing and the TE plastic shell surrounding the active volume are roughly equal to the maximum range

of ^{90}Sr electrons, the only dose expected is from ^{90}Y . However, in many spectra, there appeared to be a secondary contribution to dose from lower energies. That is, the dose spectra seemed bimodal. This contribution does not seem to match the lower energy noise spikes present in the x-ray and ^{137}Cs spectra. Rather, they had a more profound effect at higher integration times and actually seemed non-existent at lower integration times. In addition, the width of these peaks were greater

than the width of the noise spikes observed in the x-ray and ^{137}Cs spectra. This would indicate a dual contribution. For most spectra, the energy deposition was entirely above or below this energy level; however a few spectra had energy depositions straddling this level, showing a dip in dose contributions.

For integration times above 100 μ s, the dose spectra are shown in Figure 26. The typical Gaussian distribution is noted. For the 5 ms integration time, the distribution resembles other distributions discussed thus far. However, for

integration times of 2, 1 and .5 ms, the bimodal characteristics are noted. This is especially noticeable in the spectra for the 1 and 2 ms integration times. Only the 2 ms spectrum is displayed in Figure 27. This phenomenon is explained by Rossi and Zaider as being produced by the activities of not only incident electrons, but also secondary delta rays, which contribute a different LET to the chamber as the primary electrons.^[14] This would be present only at certain integration times because at those time intervals, the probability of only one event interacting in the detector is maximized, and the spectrum would closely resemble a typical microdosimetric spectrum. For higher integration times, the probability of more than one particle interacting becomes greater, and the spectrum resembles the typical Gaussian that is observed. At lower integration times, partial energies would be collected, and the incident particles would no longer deposit their total energy. In addition, because of the large amount of scattering of electrons in a material, there is a higher probability that the electron will deposit all of its energy in the detector.

Spectra for low charge collection times are displayed in Figure 28. As in previous cases, the major contribution to dose decreases with decreasing integration time. Also, a slight biphasic characteristic is noticeable as a slight dose contribution can be seen at doses of roughly 6 Gy. While the probability of an electron depositing all of its energy in scatter events during these short integration times is small, it is seen that a few integration periods do exist in which a larger dose is deposited from these electrons.

The frequency average dose contribution is also linear with integration time for electrons incident on the detector. This plot is shown in Figure 29. The linear fit was quite good with an R^2 value of 0.987. While the dose rate was slightly higher than that of the ^{137}Cs source, it was not as high as the dose rate from the x-ray source. As in Figures 21 and 24, a slight offset exists in the doses.

The absolute and relative variances of the frequency average dose for various integration times are listed in Table 7. While most of the relative variances are

quite low, those of the higher integration times seem slightly elevated, with a local maximum around 1 and 2 ms. This local maximum is to be expected as the width of the spectral peak is actually taken to be the total width of the biphasic peak and the secondary delta ray peak.

Dose distributions for the $^{90}\text{Sr}/^{90}\text{Y}$ source as $zf(z)$ and $f(z)$ given in Appendix D with the same distributions for ^{137}Cs and the x-ray source.

Table 7. Variance and Relative Variance of Dose Distributions vs. Integration Times for

$^{90}\text{Sr}/^{90}\text{Y}$ β Particles.

t_{int}	σ (Gy)	V
5 ms	2.773	1.782
2 ms	0.372	0.07
1 ms	0.517	0.195
.5 ms	0.006	0
100 μs	0.003	0
50 μs	0.04	0.009
10 μs	0.012	0.001
1 μs	0.011	0.001

IX.IV. Dose Relationships.

Given the system calibration and the above linear dose deposition relationships, the following equalities have been determined for total average dose deposited in a simulated $1\mu\text{m}$ spherical volume by the above sources:

$$D_{\gamma}(\Delta t) = 534\Delta t + 0.228 \quad (56)$$

$$D_{\beta}(\Delta t) = 841\Delta t + 0.221 \quad (57)$$

$$D_x(\Delta t) = 1480\Delta t + 0.738 \quad (58)$$

where D_{γ} , D_{β} , and D_x are the doses from gamma rays, β particles, and x rays in Gy respectively, and Δt is the integration time in seconds. The slope of these lines are significant in that they are representative of a total dose deposited in the simulated volume over a time Δt , the dose rate. However, a quick calculation to estimate expected dose rates from the sources used indicates that these slopes do not match the calculated dose rates. The explanation of this is given by Rossi and Zaider.^[14] In microdosimetric measurements, the quantity z_F is dependent on the size of the reference volume. Since the change in the number of single-hit events in small volumes changes little with size, one can imagine a sphere of tissue representing the tissue volume studied. As the tissue sphere increases in size, the amount of energy deposited from an event would increase linearly as the chord length of the detector increases. However, the simulated mass of the sphere, to which the specific energy is inversely proportional, increases according to the increased volume of the sphere, or cubically. Therefore, as simulated tissue volumes decrease (for small volumes), the specific energy increases for an event. This is important as the experimenter can measure localized volumes of energy with great resolution. This idea has connotations of a weighting factor for individual cells.

A plot of localized dose as a function of the calculated dose in tissue (one can imagine a tissue-equivalent sphere the size of a human) is shown in Figure 30. In addition, actual estimated dose rates for the sources used are

shown. The dose rate for the $^{90}\text{Sr}/^{90}\text{Y}$ source is calculated for Y only. No particular relation can be seen that relates the three sources, however, this relation is not to be expected. At the microscopic level, one expects difference in the lineal energy, and hence, the total energy deposited in the tissue sphere. Therefore, a particular dose from one source would not be related to the

Figure 30. Calculated Tissue Dose Rates vs. Measure (Microscopic) Dose Rates.

same macroscopic dose from another on a microscopic scale. However, given two different whole body doses from the same source, one might expect that the relationship to microscopic dose is linear (with no y-intercept) as the only governing variable is the particle flux through the bodies of interest. Since microscopic and macroscopic dose are both directly proportional to the incident particle flux, then the relationship is expected to be linear.

From Figure 30, one could induce that although the calculated tissue dose from the x-ray source is lower than that from the ^{137}Cs source, the specific energy (microscopic dose) is higher due to the increase in lineal energy from the lower energy photons.^[14] Similar comparisons can be drawn between the beta source and the photon source also. Calculations of tissue dose are given in Appendix E.

X. Discussion

Of primary consideration in radiation dose measurements is the biological effect of the measured dose. In normal cells, a dose deposited will have a specific endpoint. However, the dynamics of the repair process negate this effect. In considering cell damage and repair, two important considerations are the dose deposited and the time necessary to neutralize an effect.

Radiation deposition in a cell can result in indirect damage to the cell. This damage is initiated by the production of free radicals in the cell, primarily the $\cdot\text{OH}$ radical. Radical sites will then bind and damage DNA through strand breakage. Fast chemical repair is defined as a cell's ability to neutralize these radical sites before DNA damage occurs. Rates of chemical repair can be as high as 3000 repairs per second.^[3] Most repair rates are taken as an exponential constant such that the number of effects remaining at time t after an initial dose is expected

to follow the relationship:^[25]

$$N(t) = N_0 e^{-kt} \quad (59)$$

where k is referred to as the first order repair rate. As well, many studies relate the production of single strand break (SSB) and double strand break (DSB) precursor production to follow a quadratic relationship:^[4]

$$\epsilon(D) = \alpha D + \beta D^2 \quad (60)$$

where ϵ is defined as the average number of lesions produced by a given dose. This yields the familiar survival curve:

$$S(D) = e^{-\epsilon(D)} = e^{-\alpha D - \beta D^2} \quad (61)$$

The quantities α and β are empirically derived values varying with cell and irradiation type. It is noted that at lower doses, the quadratic relationship of ϵ to dose is dominated by the linear portion, and vice-versa for higher doses. Some studies have shown that ϵ is linear with dose for DSB precursor production.^[25]

Given Equations 59 and 60, for a continuous, constant low dose rate, the rate of precursor production could be primitively said to follow a relationship equal to the rate of production minus the rate of neutralization:

$$\frac{dN}{dt} = \frac{dN}{dD} \frac{dD}{dt} = \dot{N}_{\text{prod}} - \dot{N}_{\text{neut}} = \frac{d\epsilon}{dD} \cdot \frac{dD}{dt} - \frac{dN(t)}{dt} = \alpha + 2\beta t - kN \quad (62)$$

This is simply the difference in the derivatives of Equations 59 and 60 at lower integration times. It is the production and repair of radical sites that is of interest. Using the above data, a "critical dose" and "critical time" are being introduced and defined for production and neutralization of radical sites. The "critical dose" is defined here as the dose necessary to produce an effect. The "critical time" is the time necessary to neutralize the effect caused by the critical dose. If a dose to an organism exceeds the critical dose during the critical time, then it can be said that the organism is incapable of neutralizing the effect before another effect takes place, thus placing the organism in danger of the consequences of the produced effect. Two examples from research follow for which a critical dose and time are defined. These are compared to the dose depositions from the irradiation sources used for this experiment.

X.I. V79 Cells

Michael, Davies, and Held report a D_{37} dose for SSB precursors in V79 cells to be approximately 25 Gy (electrons).^[7] Mathematically, the D_{37} dose is the dose at which each cell has an average of one induced effect. This becomes a useful parameter, as it represents a population average dose for an average of one effect per organism. The study presents an exponential decay model for the neutralization of SSB and DSB precursors. The rate of fast SSB neutralization is reported to be 350 s^{-1} , or a repair time of approximately 2.9 ms per site. The rate of DSB site neutralization is listed to be 300 s^{-1} , or a repair time of 3.3 ms per site. These rates represent decay constants of radical sites, thus following the relationship of Equation 59. Examining the lines given in Figures 21, 24, and 29, it

is possible to make comparisons of dose and time against the plots using either a critical time or a critical dose. For the cells presented in this study, the critical dose is set to 25 Gy, the dose necessary to produce an average of one effect per cell. The critical time is slightly more difficult to determine since repair is not linear. However, critical time is determined to be the half-time (0.693 divided by the decay constant) of SSB and DSB repair in the cell. The half-time is taken because it represents the amount of time necessary such that the probability of repair of a single site is greater than 50%. For SSB sites, this time is 2.0 ms, and for DSB sites, the half-time is 2.3 ms.

The comparison of critical times and dose for V79 cells is shown graphically in Figure 31. One can say that for a critical time of 2 ms, the dose deposited by the source used in this work for a 1 μm site is approximately 1.3 Gy. If the dose deposited during the critical time exceeds the critical dose, then repair rate of the cell is not high enough to compensate for damage caused by the source. Alternatively, one

Figure 31. Examination of Cesium-137 Dose Using Critical Time and Critical Dose

can say that at a critical dose of 1.3 Gy (as shown in Figure 31), the critical time is 2 ms. If the dose necessary to cause an effect is deposited during a time period less than the repair time of the lesion, then the damage caused by the source will exceed the neutralization by the cell. It is at this point that the activity of the source can prove quite dangerous, and the possibility of actual SSB and DSB production increases. It is seen for the ^{137}Cs source that the dose deposition during a critical time period does not exceed the critical dose of 25 Gy. One can also use Equations 56-58 to determine the dose deposited by the source in a 1 μm site, since these equations were determined graphically. The dose from ^{137}Cs incident on the detector during the critical time of 2 ms is calculated to be 1.30 Gy, matching the value determined using Figure 30. For the x-ray source in the configuration used by the experiment, the dose deposited during a critical time period of 2 ms is calculated to be 3.7 Gy. For the $^{90}\text{Sr}/^{90}\text{Y}$ source, the dose to the detection system is calculated to be 1.9 Gy. In all three cases, the calculated dose is not more than the critical dose, so the possibility of SSB production by the indirect effect is minimized.

X.II. pBR322 Plasmid DNA

In two studies of isolated pBR322 plasmid DNA (purified from *E. coli*), scientists at the Gray Cancer Research Laboratory have established chemical repair rates as well as dose effects.^[3,25] In one study, the first order repair rate of SSB precursor repair was determined to be 1370 s^{-1} ($t_{1/2}=505 \text{ } \mu\text{s}$), while the rate for DSB precursor repair was determined to be 2900 s^{-1} ($t_{1/2}=240 \text{ } \mu\text{s}$).^[25] DNA used in this study was irradiated with 400 keV electrons. This study determined DSB precursor production to be linear with dose, while SSB precursor production followed the typical relationship given in Equation 60. The production rates given for SSB's and DSB's under oxic (normal) and hypoxic conditions are outlined in Table 8. Since *E. coli* has $8 \times 10^{-15} \text{ g}$ of DNA,^[26] or $4.818 \times 10^{10} \text{ Da}$ (one Dalton being equal to $1.661 \times 10^{-27} \text{ kg}$), then the production rate can be easily extrapolated to the entire cell. These values are also shown in Table 8.

Table 8. SSB and DSB Precursor Production Rate for pBR322 Plasmid DNA.^[25]

	SSBs		DSBs	
	<u>Oxic</u>	<u>Hypoxic</u>	<u>Oxic</u>	<u>Hypoxic</u>
Production Rate (breaks/Gy/Da)	8.4×10^{-10}	9.2×10^{-10}	1.6×10^{-11}	8.7×10^{-12}
<i>E.coli</i> Production Rate (breaks/Gy)	40	44	1	<1

Given the production rates in Table 8, the critical dose for SSB precursor production in *E. coli* cells is set at $337 \text{ } \mu\text{Gy}$ (midway between the calculated critical dose under oxic and hypoxic conditions). For DSB production, the critical dose is set at 24 mGy . The critical time for SSB production is set to be $505 \text{ } \mu\text{s}$, and the critical time for DSB precursor production is $240 \text{ } \mu\text{s}$. Using Equations 56-58, then at the critical time for SSB precursor production, the dose deposited by ^{137}Cs is calculated to be 0.50 Gy . The x-ray source would deposit 1.49 Gy , and the $^{90}\text{Sr}/^{90}\text{Y}$ source deposits 0.65 Gy . It is seen that these sources can be classified as potentially dangerous to the organism based on the above data, as the strand break precursors are not neutralized before enough dose is deposited to cause another precursor. In this case, strand break production would be inevitable.

In another study by the same people, irradiations of pBR322 plasmid DNA were done with protons, and deuterons instead of electrons. The resulting chemical repair rate was much lower - on the order of 700 s^{-1} , resulting in a much longer critical time of about 1 ms .^[3] This lower rate of repair could be due to the interactions of radicals in close proximity due to increased clustering around a single site. The clustering would be a result of the action of the more densely ionizing radiation.

X.III. Conversion to Equivalent Lineal Energy

The above examples provide a general concept of the type of analysis conducted in consideration of energy deposition in cellular volumes and the endpoint of interest. Another type of analysis that will be discussed briefly is the comparison of dose deposition and lineal energy. Although the concept of lineal energy only applies strictly to single events, comparisons can be drawn between total energy deposition in a microscopic volume by a source of

one particular lineal energy and a source of another. An application would be the determination of the amount of time necessary for x rays to deposit an amount of energy in a microscopic volume such that the energy would be equal to that deposited by a more densely ionizing alpha particle. Using Equation 49, an "equivalent" lineal energy can be calculated. An "equivalent" lineal energy spectrum for 250 kVp x rays integrated over 2 ms is shown in Figure 32. This spectrum is identical to the

spectrum of $z^2 f(z)$ vs. z with the exception of the axes. While individual events are not represented in this spectrum, the sum of energies deposited over 2 ms is treated as a single event. The largest contribution to dose is from "lineal energies" between 10 and 30 keV/ μm . The energy deposited by 250 kVp x rays in 2 ms would be equivalent to that deposited by a particle that deposits its largest contribution to dose with lineal energies between 10 and 30 keV/ μm .

This treatment can also be extended to the frequency averaged equivalent lineal energy. By adjusting Figures 21, 24, and 29, the concept of equivalent lineal energy can be seen. One can thus determine the time of exposure necessary in order to obtain a particular amount of energy deposition equal to a single particle of higher lineal energy. A plot of equivalent lineal energies (calculated using Equation 5) as a function of integration times is shown in Figure 33. As an example, given a low energy electron with a lineal energy in tissue of 10 keV/ μ m, one can say that the ^{137}Cs source used in this experiment, must deposit energy for a time of roughly 4 ms to deposit energy in a 1 μ m sphere equal to that from one passage of the electron. The β source must deposit energy for a time of roughly 2.5 ms, and the x-ray source must deposit energy for a time of roughly 1 ms. This depends heavily on the activity of the source and the geometry of the detection system.

As with Equations 56-58, relationships can be drawn between equivalent lineal energy and deposition time for all three sources. These relations are as follows:

$$y'_{cs}(\Delta t) = 2520\Delta t + .95 \quad (63)$$

$$y'_x(\Delta t) = 7270\Delta t + 3.62 \quad (64)$$

$$y'_\beta = 4120\Delta t + 1.08 \quad (65)$$

where y' is in keV/ μ m, and Δt is in seconds.

XI. Conclusions

It is interesting to note that the bulk of this project was involved in production of the instrumentation. The final product, although not the intended device, performed the intended function quite well. The counting system described represents new (yet specific in function) instrumentation in the field. Thus, the first objective of the experiment was met.

Although it did not pose a serious problem, the detector calibration proved to be tedious. The sensitivity of the system to small changes in charge and voltages made small changes in input pulse voltage difficult to accomplish. As well, a more direct calibration technique is desired. While a different calibration technique has not been proposed, it is hoped that one will be developed by those exploring this project in the future.

Perhaps the most important part of the project was the collection of the actual data. As a preliminary experiment, this project yielded results consistent with expectations. As expected, the dose deposition as a function of integration time was linear. In addition, the concept of a threshold dose is strongly evident in the nature of this project. By defining a critical time and a critical dose, this work suggests that any dose less than a critical dose or any integrated deposition time greater than the critical time will have little effect on the organism. However, the stochastic nature of radiation interactions in cells (stochastic because it is not fully understood), limits this concept. It is believed that the critical parameters defined herein are closely related with the parameters α and β given in Equations 60 and 61, since these parameters represent an exponential increase in endpoint probability with

increased dose. In the case of the critical parameters, exceeding these parameters implies a drastic increase in the specific endpoint probability - particularly, SSB and DSB production. Certainly, mathematical models would enrich understanding of this idea. While some models have been proposed, uncertainty presents a tremendous problem; furthermore, disagreement among competing models is indicative of a lack of complete understanding.

Comparisons of the data to specific values obtained from literature has been done with two examples. These particular examples were chosen because of the clarity of the data and research presented in the papers describing their studies, as well as the extensiveness of the studies. However, the broad range of cell sensitivities would certainly dictate that other cell lines be studied. As well, concurrent measurements with the microdosimetry system and cell response would allow the investigator to pinpoint precise values of the time relation to energy deposition and cell response and repair.

XII. Recommendations

The biological application of measurements of dose in small volumes over short time spans has been presented along with practical applications. The application to medical purposes would be a worthy pursuit in light of the data presented here. While the detector used in this experiment was defined to be tissue equivalent, the actual variation in types of tissue and their reaction to types of radioactivity is evident as can be seen by the cell examples in the preceding section.

While this experiment made good progress into a little explored realm of microdosimetry, there is further research that could be done to enhance the current state of knowledge concerning microdosimetry in the temporal domain. Current knowledge in the field is greatly limited to theory. Further experimentation that could be done (some including the apparatus used for this experiment) include decreasing the integration time; adjusting the experimental setup to a time-of-flight (TOF) system; examination of specific sources, integration times, and gas (tissue) types; and possible incorporation of solid state detection methods.

Decreasing the integration time is closely related to the creation of a TOF system. Since energy deposition usually takes place in a very short time, it would be interesting to adjust the system such that the energy from only a partial particle passage is collected. This will enable the experimenter to monitor changes in lineal energy as the particle crosses the active volume. An experiment of this sort would require fast electronics and a certain amount of creativity. While the integrator chip used for this experiment is considered fast and sensitive, its speed would not support a TOF system. Implementation of a system of this sort would possibly require a change in system geometry. Rather than change the speed of the electronics, one could alter the particle path, somehow simulating a small tissue volume with a larger detector at very low pressure and using an anode of finite resistivity. The resulting signal could then be deconvolved to pick out specific secondary events. As well, multiple anode detectors might be useful. These are only two suggestions.

Creation of a TOF system with shorter integration times would certainly require the experimenter to move the integrator closer to the detector in the signal chain. For this experiment, the integrator integrated a pulse after it

had passed through an amplifier. For the integration times used, this was not a problem, as the time characteristics of individual events in the detector were much lower than the integration times. In addition, switching in the middle of an amplified pulse was purely arbitrary if it happened. However, if one wishes to study actual charge characteristics in the detector, then the rise time of the signal is important, and the integrator must be able to read this fast, low-amplitude signal. Most likely, it will be necessary to replace the pre-amplifier with a modified integrator circuit.

The experimental method outlined in this paper could feasibly be applied to any radiation source, allowing the researcher to determine the ability of the source to produce desired (or undesirable) endpoints in tissue. While only three sources were examined in this experiment for the purpose of example, any source can be explored. As well, any critical dose or any critical time can be explored. By examination of a particular source from the standpoint of a critical time or dose, irradiation regimens can easily be established by medical professionals administering treatments on a small scale.

Solid state measurements are mentioned because of the electronics used. The integrator chip used is simply a collection of solid state junctions. The idea of using its switches as surface barrier detectors would not be a bad one. The feedback switch in the integrator could very well present itself as a solid state microdosimeter. In the past, solid state microdosimetry has made use of CCDs and memory chips as the primary detection components. This might be a fairly interesting project.

Appendix A

SPHERICAL TISSUE-EQUIVALENT PROPORTIONAL COUNTER MODEL LET-1/2

Description

The FWT LET-1/2 is a spherical tissue-equivalent (TE) proportional counter suitable for measuring the spectra of absorbed dose in LET (actually P(Y) spectra). It is based upon designs originated by Dr. H. H. Rossi of Columbia University and uses a spiral grid for uniform collection characteristics.

The detector is housed in a 0.007-inch-thick aluminum shell which serves as a vacuum tight housing. The aluminum shell and attached stem are completely submersible and capable of being operated in any of the common TE fluids. The active volume of the detector is 0.500 inch in diameter and has a 0.05-inch-thick wall of TE plastic. The stem is approximately 10 inches long and has a connector block at one end carrying the

high voltage and signal connectors and a vacuum fitting. The block is suitable for clamping the detector to laboratory fixtures and is gasketed for moisture resistance.

The counter is designed to be evacuated and filled by the user with TE proportional counting gas at a low pressure, typically 2.5 to 40 cm Hg. These low pressures enable the spherical detector volume to simulate small tissue volumes from less than 1 to more than 8 microns in diameter.

Spectral Measurements

Connection of the detector to a preamplifier, such as used with Ge(Li) detectors, and a linear amplifier will enable a P(Y) spectrum to be accumulated on a pulse-height analyzer when the detector is exposed to a radiation source. Proper choice of amplifier gain settings will enable data to be obtained for Y values from 0.5 to 1000 keV per micron. These data may be converted to a spectrum of absorbed dose in LET with the aid of computer techniques or by hand if less resolution is satisfactory.

The internal ^{244}Cm alpha source makes calibration of the Y axis in keV per micron particularly simple. The source allows collimated alpha particles to cross the detector on a diameter, producing a peak in the pulse height analyzer display. The peak channel is characteristic of the average keV per micron expended by the alpha particles in traversing the sphere. The alpha source has a gravity-operated shutter mechanism that allows operation free of unwanted alpha counts when desired.

Dose Measurements Above a Bias in LET

Absorbed dose measurements above a bias level set at a particular LET value permits application of Quality Factors (QF) to the rads-tissue measured, a particularly easy technique with this detector and one that may be effected by either of two methods. One method takes the pulses from a linear amplifier and presents them to a linear gate. The discriminator driving the linear gate is calibrated in keV per micron with the aid of the ^{244}Cm source. The pulses passing through the gate (i.e., above the keV per micron bias level set on the discriminator) are then summed in a charge sensitive integrator: either a suitable electrometer, special ratemeter, or a current digitizer (ORTEC-439). This integral is proportional to the dose in rads-tissue above the bias and thus may be assigned a QF.

The other method uses an ADC, either as a separate unit or as part of a pulse-height analyzer. The linear amplifier pulses are presented to the ADC, whose discriminator controls the internal linear gate in a manner similar to the first method. The ADC produces a train of oscillator pulses that are eventually counted by the address scaler; if the pulses are scaled externally, their number is directly proportional to the absorbed dose in rads-tissue above the bias set with the discriminator. The internal ^{244}Cm alpha source may be used for calibration in this case. This arrangement offers the additional advantage of allowing the detector to be used as an absolute dosimeter, i.e., no reference to a standard source is necessary to produce accurate rad-tissue measurements.

SPECIFICATIONS

Tissue-Equivalent (TE) Plastic Sphere:	Wall thickness, 0.050 inch (0.127 cm); inside diameter, 0.500 inch (1.27 cm); Shonka Type A-150 plastic, 1.12 g/cm ³ .
Gas Filling Requirements:	TE proportional counting gas; 5.25 cm Hg of TE gas of density 1.062 g/liter corresponds to a cavity of 1 micron diameter; gas must be made up of high purity components and be moisture free.
Gas Filling Connector:	Mates with Swagelok type B-QC4-D-400
Signal Output:	Suitable for use directly into a low noise preamplifier (ORTEC 142A, ORTEC 142PC, TENNELEC TC 174, etc.). Output is AC coupled.
Signal Connector:	Mates with BNC UG260 B/U or equal.
Signal Cable:	Short length (<6 in.) of low noise coax. MALCO 250-3834-0000 for example.
High Voltage Requirements:	Positive polarity, ~450 volts at 2.6 cm Hg, ~550 volts at 5.25 cm Hg, ~650 volts at 10.5 cm Hg.
High Voltage Connector:	Mates with UG932 A/U or equal.
High Voltage Cable:	RG-59/U or equal.
Calibration Source:	^{244}Cm , ~1.0 microcuries; average alpha energy ~5.80 MeV; gravity operated shutter.
Exterior Can:	Vacuum tight aluminum; wall thickness, 0.007 inch (0.0178 cm); grounded to serve as electrostatic shield; maximum O.D. of detector end 0.825 inch (2.1 cm).
Overall Size and Weight:	Length including vacuum connector 15.75 inches (40.0 cm); weight, 230 grams; wooden storage box weight, 1540 grams.
Modifications Available:	Thicker aluminum can — up to 0.020 inch (0.051 cm); no alpha source; 0.100 inch thick wall.

June 1987

Appendix B

GAS FILLING SYSTEM MODEL GFS-1

FWT's gas filling system is an accessory for the various LET counters and Multiplying Ionization Chambers presently available. Its principal purpose is to serve as an interface between a mechanical vacuum pump, a supply of tissue-equivalent gas, and the LET counter.

DESCRIPTION

The gas filling system is composed of 1) an accurate mechanical vacuum gauge with a range from 0 to 20 cm Hg; 2) connections to allow attachment of a mechanical vacuum pump and a supply of tissue equivalent gas; 3) bellows-sealed valves with teflon seats to control the vacuum and gas inlets and to serve as a protection device for the gauge; and 4) a connector to allow attachment of the LET counter. The parts are assembled on a rigid aluminum plate with deep engravings showing the vacuum piping and component function.

The system has been assembled with extreme care so that contamination is at a minimum. Individual components and the assembled complete system have been helium leak tested to insure leak-free performance.

SPECIFICATIONS

Valves:	Bellows type, Teflon seats for positive shutoff
Gas Inlet Connector:	Mates with Swagelok B-QC4-D-400
Counter Inlet Connector:	Mates with Swagelok B-QC4-D-400
Vacuum Pump Connector:	Pipe suitable for clamping 1/2-in. to 3/4-in. I.D. vacuum hose
Vacuum Pump Requirements:	Mechanical pump, >30 liters per minute, 5×10^{-5} mm Hg ultimate, liquid nitrogen cold trap desirable
Vacuum Gauge:	Aneroid type, 0 to 20 cm Hg standard, 6-inch dial, accuracy $<\pm 0.012$ cm Hg
Leakage:	Individual parts and complete system helium leak checked, leak rate $<2 \times 10^{-5}$ std cc/sec
Dimensions:	16 in. (40.5 cm) long, 13 in. (33 cm) wide, and 10 in. (25 cm) high
Weight:	19 lbs (8.6 kg) with cover

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Various modifications of the system are possible, such as vacuum gauges of different ranges, additional valves and connectors, rack mounting, etc. Please contact FWT for additional information.

Data and Specifications Subject to Change Without Notice

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Appendix D: Dose Spectra

Part 1: Typical Dose Spectra

Figure D2. Cesium-137 Spectrum
Typical Second Moment Spectrum

Figure D3. Sr/Y Dose Spectrum
Typical Frequency Spectrum

Figure D4. Sr/Y Dose Spectrum
Typical First Moment Spectrum

Figure D5. Sr/Y Dose Spectrum
Typical Second Moment Spectrum

Figure D6. Typical X-Ray Dose Spectrum

250 kVp x rays, Frequency Spectrum

Figure D7. Typical X-Ray Dose Spectrum

250 kVp x rays, First Moment Spectrum

Figure D8. Typical X-Ray Dose Spectrum

250 kVp x rays, Second Moment Spectrum

Part 2: Integrated Signal Dose Spectra For ^{137}Cs

Figure D9. Cesium Dose Spectra for Various Integration Times

Second Moment Spectra

Figure D10. Cesium Dose Spectra for Various Integration Times

Second Moment Spectra

Figure D11. Cesium-137 Spectrum

5 ms Integration Time

Figure D12. Cesium 137 Spectrum

2 ms Integration Time

Figure D13. Cesium-137 Spectrum

1 ms Integration Time

Figure D14. Cesium-137 Spectrum

500 Microsecond Integration Time

Figure D15. Cesium-137 Spectrum

100 Microsecond Integration Time

Figure D16. Cesium-137 Spectrum

50 Microsecond Integration Time

Figure D17. Cesium-137 Spectrum

10 Microsecond Integration Time

Figure D18. Cesium-137 Spectrum

1 Microsecond Integration Time

Figure D19. Cesium Dose Spectra for Various Integration Times

First Moment Spectra

Figure D20. Cesium Dose Spectra for Various Integration Times

First Moment Spectra

Figure D25. Cesium-137 Spectrum

100 Microsecond Integration Time

Figure D26. Cesium-137 Spectrum

50 Microsecond Integration Time

Figure D27. Cesium-137 Spectrum

10 Microsecond Integration Time

Figure D28. Cesium-137 Spectrum

1 Microsecond Integration Time

Figure D29. Cesium Dose Spectra for Various Integration Times

Frequency Spectra

Figure D30. Cesium Dose Spectra for Various Integration Times

Frequency Spectra

Figure D31. Cesium-137 Spectrum

5 ms Integration Time

Figure D32. Cesium 137 Spectrum

2 ms Integration Time

Figure D33. Cesium-137 Spectrum

1 ms Integration Time

Figure D34. Cesium-137 Spectrum

500 Microsecond Integration Time

Figure D35. Cesium-137 Spectrum

100 Microsecond Integration Time

Figure D36. Cesium-137 Spectrum

50 Microsecond Integration Time

Part 3. Integrated Signal Dose Spectra For ²⁵⁰ kVp X Rays

Figure D41. X-Ray Dose Spectrum

5ms Integration Time

Figure D42. X-ray Dose Spectrum

2 ms Integration Time

Figure D45. X-Ray Dose Spectrum

100 Microsecond Integration Time

Figure D46. X-Ray Dose Spectrum

50 Microsecond Integration Time

Figure D47. X-Ray Dose Spectrum

10 Microsecond Integration Time

Figure D48. X-Ray Dose Spectrum

1 Microsecond Integration Time

Figure 49. X-Ray Dose Spectra for Various Integration Times

First Moment Spectra

Figure D50. X-Ray Dose Spectra for Various Integration Times

First Moment Spectra

Figure D53. X-Ray Dose Spectrum

1 ms Integration Time

Figure D54. X-Ray Dose Spectrum

500 Microsecond Integration Time

Figure D55. X-Ray Dose Spectrum

100 Microsecond Integration Time

Figure D56. X-Ray Dose Spectrum

50 Microsecond Integration Time

Figure D57. X-Ray Dose Spectrum

10 Microsecond Integration Time

Figure D58. X-Ray Dose Spectrum

1 Microsecond Integration Time

Figure D59. X-Ray Dose Spectra for Various Integration Times
Frequency Spectra

Figure D60. X-Ray Dose Spectra for Various Integration Times
Frequency Spectra

Figure D61. X-Ray Dose Spectrum

5ms Integration Time

Figure D62. X-ray Dose Spectrum

2 ms Integration Time

Figure D65. X-Ray Dose Spectrum

100 Microsecond Integration Time

Figure D66. X-Ray Dose Spectrum

50 Microsecond Integration Time

Figure D67. X-Ray Dose Spectrum

10 Microsecond Integration Time

Figure D68. X-Ray Dose Spectrum

1 Microsecond Integration Time

Part 4. Integrated Signal Dose Spectra For $^{90}\text{Sr}/^{90}\text{Y}$ Beta Particles

Figure D71. Sr/Y Dose Spectrum

5 ms Integration Time

Figure D72. Sr/Y Dose Spectrum

2 ms Integration Time

Figure D73. Sr/Y Dose Spectrum

1 ms Integration Time

Figure D74. Sr/Y Dose Spectrum

500 Microsecond Integration Time

Figure D75. Sr/Y Dose Spectrum

100 Microsecond Integration Time

Figure D76. Sr/Y Dose Spectrum

50 Microsecond Integration Time

Figure D77. Sr/Y Dose Spectrum

10 microsecond Integration Time

Figure D78. Sr/Y Dose Spectrum

1 microsecond Integration Time

Figure D81. Sr/Y Spectrum

5 ms Integration Time

Figure D82. Sr/Y Dose Spectrum

2 ms Integration Time

Figure D83. Sr/Y Dose Spectrum

1 ms Integration Time

Figure D84. Sr/Y Dose Spectrum

500 Microsecond Integration Time

Figure D87. Sr/Y Dose Spectrum

10 microsecond Integration Time

Figure D88. Sr/Y Dose Spectrum

1 microsecond Integration Time

Figure D89. Sr/Y Dose Spectra Based on Integration Time

Frequency Spectra

Figure D90. Sr/Y Dose Spectra Based on Integration Time

Frequency Spectra

Figure D93. Sr/Y Dose Spectrum

1 ms Integration Time

Figure D94. Sr/Y Dose Spectrum

500 Microsecond Integration Time

Figure D97. Sr/Y Dose Spectrum

10 microsecond Integration Time

Figure D98. Sr/Y Dose Spectrum

1 microsecond Integration Time

Appendix E: Calculation of Tissue Dose from Radiation Sources Used

Dose rates in tissue for the three sources used in the configurations used is presented separately here for each source. References for this section are presented within the section.

E.I. Cesium-137

The activity of the ^{137}Cs source was 26.37 mCi (9.757 Bq) at the time of the experiment. The total photon flux at the detector is calculated as:

$$\phi = \frac{AY}{4\pi d^2} \quad (\text{E1})$$

where A is the activity in Bq, Y is the photon yield for ^{137}Cs (in secular equilibrium with its radioactive product $^{137\text{m}}\text{Ba}$), and d is the detector distance from the source (2 cm). The calculated photon flux is then $1.650 \times 10^7 \text{ } \gamma/\text{s-cm}^2$. (Air attenuation is neglected at this close distance.) The conversion to exposure rate in air is given by:

$$\dot{X} = \frac{\phi}{\phi'} \quad (\text{E2})$$

where ϕ' is given by the empirical formula taken using ICRU tissue:

$$\phi' = \frac{1.51 \times 10^4}{\mu_{en}/\rho_{air} E(\text{MeV})} \quad (\text{E3})$$

for 0.662 MeV gamma photons, μ_{en}/ρ for air is $0.0293 \text{ cm}^2/\text{g}$. The conversion factor is then calculated to be $7.785 \times 10^5 \text{ } (\gamma/\text{s-cm}^2)/(\text{R/h})$. The exposure rate in air is then calculated to be 21.195 R/h. Using the NIST conversion factor of $8.78 \times 10^{-3} \text{ Gy/R}$, then the air kerma dose is calculated to be 0.186 Gy/h or $5.169 \times 10^{-5} \text{ Gy/s}$. To convert this kerma dose to dose in tissue, the general formula for kerma is used:

$$K = \Psi \cdot \frac{\mu_{tr}}{\rho} \cdot 10^{-4} \quad (\text{E4})$$

where the energy transfer coefficient μ_{tr} is calculated by:

$$\mu_{tr} = \mu \frac{\bar{E}_k}{E_o} \quad (\text{E5})$$

where μ is the total linear attenuation coefficient of the material. Thus, the ratio of the dose in tissue to that in air is given by:

$$S_{a \rightarrow t} = \frac{K_t}{K_a} = \frac{\mu_t}{\mu_a} \quad (\text{E6})$$

However, for energy deposition, the f factor is used (which incorporates μ_{en}) in dose calculations:

$$f = \frac{\mu_{en_t}}{\mu_{en_a}} \quad (\text{E7})$$

so that the dose in tissue is estimated to be equal to the air kerma dose times the conversion factor given in Equation E7. The attenuation coefficients are taken for adipose tissue and dry air at sea level from the *Radiological Health Handbook*. The calculated value for f for ^{137}Cs photons is then 1.120. (This is also equal to the S .) Thus the calculated dose in tissue (at the surface, the shallow depth) is then calculated to be 5.789×10^{-5} Gy/s.

E.II. X Rays

The dose from 250 kVp x rays is estimated by first measuring the exposure from the source. Using an NIST calibrated Shonka-Wyckoff ionization chamber at an anode voltage of 500V, the ionization current was measured, and the following relation was applied to obtain the exposure rate in air:

$$\dot{X} = 60000NiC_{air} \quad (\text{E8})$$

where N is the calibration factor for the chamber and x ray method used (in this case, taken to be 3.017×10^{-5} Gy/C), i is the measured ionization current (corrected for leakage), and C is the correction factor for air, calculated by:

$$C_{air} = \frac{(273.15 + T)760}{295.15P} \quad (\text{E9})$$

where T is the air temperature, and P is the pressure.

It is noted that the calibration factor is given in Gy/C. Since the air kerma dose is directly proportional to the exposure, a direct relation has been made during detector calibration, so that the resulting value is in mGy/min. Using this method, the air kerma dose was measured to be 0.441 mGy/min or 7.348×10^{-6} Gy/s.

Using Equation E6, the conversion factor from air to tissue is made for x rays. Since the attenuation coefficients are energy dependent, and the x ray source emits a spectrum of energies, the average energy was estimated to be half of the maximum energy value, or 115 keV. The conversion factor f was then calculated to be 1.064, so that the tissue dose rate of the x rays is then estimated to be 7.817×10^{-6} Gy/s.

E.III. Beta Particles

Since the electrons from the ^{90}Sr source do not penetrate the aluminum of the detector, the dose rate from only the ^{90}Y electrons are calculated. Calculation of the dose rate is made using the empirical Loevinger's equation:

$$D = \frac{k}{(vx)^2} \left\{ c \left[1 - \frac{vx}{c} e^{(1-vx/c)} \right] + vx e^{(1-vx)} \right\} \quad (\text{E10})$$

where D is the dose per electron, x is the mass distance from the source (g/cm^2), v is the apparent attenuation coefficient (taken in air since that is the medium through which the particles passed), and k and c are empirical values taken as:

$$k = 1.28 \times 10^{-9} \rho^2 v^3 \bar{E}_\beta \alpha \quad (\text{E11})$$

$$c = 3.11 e^{-0.55 E_{\max}} \quad (\text{E12})$$

For ^{90}Y electrons, the average particle energy is taken from the *Radiological Health Handbook* to be 0.931 MeV and the maximum energy is 2.245 MeV. The apparent attenuation coefficient is calculated as:

$$\nu = \frac{16}{(E_{\text{max}} - 0.036)^{1.40}} \quad (\text{E13})$$

Given the above data and the experimental conditions, the dose per incident particle in air is calculated as 1.042×10^{-11} rad/ β . Conversion to a dose rate with respect to time is calculated by multiplying the dose per particle by the source activity in Bq and the abundance (100% for ^{90}Y). The dose rate in air is then calculated to be 8.096×10^{-3} rad/s. Conversion can then be made to tissue in a similar manner as given in Equation E7 - multiplying by the ratios of the stopping powers of the incident particles in tissue to air. This can be calculated using Equation 42 to be 1.132. Therefore, the dose of the incident electrons from the ^{90}Y source in tissue is calculated to be 9.165×10^{-3} rad/s or 9.165×10^{-5} Gy/s.

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