

5. Tissue equivalence correction for silicon microdosimeters

ICRU report 44 [32] states "Tissue substitutes are often mixtures formulated so that their radiation interaction properties rather than their atomic composition, match those of the body tissue to the degree necessary for the specific application". Similarly, radiation detectors due to operational and construction requirements (such as electrical conductivity, stability, mechanical strength, and shape construction) are rarely made of components with identical composition to tissue. Instead, the pertinent radiation characteristics and mass densities of the body tissue and detector are evaluated and correction factors frequently employed.

The *macrodosimetric* tissue equivalency of silicon or silicon dioxide for integral low LET photon irradiation is well established [175]. However, methods for converting silicon based *microdosimetric* measurements to tissue volumes have previously not been investigated.

For macrodosimetric measurements, dose may be scaled according to the Bragg-Gray theorem [176-178]. Assuming that the detector is small in comparison with the range of secondary particles then the relationship between detector dose $D_{detector}$ and tissue dose D_{tissue} is given by:

$$D_{tissue} = \frac{(\bar{S} / r)_{tissue}}{(\bar{S} / r)_{detector}} D_{detector} \quad (5.1)$$

where $\bar{S} / r$ is mass collision stopping power averaged over the energy spectrum of charged particles. For photons, provided the detector does not perturb the photon spectrum, equation (5.1) becomes

$$\begin{aligned} D_{tissue} &= \frac{(\bar{m}_{en} / r)_{tissue}}{(\bar{m}_{en} / r)_{detector}} D_{detector} \\ &= k D_{detector} \quad \text{for Silicon Detector} \end{aligned} \quad (5.2)$$

where $\bar{m}_{en} / r$ is the mass energy absorption coefficient averaged over the photon spectrum. The scaling factor k is simply 1.09 for silicon with photons in the energy range 0.3 to 3 MeV [179].

For microdosimetric measurements, a spectrum is produced as opposed to a single dose measurement. A simple method for interpreting silicon microdosimetric spectra as equivalent tissue measurements is described in this chapter.

The results are presented using boron neutron capture therapy (BNCT) as a test case. We first compare silicon and tissue range-energy relationships for H, He, Li and C ion deposition. The Monte-Carlo method is used to compare energy deposition spectra for some simple geometrical cases relevant to boron neutron capture therapy. The tissue equivalency of silicon detector/converter systems under fast neutron and proton irradiation is also considered.

5.1 Definition of Tissue and Detector Materials

Before comparing silicon to tissue, the composition of tissue must first be defined. Biological tissue composition varies significantly between tissue types and individuals. The subject of tissue references is comprehensively discussed in ICRU report #44 [32] and a report by the ICRP [180]. These reports identified the most important tissues needing simulation as muscle, adipose tissue and the skeleton. Together these make up over 70% of the body mass. Lung tissue is also important due to its large density difference. Muscle includes the connective tissue, blood vessels, blood, lymph, etc., generally associated with skeletal (striated) muscle. Adipose tissue is composed of a protein matrix supporting cells (adipocytes) highly specialized for the storage of lipids. The most common tissue reference in microdosimetric literature [11], and the one we shall use here, is commonly identified as muscle (striated, ICRU, 1964) [181]. ICRU report 44 [32] identifies a composition called muscle (skeletal, ICRU, 1989) as the recommended muscle composition. However, the difference with muscle (striated) is small (an additional 2% carbon content and 2% less oxygen content) and we maintain conformity with the literature in using muscle (striated). The composition of muscle (striated ICRU, 1964), heretofore called muscle, is defined in Table 5.1. In addition, the term "tissue equivalent" refers to muscle (striated ICRU, 1964) equivalence. The primary elements in tissue are hydrogen, carbon, oxygen and nitrogen. For a given tissue substitute the quality of the tissue approximation depends on the composition and density matching along with the type and energy of the radiation being used. Also described in Table 5.1 are the composition and some relevant properties of several materials (A150, Lucite, and silicon) which will be compared to muscle.

The standard wall material used in the majority of proportional gas counters is a plastic designated A150, which is considered "muscle equivalent". A150 is a mixture of polyethylene, nylon, carbon and calcium fluoride, which is easily molded, welded and machined. The material was originally developed by Shonka and collaborators [33] and subsequently modified by Smathers [182]. The elemental composition of A150 is very close to muscle in regard to hydrogen and nitrogen content. However, the oxygen content is much lower and this has been substituted by an increased carbon content to obtain suitable electrical conductivity. Note that no usable solid exists with the required oxygen content of tissue. The effect of this discrepancy in elemental composition on neutron microdosimetric measurements is discussed further in section 5.5.

Lucite (or Perspex) is a readily available material often used as a tissue substitute in phantoms. The tissue converter placed above the silicon microdosimeter in this work was constructed of Lucite. This material was selected due to availability, low cost and construction ease. The hydrogen content is slightly lower than for muscle and there is no nitrogen. Like A150, the carbon and nitrogen contents do not correctly reflect muscle. Again, the effect of this discrepancy in elemental composition will be discussed further in section 5.5.

Table 5.1. Elemental composition of important tissue types, substitutes and detector materials.

			Muscle(Striated ICRU, 1964)		A150		Lucite		Silicon
Atom	Z	A	Atomic %	Mass %	Atomic %	Mass %	Atomic %	Mass %	Mass & Atomic %
H	1	1.01	63.31	10.2	58.26	10.1	53.33	8.05	-
C	6	12.01	6.41	12.3	37.61	77.7	33.33	59.98	-
N	7	14.01	1.56	3.5	1.45	3.5	-	-	-
O	8	16.00	28.51	72.91	1.89	5.2	13.33	31.96	-
F	9	19.00	-	-	0.52	1.7	-	-	-
Na	11	22.99	0.02	0.08	-	-	-	-	-
Si	14	28.09	-	-	-	-	-	-	100.00
P	15	30.97	0.04	0.2	-	-	-	-	-
S	16	32.06	0.10	0.5	-	-	-	-	-
K	19	39.10	0.05	0.3	-	-	-	-	-
Ca	20	40.08	0.001	0.007	0.26	1.8	-	-	-
Mean Atomic Mass				14.10		11.80		12.40	28.09
Density (g/cm ³)				1.04		1.127		1.2	2.32
Atomic Density ×10 ²³ atoms/cm ³				1.00		1.17		1.08	0.50
Electron Density ×10 ²³ atoms/cm ³				3.44		3.73		3.90	6.99

Note 1: The Mean Atomic Mass was calculated as the mass weighted average

(i.e. $\sum(\text{mass fraction}_i \times A_i)$)

2: Lucite (or Perspex) is Polymethyl Methacrylate, C₅H₈O₂ [126].

3: Definition of Muscle (Striated ICRU,1964) and A150 from Appendix C [11]

5.2 Microdosimetry TE Correction Factor

Frequently when analyzing microdosimeter performance researchers have often concentrated on LET. However, the energy deposited in a microdosimeter depends upon the *integral* of the particle LET over the chord length traversed. This implies that one should consider the particle range-energy relationship rather than the LET-range relationship as a starting point in analyzing microdosimetry tissue equivalence. The equation for range of a particle with initial energy E_{max} is

$$R = \int_0^{E_{max}} \left(-\frac{dE}{dx} \right)^{-1} dE \quad (5.3)$$

where dE/dx is the particle LET or linear stopping power. Particles primarily lose energy via coulomb interactions with the material's electrons. An approximate relationship for dE/dx is given by the Bethe formula based on relativistic quantum mechanics [23]:

$$-\frac{dE}{dx} = \frac{4pe^4 z^2}{m_0 v^2} NZ \left[\ln \frac{2m_0 v^2}{I} - \ln \left(1 - \frac{v^2}{c^2} \right) - \frac{v^2}{c^2} \right] \quad (5.4)$$

where v and ze are the velocity and charge of the particle, N and Z are the number density and atomic number of the absorber atoms, m_0 is the electron rest mass, and e is the electronic charge. The parameter I represents the average excitation and ionization potential of the absorber and is normally treated as an experimental parameter for each element. The following approximate empirical formula may be used to estimate the I value in eV for an element with atomic number Z [183].

$$I = \begin{cases} 19.0 \text{ eV}, Z = 1 \text{ (hydrogen)} \\ 11.2 + 11.7Z \text{ eV}, 2 \leq Z \leq 13 \\ 52.8 + 8.71Z \text{ eV}, Z > 13 \end{cases} \quad (5.5)$$

When the material is a compound or mixture, the stopping power may be calculated by simply adding the contributions from the individual constituents (Bragg additivity rule). If there are T elements with the i -th element described by an atomic density N_i atoms/cm³, atomic number Z_i and mean excitation energy I_i then we use:

$$n \ln I = \sum_{i=1}^T N_i Z_i \ln I_i \quad (5.6)$$

where n is the total number of electrons/cm³ in the material ($= NZ$). For non-relativistic particles, only the first term in the brackets of equation (5.4) is significant. In comparing different materials as absorbers dE/dx depends primarily on NZ (the electron density), which is outside the logarithmic term.

Unfortunately, the integral formed by applying equation (5.4) into (5.3) does not have a simple analytic solution. Nevertheless, a semi-empirical scaling formula for range (sometimes called the Bragg-Kleeman rule) enables the comparison of ranges in different materials [23].

$$\frac{R_2}{R_1} \cong \frac{r_1 \sqrt{A_2}}{r_2 \sqrt{A_1}} \quad (5.7)$$

where r and A represent density and atomic weight and the subscripts refer to different materials. The accuracy of this estimate diminishes with large differences in atomic number. Note that the equation is independent of the ion type, charge or energy. Higher accuracy in range comparisons may be gained by using Ziegler's SRIM (Stopping and Ranges of Ions in Matter) program [126, 184]. Note that the range considered in this discussion is the mean range since as discussed in chapter 3, range and energy straggling produce a distribution of ranges for a given incident particle energy.

The general shape of the range-energy and LET-range relationships is quite similar between different materials for all ion types. Therefore, the range-energy curve in one material $E_1(R)$ will approximate the range-energy curve of another $E_2(r)$ provided the ranges are scaled appropriately according to equation (5.7) or using other range calculations. That is,

$$E_1(R) \cong E_2(zR) \quad (5.8)$$

where $R = \text{range}$ and $z = \frac{R_2}{R_1} \cong \frac{r_1 \sqrt{A_2}}{r_2 \sqrt{A_1}}$

The energy deposited E_{dep1} by a charged particle traversing a reference volume with material denoted by subscript 1 is given by

$$\begin{aligned} E_{dep1} &= \int_{a1}^{b1} \frac{dE_1}{dx} dx \\ &= E_1(b_1) - E_1(a_1) \end{aligned} \quad (5.9)$$

where $s_1 = b_1 - a_1$ is the chord length. Similarly, for another volume of differing material denoted by subscript 2 the energy deposited E_{dep2} is given by

$$\begin{aligned} E_{dep2} &= \int_{a_2}^{b_2} \frac{dE_2}{dx} dx \\ &= E_2(b_2) - E_2(a_2) \end{aligned} \quad (5.10)$$

where $s_2 = b_2 - a_2$ is the chord length. Using equation (5.8) substituted into (5.10) gives

$$E_{dep2} \cong E_1(b_2 / z) - E_1(a_2 / z) \quad (5.11)$$

Comparing equations (5.9) and (5.11) indicates that the energy deposited in the two volumes will be equal ($E_{dep1} = E_{dep2}$) provided that $b_1 = b_2/z$ and $a_1 = a_2/z$ which equates to $s_2 = zs_1$. That is, the energy deposited is equal in the two volumes if the chord lengths are scaled by the factor z . For multiple chords, the scaling requires equivalent shapes (and chord length distribution shape) with a scaling of the linear dimensions. *More generally, assuming identical charged particle spectra, the microdosimetric energy spectrum will be equivalent in two volumes of differing composition if the shape of the volumes is the same and the dimensions are scaled by the factor z .*

For a tissue rectangular parallelepiped (RPP) with dimensions $x \times y \times z$ the required scaling is $z \times x \times z \times y \times z$ of silicon. Note that this holds for any RPP (it is not necessary that $x=y=z$) and for other shapes such as a cylinder (the radius and length are scaled) or a sphere (radius scaling only). The scaling factor simply applies to each linear dimension. To obtain equal TE lineal energy ($E/\bar{l}$) in both a tissue volume and a correctly scaled silicon volume we must use the same mean chord length $\bar{l}$, corresponding to the *tissue* volume, since the energy deposited is the same.

The required scaling factor z may be estimated by equation (5.7). Values calculated for various materials with respect to muscle are shown in Table 5.2. The scaling factor calculated for 1 MeV protons, using the projected range algorithm (PRAL) option of SRIM [126, 184], is within 7% of the factor calculated using the approximate equation (5.7). The tissue equivalence factor for Lucite is quite close (within 5%) of the factor for A150.

Note that the possible equivalence of silicon and tissue via a geometrical scaling factor assumes that the charged particle spectra are equivalent. To obtain equivalent charged

particle spectra the atomic composition of the volumes should closely match in order to obtain the same interactions (capture, elastic, inelastic and nuclear reactions) and subsequent reaction products. In the case of a silicon detector, we require the use of a tissue substitute converter lying above the detector. Furthermore, the range of recoil particles should be such that crossers and stoppers dominate the microdosimetric spectrum as discussed extensively in chapter 3.

Table 5.2. Tissue equivalence scaling factor for A150, Lucite and Silicon

	A150	Lucite	Silicon
z from equation (5.7)	0.84	0.81	0.63
z from PRAL (1MeV protons)	0.88	0.84	0.68

Note: The range of 1 MeV proton in muscle is 23.03 μm calculated using PRAL, the projected range algorithm option of SRIM [126, 184]

5.3 Summary of Conditions for Tissue Equivalence

In summary, to obtain tissue equivalence the silicon detector/ TE converter must satisfy four main conditions

1. The geometrical TE scaling factor for the silicon detector ($z = 0.63$) should apply to all ion products
2. T.E converter must closely approximate tissue atomic composition with respect to atoms generating the dominant secondary charged particles.
3. The proportion of interactions with silicon must be negligible in comparison with converter interactions. That is, events should originate from converter recoil products or primary ions rather than from within silicon.
4. The range of recoil particles should be such that crossers and stoppers dominate the microdosimetric spectrum as discussed extensively in chapter 3

The effect of tissue substitute composition will be further discussed in section 5.5 where we consider Fast Neutron Therapy (FNT) and Proton Therapy (PT) conditions. The concept of scaling ranges to establish tissue equivalence will be further clarified by consideration of the Boron Neutron Capture Therapy (BNCT) reactions in the following section.

5.4 Boron Neutron Capture Therapy Tissue Equivalence

The principle and application of microdosimetry to BNCT is described in chapter 7. In this section, the interaction of neutron capture products with silicon and tissue will be considered in detail to test TE relationships uncovered in the previous section.

5.4.1 Comparison of Silicon and Tissue BNCT Range-Energy Relationships¹

The BNCT reactions provide an excellent platform for testing the geometrical TE factor discussed in the previous section. Four ion types over a range of energies are involved. The primary thermal neutron reactions with tissue are as follows:

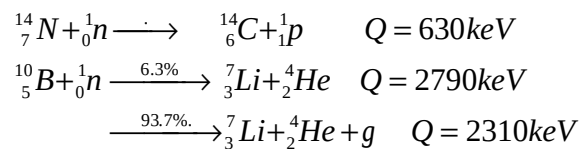


Table 5.3. Summary of Energy-Range data for ions produced by thermal neutron interactions with nitrogen and boron

Reaction	Ion	Energy (keV)	Range(μm)	
			Tissue	Silicon
${}^{14}\text{N}(\text{n,p}){}^{14}\text{C}$	p	590	10.5	7.2
	${}^{14}\text{C}$	40	0.20	0.12
${}^{10}\text{B}(\text{n},\alpha){}^7\text{Li}$	α (6.3%)	1780	9.3	6.3
	${}^7\text{Li}$ (6.3%)	1010	4.6	2.8
	α (93.7%)	1470	7.7	5.1
	${}^7\text{Li}$ (93.7%)	840	4.1	2.5

Table 5.4. Summary of interaction data for selected BNCT conditions♣

Reaction	σ (barns/atom)	Interactions/ μm^3	Ions	P(ion pair emitted)
${}^{14}\text{N}(\text{n,p}){}^{14}\text{C}$	1.81	0.0272	p, ${}^{14}\text{C}$	0.37
${}^{10}\text{B}(\text{n},\alpha){}^7\text{Li}$	3837	0.0462	α , ${}^7\text{Li}$ (6.3%)	0.04
			α , ${}^7\text{Li}$ (93.7%)	0.59

♣ Both reactions are for a neutron fluence of 10^{13} n/cm² [186]

Nitrogen concentration 3.5g/100g, Boron concentration 20μg/g.

For other concentrations and fluence the interactions per unit volume changes proportionally.

P(ion pair emitted) is the proportion of interactions which produce the respective ion pair.

The ion products and their energies are summarized in Table 5.3. Ion range and energy data for each of these ions was calculated using the computer code SRIM [126]. The data was calculated for both silicon and tissue using the projected range algorithm

¹ The work presented in this section and the following one was originally published by the author in 1998 [185] P. D. Bradley and A. B. Rosenfeld, "Tissue Equivalence Correction for Silicon Microdosimetry Detectors in Boron Neutron Capture Therapy," *Med. Phys.*, vol. 25, pp. 2220-2225, 1998..

(PRAL) option of SRIM [126, 184]. This option provides range-energy tables that are within a few percent of the more accurate but time consuming TRIM-Monte-Carlo calculations available in SRIM. The accuracy of PRAL was considered adequate for our purposes given that we are primarily interested in a comparative study between silicon and tissue. TRIM-Monte-Carlo calculations using the ion/energy combinations given in Table 5.3 yielded average longitudinal range estimates within 2% of the PRAL estimated ranges.

Tissue was modeled as per ICRU striated muscle, which is defined with an elemental composition and density given in Table 5.1. Using these models, the range-energy data is plotted in Figure 5.2 for the ions relevant to BNCT. The corresponding LET versus energy and LET versus range relationships are provided in Figure 5.3 as a useful reference. A 5th order polynomial was fitted to the range-energy data for each of the ion/target combinations with the results given below.

H, He, Li and C ions in Silicon:

$$E(r, H, Silicon) = -0.07492r^5 + 1.713r^4 - 13.27r^3 + 34.89r^2 + 80.99r$$

$$E(r, He, Silicon) = -0.2455r^5 + 15.81r^4 - 51.49r^3 + 198.7r^2 + 9.229r$$

$$E(r, Li, Silicon) = 4.398r^5 - 44.75r^4 + 141.6r^3 - 63.99r^2 + 141.1r$$

$$E(r, C, Silicon) = -2.442 \times 10^6 r^5 + 862600r^4 - 109200r^3 + 5922r^2 + 211.8r$$

H, He, Li and C ions in Muscle:

$$E(r, H, Muscle) = -0.01148r^5 + 0.398r^4 - 4.787r^3 + 21.2r^2 + 40.05r$$

$$E(r, He, Muscle) = 0.00564r^5 + 0.1188r^4 - 5.84r^3 + 58.06r^2 + 17.36r$$

$$E(r, Li, Muscle) = 0.03762r^5 - 1.534r^4 + 11.73r^3 + 6.404r^2 + 72.66r$$

$$E(r, C, Muscle) = -1.660 \times 10^5 r^5 + 97530r^4 - 20210r^3 + 1727r^2 + 146.5r$$

The above equations are only valid up to the ion ranges specified in Table 5.3. Comparison of the range-energy relationships for tissue and silicon indicates that the data may be related by a simple scaling factor, z . Scaling the silicon plots by substituting $0.63 r$ ($z = 0.63$) into the above equations yields a close approximation to the tissue range-energy relationship as shown in the right hand plots of Figure 5.2. These results are completely consistent with the previous discussion and value of z given in Table 5.2. As predicted by equation (5.7), the ratio z of the ion range in silicon to the range in tissue is only weakly dependent on the ion species and energy. Note that

the geometrical scaling is not obvious by inspection of the LET curves provided in Figure 5.3.

An average value of z was calculated for each ion using the following equation:

$$z(\text{ion}) = \frac{1}{E_{\max}} \int_0^{E_{\max}} \frac{R(E, \text{ion}, \text{silicon})}{R(E, \text{ion}, \text{tissue})} dE \quad (5.12)$$

where $R(E, \text{ion}, \text{target})$ is the range as a function of energy (E) for the required *ion* and *target*. Values of z for each ion are shown in Table 5.5. The optimum scaling factor, z , is dependent on the contribution of each ion to the energy deposition spectra which in turn is dependent on the segment length distribution of the sensitive volume, ion energy and relative frequency of ion generation. An approximate scaling factor was calculated by weighting each ion according to the energy of the ion and the relative frequency of generation. The required cross-section and boron and nitrogen concentrations are given in Table 5.4 taken from typical values used by Charlton [186]. The final weighted average scaling factor is $z = 0.63$. The maximum deviation from this scaling factor is 8% and occurs for the highest energy proton. This factor is in excellent correspondence with the approximate range ratio equation (5.7). Note, that a more accurate weighting of ions would consider the difference in contribution to dose mean lineal energy $\bar{y}_D$ of the various ions. Ions depositing a higher energy (usually higher LET ions) contribute more to $\bar{y}_D$ and thus should be assigned a higher weighting factor.

Table 5.5. Mean ratio of BNCT ion ranges in silicon and tissue

Ion	z	Weight
H	0.65	0.13
He	0.64	0.55
Li	0.60	0.31
C	0.60	0.01
Weighted Av.	0.63	

Note: Weight is an approximate measure of the contribution of the ion to BNCT energy deposition spectra

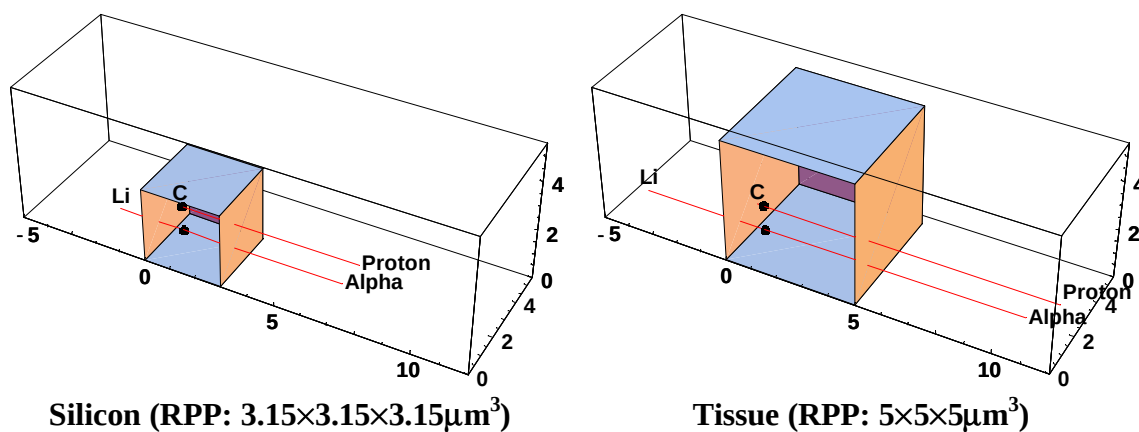


Figure 5.1. Comparison of BNCT ion product ranges in tissue and silicon. The silicon RPP dimensions have been scaled by 0.63 times the tissue dimensions. Note that the figure on the left (silicon) is an exact scale miniaturization of the figure on the right (tissue). These volumes will have equivalent microdosimetric spectra assuming equal concentrations of nitrogen and boron.

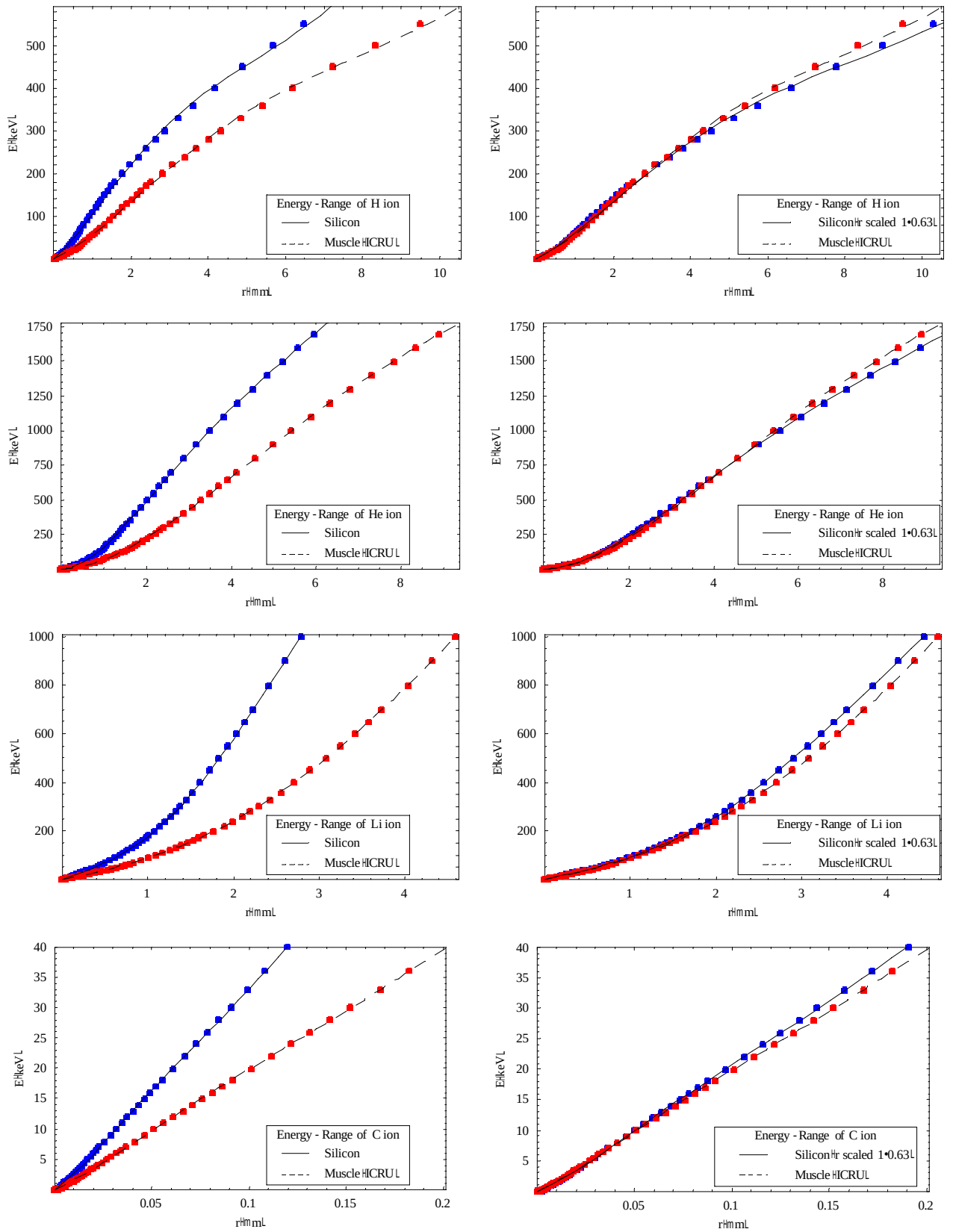


Figure 5.2. Range-Energy relationships for BNCT products (H, He, Li and C) in ICRU muscle and Silicon. The right-hand side shows the same plot as the left except for the silicon range scaled by 1/0.63. In addition, the plots display original SRIM generated data points used for the 5th order polynomial fit.

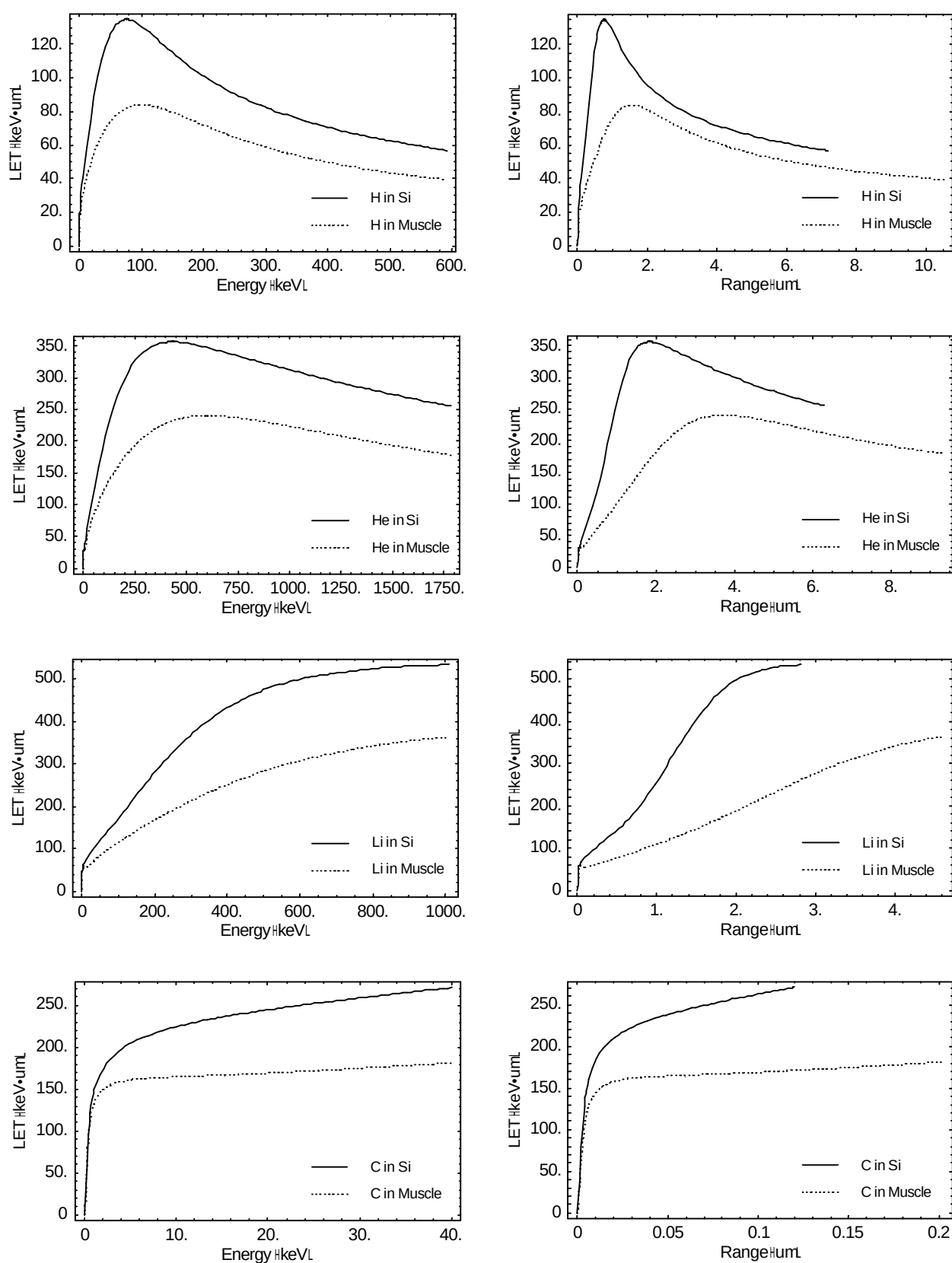


Figure 5.3. LET versus energy (left) and LET versus range(right) for BNCT Products (H, He, Li and C) in ICRU muscle and Silicon. The maximum range is plotted only up to the corresponding maximum energy.

5.4.2 Comparison of Silicon and Tissue BNCT Energy Deposition Spectra

5.4.2.1 Method: Monte Carlo Program

A Monte Carlo program was developed in order to confirm that the energy deposition in appropriately scaled silicon volumes may approximate the deposition in tissue. The use of Monte Carlo methods to calculate microdosimetric spectra in cell sized volumes exposed to BNCT and other high LET ions is well established [186]. It is generally assumed that the path of the ions follow straight lines with negligible straggling and that the width of the track is negligible in comparison to the volume size. These assumptions are valid for micrometer sized volumes and low energy ions as confirmed by comparative calculations performed by Charlton [187] on detailed track structure versus simple chord length/LET calculations.

The program description begins by defining two types of volumes:

1. *Generation volume (GV)* in which ions are generated assuming a nitrogen concentration of 3.5 g/100g and a boron concentration of 20 $\mu\text{g/g}$. These are typical values as used by Charlton [186]. The probability of formation of the various ion pairs and the number of interactions per μm^3 are given in Table 5.4.
2. *Sensitive volume (SV)* in which we compute the energy deposited by the traversal of ions generated in the GV.

The shape of both volumes is defined as an RPP with dimensions $x \times y \times z$ all in μm . Such a shape is selected on the basis that the SV of silicon reverse biased p-n junctions has traditionally been modeled as an RPP (See chapter 3 for a complete discussion on detector shape). The position of the GV with respect to the SV has no restriction with partial or even complete overlapping allowed.

The Monte Carlo process may be summarized by the following steps:

1. Randomly select an ion pair with initial energy and range given by Table 5.3 according to the probability specified in Table 5.4.
2. Randomly select the position of the interaction within the GV and the angle of ion emission with equal probability for all points within the GV and all angles. Note that ion pairs are emitted at 180 degrees.

3. Calculate the points of intersection of the emitted ions with the SV. This may be approached by solving the simultaneous equations of a line intersecting an RPP in a similar manner as Charlton [188] for an ellipsoid. Alternatively, one may incrementally follow the ion path at sufficiently small increments ($0.01\mu\text{m}$), testing at each increment to check for traversal of the SV boundary. The latter approach was adopted despite the longer computation time since it is more adaptable to future versions that may operate on complex geometry.

The main Monte-Carlo routines were coded in C for computational speed. These routines were encapsulated within a Mathematica [141] based input/output framework to provide flexibility in program input and output analysis.

Software validation of such programs is a non-trivial task since it is often difficult to construct cases of sufficient simplicity that analytic solutions may be derived for result comparison. The software was modified slightly to calculate segment length distributions. Such distributions have been analytically derived by Bradford [122, 137] (based on the work of Kellerer [128]) for an RPP volume under the condition of μ -randomness and fixed length tracks as discussed in chapter 3. Exact segment length distributions were calculated using the Mathematica notebook shown in Appendix B (and described in chapter 3) and compared against the Monte Carlo results. Figure 5.4 displays the close correspondence between the analytic and Monte Carlo derived distributions thus validating a substantial portion of the software. The remainder of the software was tested by thorough review and testing of individual functions.

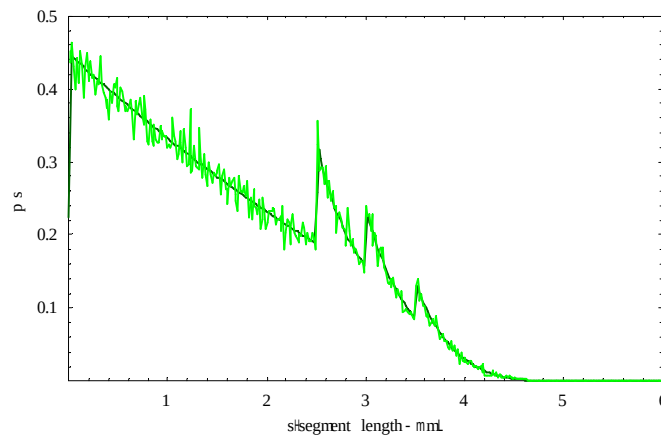


Figure 5.4. Comparison of analytic and Monte Carlo generated segment length distribution ($P(s)$) for an RPP volume $3.5 \times 3 \times 2.5 \mu\text{m}$ enclosed in a uniform isotropic field of $7 \mu\text{m}$ length rays.

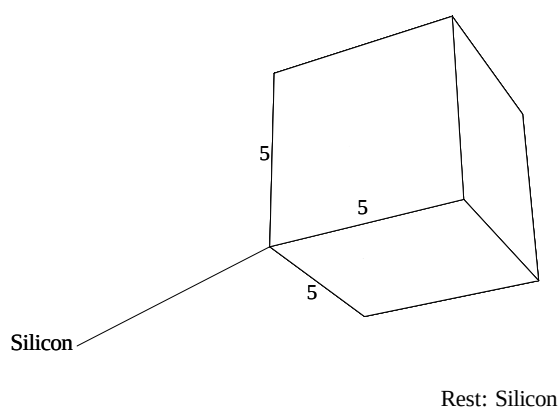


Figure 5.5. Geometry Case 1 - SV and GV are identical in size and position

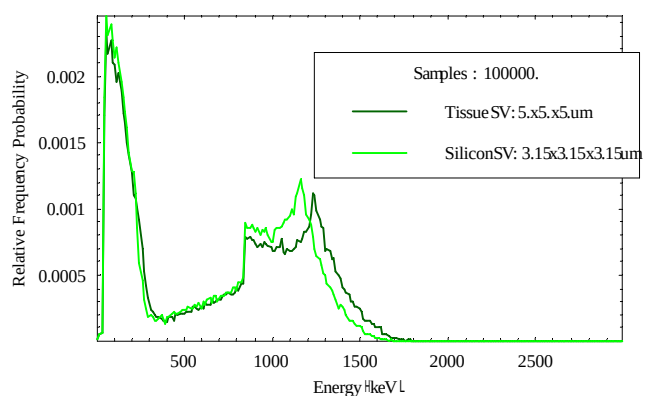


Figure 5.6. Geometry Case 1 (Example 1):
Energy Deposition Spectra comparison of Silicon ($3.15 \times 3.15 \times 3.15 \mu\text{m}$) and Tissue ($5 \times 5 \times 5 \mu\text{m}$).

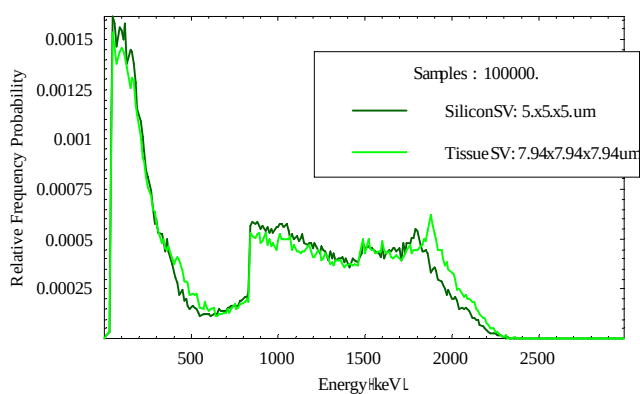


Figure 5.7. Geometry Case 1 (Example 2):
Energy Deposition Spectra comparison of Silicon ($5 \times 5 \times 5 \mu\text{m}$) and Tissue ($7.9 \times 7.9 \times 7.9 \mu\text{m}$).

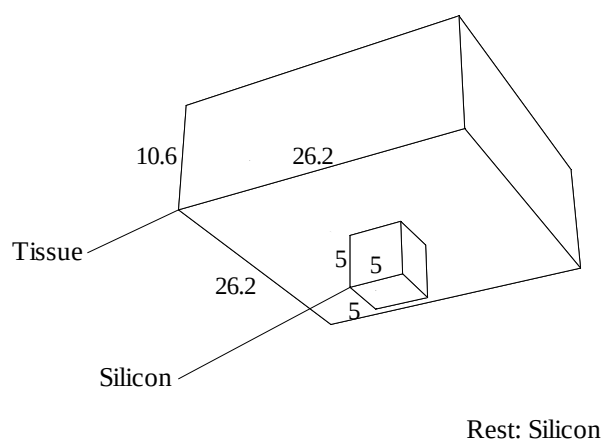


Figure 5.8. Geometry Case 2 - GV is a large tissue volume above a much smaller silicon/tissue SV

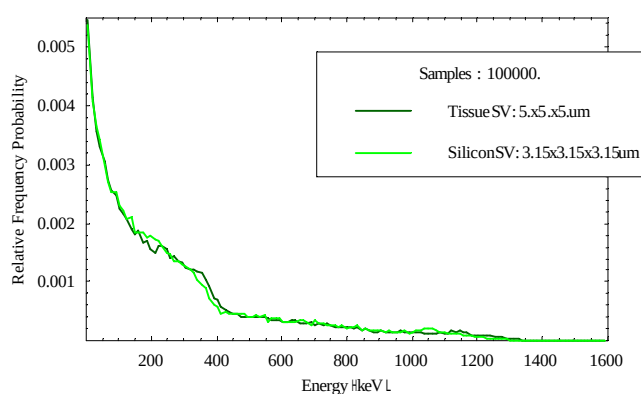


**Figure 5.9. Geometry Case 2 (Example 1):
Energy Deposition Spectra comparison of Silicon ($3.15 \times 3.15 \times 3.15 \mu\text{m}$) and Tissue ($5 \times 5 \times 5 \mu\text{m}$).**

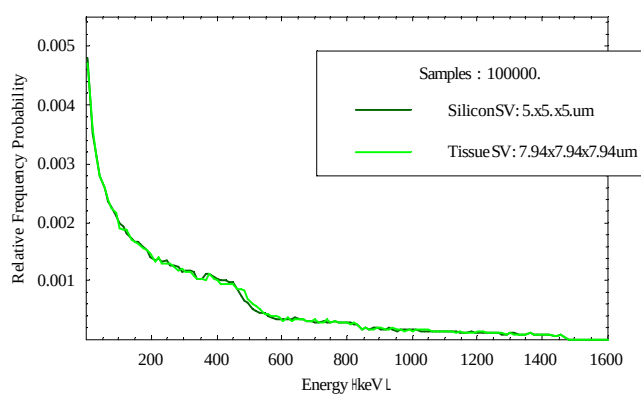


**Figure 5.10. Geometry Case 2 (Example 2):
Energy Deposition Spectra comparison of Silicon ($5 \times 5 \times 5 \mu\text{m}$) and Tissue ($7.9 \times 7.9 \times 7.9 \mu\text{m}$).**

5.4.2.2 Results: Geometry Case 1 - Ion generation in RPP volumes with SV=GV.

Geometry Case 1 consists of identical SV and GV of the same material (silicon or tissue). Figure 5.5 illustrates the case for a $5 \times 5 \times 5 \mu\text{m}^3$ volume of silicon. We assume that both silicon and tissue contain the same concentration of boron and nitrogen. This case corresponds to the BNCT scenario in which a B-10 compound is uniformly distributed throughout the cell. However, a corresponding silicon detector cell is limited to measuring only the boron capture component. Boron may be easily introduced by using *p* doped Si arrays as proposed by Rosenfeld [71]. The introduction of nitrogen into the body of a silicon diode is a much more difficult proposition. Nitrogen concentrations required for an acceptable probability of nitrogen capture are too high for maintaining device operation. Note, that nitrogen may be introduced into surrounding insulating layers (e.g. Si_3N_4 is commonly used for passivation). Nevertheless, for the purpose of testing the silicon/tissue scaling factor, we compare a hypothetical device assuming tissue concentrations of nitrogen and boron.

Figure 5.6 and Figure 5.7 illustrate the energy deposition spectra (normalized) for silicon and tissue with the dimensions of silicon scaled by 0.63 in each case. Clearly, the energy deposition spectra are vastly different for similar volumes of tissue and silicon. However, a comparison of silicon volumes scaled by $z = 0.63$ with tissue volumes indicates quite good correspondence between spectra. Note that the step in probability at 840 keV corresponds to the lithium ion as given in Table 5.3. This ion has a range small enough to be a complete insider. A similar step occurs at 40 keV corresponding to the carbon recoil.

5.4.2.3 Results: Geometry Case 2-Ion generation above RPP volume

Geometry Case 2 consists of a large GV placed directly on top of a much smaller SV. The GV material is tissue whilst the remaining volume including the SV consists of either tissue or silicon. This case is not hypothetical since it is easy to construct an overlayer with the appropriate concentrations of boron and nitrogen without compromising silicon device operation. Figure 5.8 illustrates the case for a silicon SV of dimensions $5 \times 5 \times 5 \mu\text{m}^3$. The GV dimensions are selected such that the boundaries are further from the SV than the longest range ion (590 keV proton -10.5 μm). Thus, we effectively model a volume of infinite extent away from the SV. Such a situation models a layer of tissue equivalent plastic (or real cells) impregnated with appropriate boron concentrations and placed above a silicon cell (within an integrated circuit). We

then compare the energy deposition in an appropriately scaled silicon cell to that which may occur if a tissue cell was substituted as the SV. Again the normalized results shown in Figure 5.9 and Figure 5.10 confirm that, with appropriate geometrical scaling, silicon detectors with well known geometry will record energy deposition spectra representative of tissue cells of equivalent shape. Again, note the significant differences in spectra between the two figures (tissue $5 \times 5 \times 5 \mu\text{m}^3$ in Figure 5.9 and silicon $5 \times 5 \times 5 \mu\text{m}^3$ in Figure 5.10) indicating the importance of using an appropriate scaling factor.

5.5 Fast Neutron Therapy and Proton Therapy Tissue Equivalence

Experimental microdosimetry is commonly applied to the characterization of fast (>1 MeV) neutron and proton beams. The conditions necessary for TE of a silicon/TE converter microdosimeter were summarized in section 5.3. Each of these conditions will now be addressed for the case of fast neutron therapy (FNT) and proton therapy (PT). The emphasis will be on FNT since for PT the absorbed dose for proton energies less than 250 MeV is dominated by electronic interactions with the protons. The recoil products in PT are similar in energy and type (but not relative contribution) to FNT so the discussion on TE of FNT is largely applicable to PT. Further discussion of neutron therapy and proton therapy are provided in chapter 8 and chapter 9, respectively.

5.5.1 Geometrical TE scaling factor for FNT and PT

The main ion products from fast neutron reactions include protons, alphas, and heavy ion recoils (carbon, oxygen recoil and to a lesser extent nitrogen). As an example, the initial spectrum of charged particles for 14 MeV neutrons in ICRU tissue was calculated by Caswell and Coyne [125] and shown in Figure 5.11. From this data for 14 MeV neutrons; protons have energies up to 14 MeV, alphas may have energies up to about 13 MeV, carbon up to 6 MeV and oxygen up to 3 MeV. The range-energy relationships for these ions are shown in Figure 5.14 and the LET relationships are given in Figure 5.15. In addition, the range-energy relationship for fast protons is displayed up to 200 MeV in Figure 5.13. The mean ratios of ranges in tissue and silicon, calculated as per equation (5.12), are displayed in Table 5.6. In general, the previously calculated scaling factor of 0.63 is reasonable, although the mean ratio drops to 0.58 for high energy protons. For microdosimetric spectrum dominated by high energy protons, one may consider reducing the geometric scaling factor. Note, that in

many cases the higher LET ions, although relatively infrequent, contribute significantly to the dose mean lineal energy. Additional weighting should then be applied to such ions when estimating the optimal scaling factor.

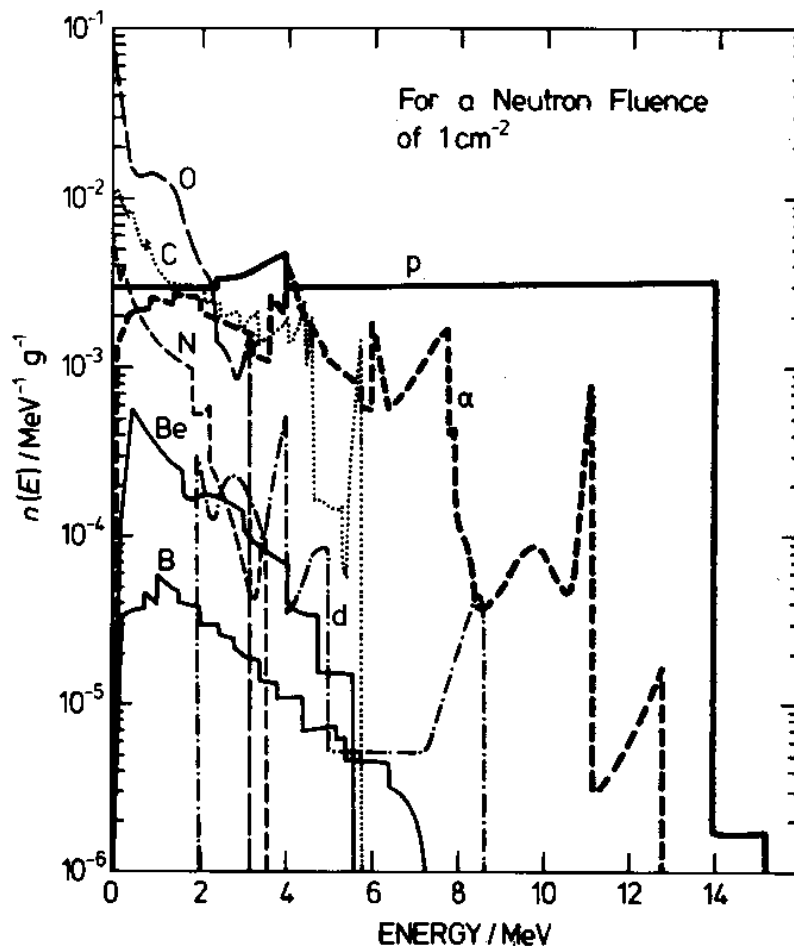


Figure 5.11. Initial spectrum of charged particles for a neutron fluence of 1 cm^{-2} of 14 MeV neutrons in ICRU tissue (from Caswell and Coyne [125] and ICRU report 36 p39 [11]). The symbols p, d and α refer to recoil protons, deuterons and alpha particles, respectively,

Table 5.6. Mean ratio of Fast Neutron/Proton Therapy ion ranges in silicon and tissue

Ion	z
H (0-14 MeV)	0.62
H (0-200 MeV)	0.56
He	0.66
C	0.63
O	0.61

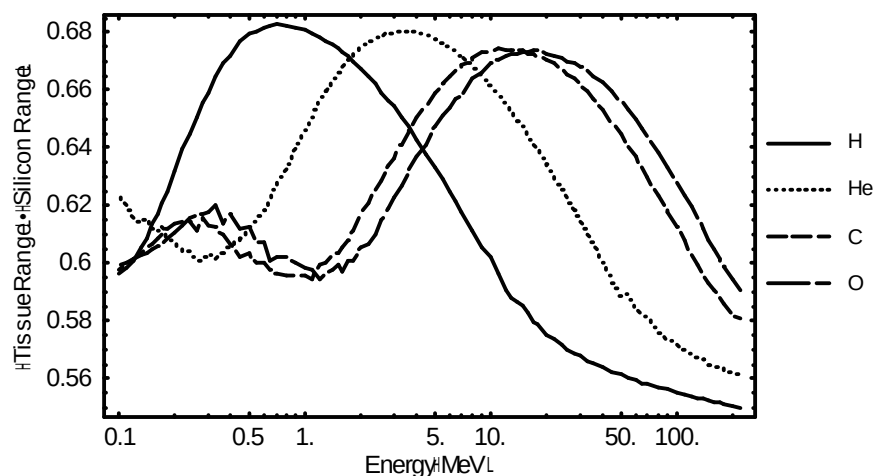


Figure 5.12. Ratio of tissue range over silicon range as a function of ion energy for main ions in Neutron and Proton Therapy.

The ratio of tissue range over silicon range as a function of ion energy is shown for various ions up to 200 MeV in Figure 5.12. This provides a useful reference for calculating the required z factor given information on the energy spectrum of secondary recoil products. Curves for different ions follow a common shape shifted to the right according to the ion's atomic number. The Bragg peak in the LET curve for each ion shifts to higher energies as the atomic number rises, although the main peak in Figure 5.12 is at an energy somewhat higher than the Bragg peak for each ion. An empirical equation should be extractable from Figure 5.12, with ion atomic number as a parameter, although this was not attempted.

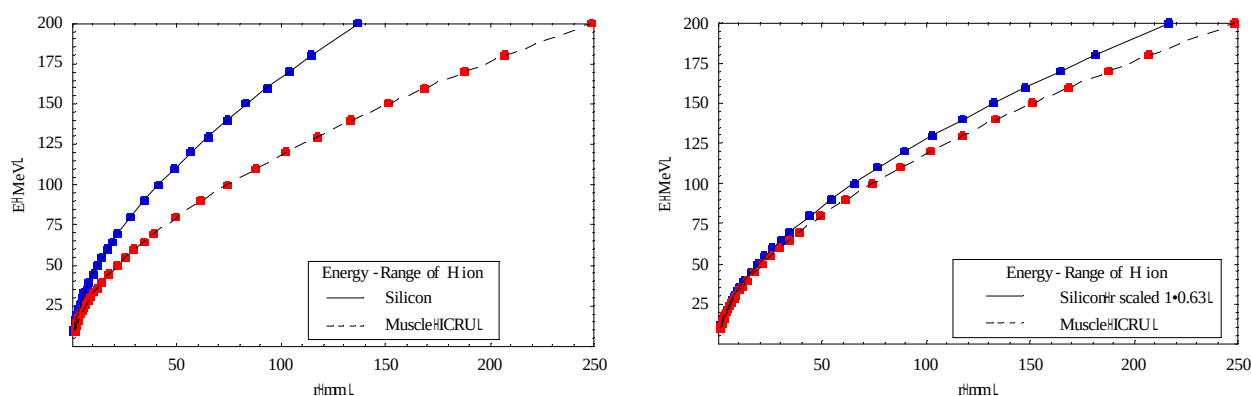


Figure 5.13. Range-Energy relationships for fast protons in ICRU muscle and Silicon. The right-hand side shows the same plot as the left except for the silicon range scaled by $1/0.63$.

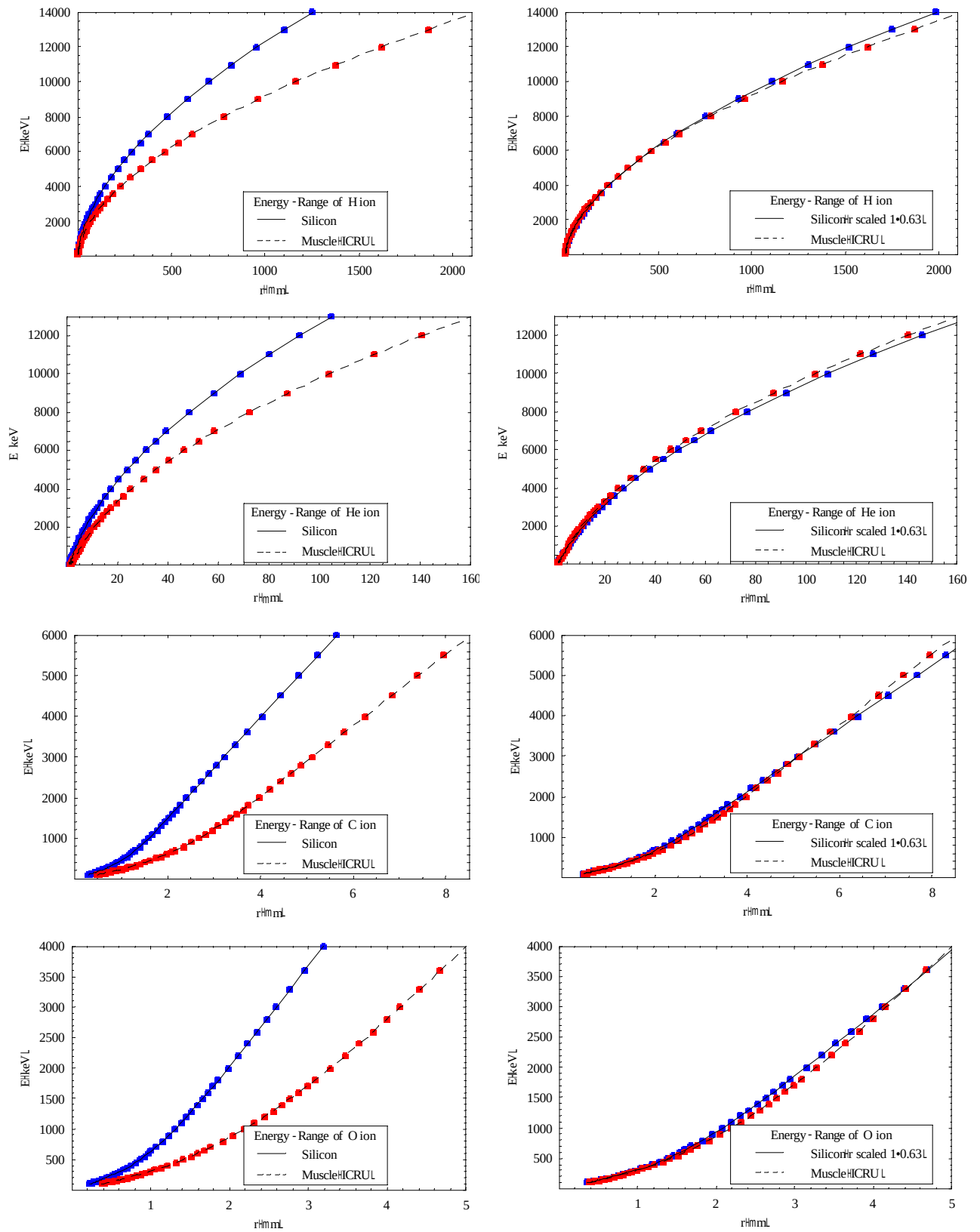


Figure 5.14. Range-Energy relationships for the main reaction products of fast neutron beams (H, He, C and O) in ICRU muscle and Silicon. The right-hand side shows the same plot as the left except for the silicon range scaled by $1/0.63$.

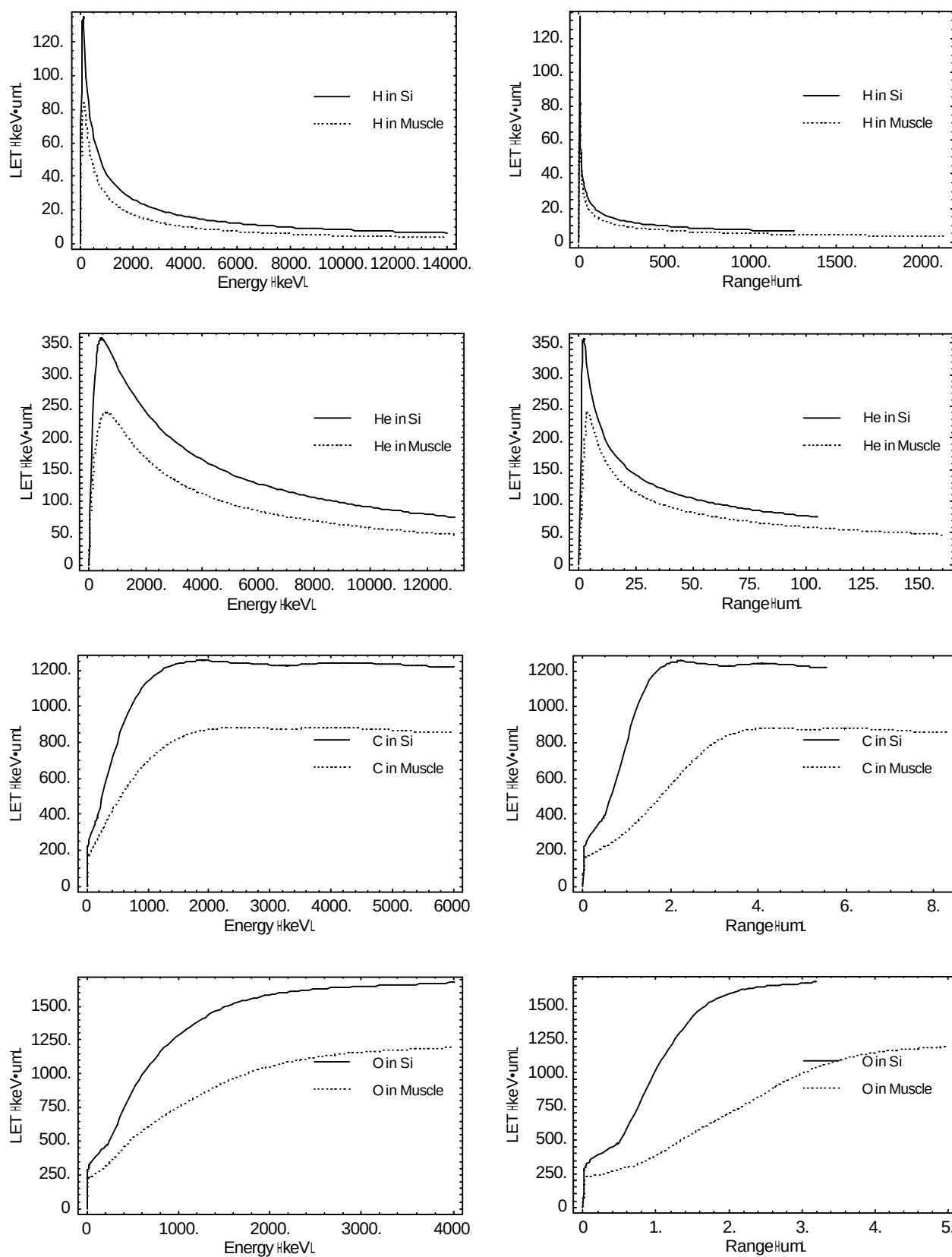


Figure 5.15. LET versus energy (left) and LET versus range (right) for Fast Neutron Therapy Products (H, He, C and O) in ICRU muscle and Silicon. The maximum range is plotted only up to the corresponding maximum energy.

5.5.2 Effect of converter atomic composition in FNT

Given that the TE geometrical scaling factor $z = 0.63$ is a reasonable approximation for a wide range of ion types and energy, one must now consider the production of secondary particles in a fast neutron environment. At neutron energies below 10 MeV the bulk of absorbed energy is due to hydrogen recoils except for some narrow resonances in heavy elements. The proportion of hydrogen in muscle is well approximated by A150 and 20% lower than desired in Lucite. Both A150 and Lucite substitute carbon for oxygen in their atomic composition when compared with the desired muscle composition as shown in Table 5.1. At higher neutron energies elastic and inelastic collisions with carbon and oxygen become important and appreciable ($>20\%$) TE induced errors in $\bar{y}_d$ may occur. Table 5.7 gives values of $\bar{y}_d$ for various substances calculated theoretically by Green [189]. Furthermore, calculations by Caswell and Coyne [190] have shown considerable differences in the lineal energy spectra for 20 MeV neutrons between a simulated A150 proportional counter and ICRU muscle as shown in Figure 5.16.

Table 5.7. Calculated values of $\bar{y}_d$ in A150 and Muscle for various neutron energies E_n

Material	$E_n=5$ MeV	$E_n=10$ MeV	$E_n=15$ MeV	$E_n=19.5$ MeV
A150	53.8	73.4	101.1	124.7
Muscle	49.3	60.3	68.9	64.7

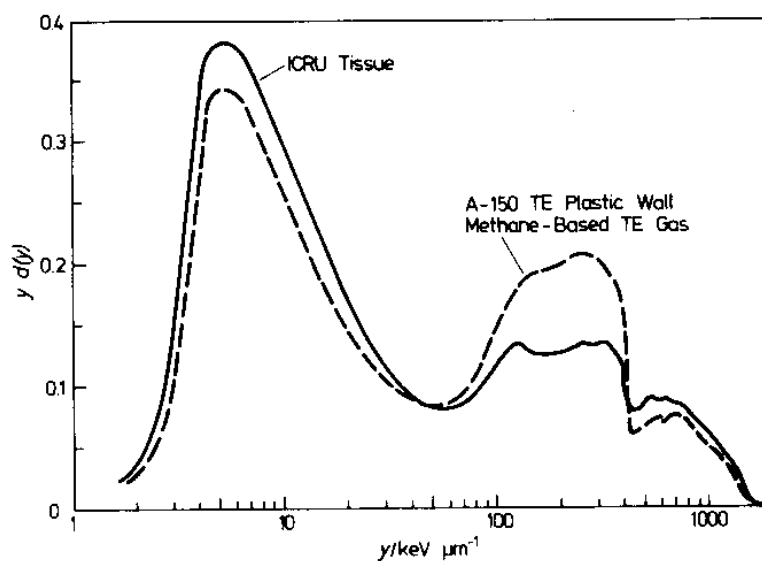


Figure 5.16. Comparison of analytic calculations of $yd(y)$ of 20 MeV neutrons performed for ICRU tissue and for A150 plastic. (from Caswell and Coyne [190] and ICRU report 36 p42 [11])

Aside from simulation comparisons, a first order comparison of tissue equivalent performance may be gauged by comparing kerma ratios. Kerma is defined as the initial kinetic energy of all charged particles liberated by the radiation per unit mass and has the unit of Grays. Kerma data for various materials are tabulated in ICRU report 44 [32] based on the work of Caswell and Coyne [191]. Data is readily available for Acrylic, A150 and Muscle (Skeletal-ICRU 44). Acrylic ($C_6H_4O_2$, $\rho = 1.17 \text{ gm/cm}^3$, H = 7%, C = 56%, O = 37%) is quite close in composition and density to Lucite ($C_5H_8O_2$, $\rho = 1.2 \text{ gm/cm}^3$, H = 8%, C = 60%, O = 32%) and will be used in this comparative study. Note that the Acrylic mass composition provided in Appendix A of ICRU report 44 is fortunately in error since it corresponds to Lucite's mass composition. As mentioned in section 5.1, Muscle (Skeletal-ICRU 1989) is quite close in composition to Muscle (Striated-ICRU 1964) and sufficient for this comparative study.

The kerma ratio of A150 and Lucite to muscle (Skeletal, ICRU 1989) is shown in Figure 5.18. The Lucite kerma is generally 20% lower than A150 due to the 20% lower hydrogen content. The dips at 0.4 and 1 MeV occur due to oxygen elastic reactions, which are present in muscle and not in the plastics due to their lower oxygen content. These reactions have several resonances in the cross-section above 0.4 MeV as shown in Figure 5.22. The relative contribution of each element to tissue kerma is shown in Figure 5.17. Above 10 MeV the ratio rise is due to the effect of increasing carbon kerma contribution from the plastics.

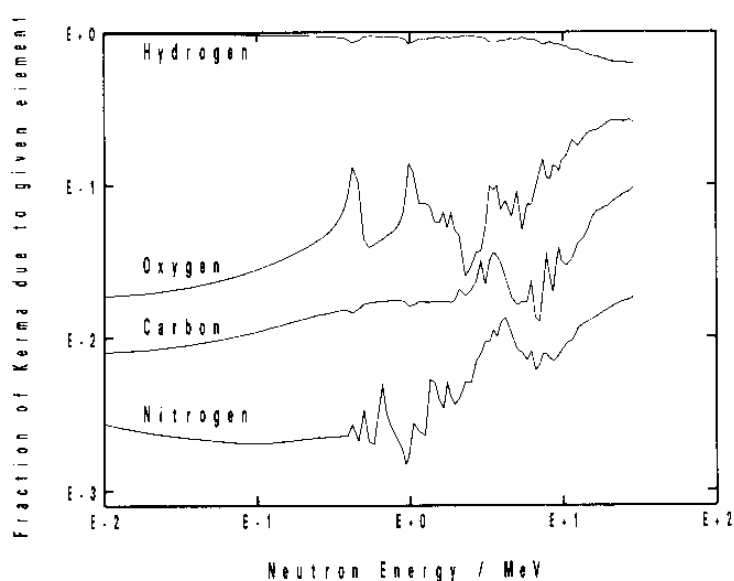


Figure 5.17. Relative contribution from interaction processes with different elements to kerma in muscle (skeletal) over the neutron energy interval 10 keV-30 MeV. (From p9 ICRU report 44 [32])

The final microdosimetric spectrum is normalized to unity since it is a probability distribution. Therefore in comparing kerma it is reasonable to normalize to unity at a low energy (where hydrogen interactions dominate) as shown in the right hand side of Figure 5.18. The normalization is slightly more complicated in that we only normalize the percentage of the kerma due to hydrogen interactions as provided by Figure 5.17. Deviations in the normalized kerma ratio from unity will crudely translate to differences in the normalized microdosimetric spectrum, particularly at higher lineal energies where the carbon and oxygen recoil contribute. Viewed in this manner, for neutron energies below 30 MeV, Lucite is a better approximation to tissue than A150 since its carbon and oxygen composition is better than A150. This is reflected in a normalized kerma ratio for Lucite which is closer to 1 than the A150 ratio.

A final important comment evolves from this discussion. The silicon/converter microdosimeter offers a potential tissue equivalent advantage over the A150 based proportional counter. Specifically, the silicon microdosimeter converter may be constructed using actual tissue cells or materials with precise muscle equivalent composition without regard for electrical conductivity, mechanical and construction requirements. This approach was not investigated in this work but is highly recommended for future consideration. A multi-layer converter structure may prove useful and amenable to construction. A thin layer of tissue cells may be sandwiched between TE plastic and the detector below. The longer range protons may primarily originate from the plastic whilst oxygen and carbon recoils originate from the cells. Each layer has the correct atomic composition for recoil products that are likely to intercept the detector.

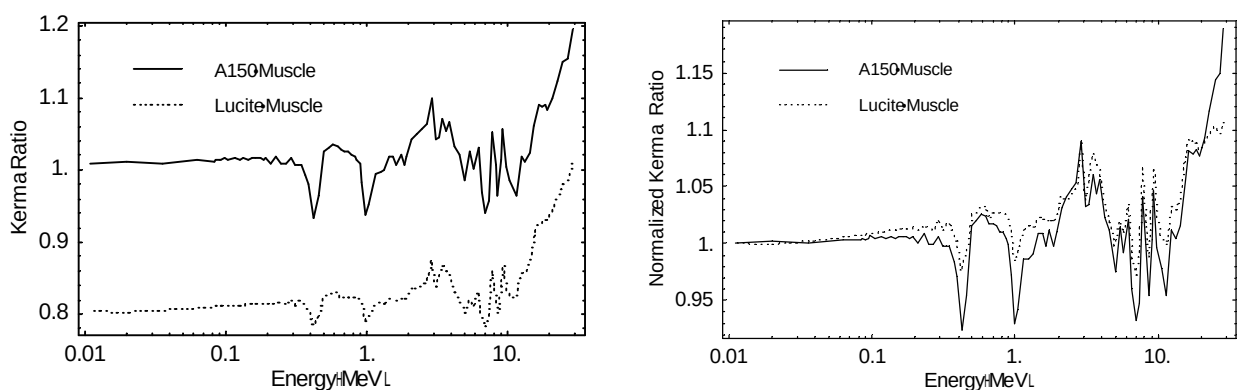


Figure 5.18. Kerma Ratio of A150 and Acrylic to Muscle (Skeletal, ICRU 1989) as a function of neutron energy. The right hand side plot has both kerma ratios normalized to 1 at 0.01 MeV (see text for explanation). The left hand side is unnormalized.

5.5.3 Silicon versus tissue interactions in FNT and PT and range of secondary products.

To maintain tissue equivalence the silicon microdosimeter must generate a relatively low proportion of insiders and starters compared with the tissue converter crossers and stoppers. Therefore, a brief summary of neutron-silicon and proton-silicon interactions is appropriate, followed by an analysis of relative kerma with tissue.

The important nuclear interactions of neutrons with silicon are summarized as

- $\text{Si}^{28}(\text{n},\text{n})\text{Si}^{28}$ elastic scattering
- $\text{Si}^{28}(\text{n},\text{n}')\text{Si}^{28}$ inelastic scattering
- $\text{Si}^{28}(\text{n},\text{p})\text{Al}^{28}$ nuclear reaction
- $\text{Si}^{28}(\text{n},\alpha)\text{Mg}^{25}$ nuclear reaction $Q=-2.654$ MeV

For neutron energies above a few MeV, nuclear reactions begin to contribute to the silicon kerma. Nevertheless, elastic scattering dominates the total cross-section (~ 2 barns) for energies in the tens of MeV range. The silicon recoil energy is often small, with the maximum recoil energy from an elastic scatter reaction occurring for a backward scatter event. In this special case, the recoil energy is $0.133 E$, where E is the nucleon incident energy [192]. Small recoil energies are much more probable and these result in much lower silicon recoil energies.

Similar nuclear reactions also occur for protons incident on silicon. The characteristics of high-energy proton-nucleus and neutron-nucleus reactions are very similar above 100 MeV. Below 100 MeV, the proton cross-section for silicon interactions is slightly lower than for neutrons. From about 20 MeV or below, the proton reaction cross section decreases rapidly because of the proton-nucleus Coulomb barrier. This barrier, however, does not apply to neutrons.

In general, microdosimetric measurements in proton therapy are only important at the end of the proton range. In this region, the relevant interactions correspond to no more than the tens of MeV energy range of fast neutron therapy. Therefore, we restrict the remainder of this comparative study to fast neutron therapy with typical neutron energies of less than 30 MeV.

We may consider the kerma ratio with tissue to ascertain the relative probability of silicon interactions as shown in Figure 5.19. The kerma ratio of silicon to tissue is less than 0.05 for neutron energies less than 5 MeV. Above this value, the cross-section for silicon neutron interactions rises more rapidly than for tissue resulting in the kerma ratio increasing to 0.2 at 15 MeV. The total cross-section (elastic + reaction(inelastic)) is high for silicon in the tens of MeV range, typically about 2 barns, with the inelastic reaction cross-section being 40% of the total cross-section as shown in Figure 5.22 and Figure 5.23.

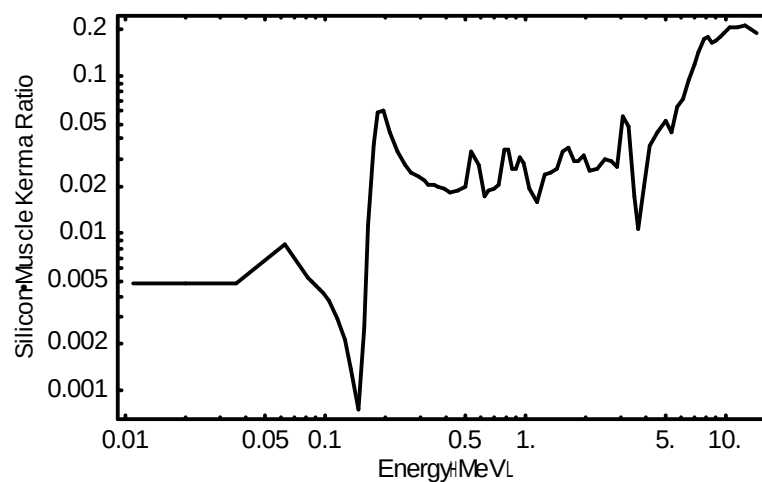


Figure 5.19. Kerma Ratio of Silicon to Muscle (Skeletal, ICRU 1989) as a function of neutron energy .

Some useful data is provided by Schrewe [37] in a very recent and excellent paper on the experimental estimation of kerma coefficients and dose distributions in various materials (C, N, O, Mg, Al, Si, Fe, Zr, A-150 plastic, Al_2O_3 , AlN, SiO_2 and ZrO_2) for neutron energies up to 66 MeV. Proportional counters of various wall construction were used to measure dose in mono-energetic neutron fields which under charged particle equilibrium (CPE) conditions is almost equal to kerma. More precisely, the kerma coefficients (k_F or $k = \text{kerma/fluence}$) were derived by the following sequence: measure the absorbed dose in the cavity gas of the microdosimeter, derive the absorbed dose in the wall using Bragg-Gray cavity theory, determine the kerma in the wall under CPE conditions, and then correct for the neutron fluence attenuation in the wall. The kerma coefficients for materials in the silicon microdosimeter and their ratio to A150 plastic are shown in Table 5.8.

Table 5.8. Neutron kerma coefficients (k in fGy m²) for Si microdosimeter materials and ratio to A150 from Schrewe [37]. The ratio of Si to A150 kerma is shaded for clarity.

Material	5 MeV	8 MeV	15 MeV	17 MeV	34 MeV	44 MeV	66 MeV
A150 (k)	4.546	5.364	6.732	7.505	8.67	8.07	7.95
(k/k_{A150})	1	1	1	1	1	1	1
Si (k)	0.175	0.812	1.436	1.571	2.75	3.05	3.84
(k/k_{A150})	0.038	0.151	0.213	0.209	0.32	0.38	0.48
SiO ₂ (k)	0.406	0.757	1.914	2.048	3.62	3.99	4.77
(k/k_{A150})	0.089	0.141	0.284	0.273	0.42	0.49	0.60
Al (k)	0.196	0.438	1.140	1.436	2.40	2.82	3.82
(k/k_{A150})	0.043	0.082	0.169	0.191	0.28	0.35	0.48

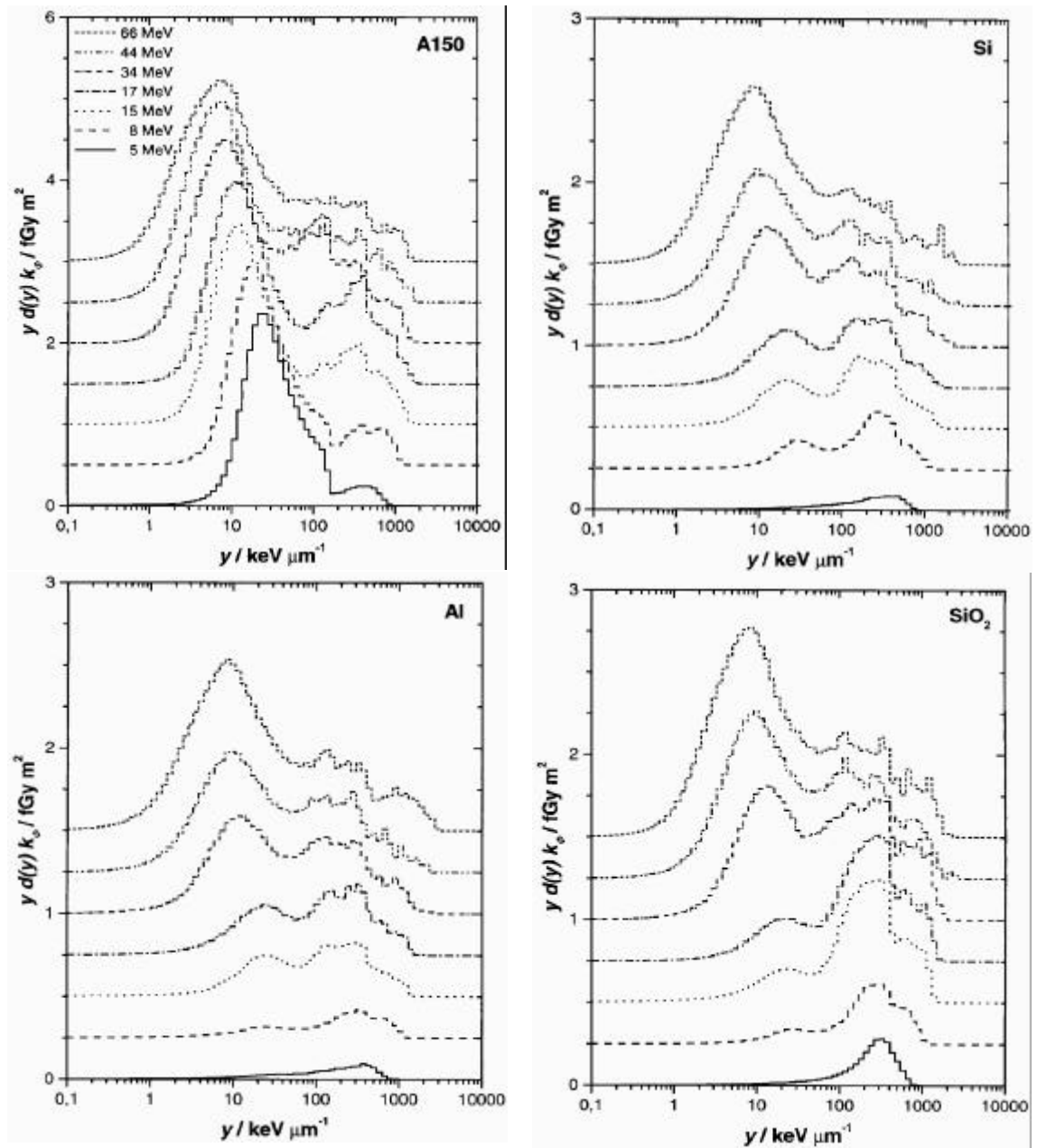


Figure 5.20. Dose distributions $d(y)k$ for A150, Si, Al and SiO₂ at neutron energies of 5, 8, 15, 17, 34, 44 and 66 MeV (from Schrewe [37]). Note, to improve presentation, fixed offsets were added. In addition, take care to consider the different scale for the A150 plot.

The values of tissue to silicon kerma derived from Schrewe's data [37] and given in Table 5.8 are in good agreement with Figure 5.19 although they extend the ICRU data to several higher energies. The kerma ratios with A150 for the main materials present in the detector overlayer (Al, SiO₂) are similar to Si. The Al ratio is generally lower by a few percent and the SiO₂ kerma ratio with A150 is 20-25% higher than Si at neutron energies above 15 MeV due to inelastic reactions with oxygen.

Schrewe [37] also provides dose distributions $d(y)k_F$ weighted by the kerma coefficient for measurements made using microdosimeters constructed with various materials and filled with a tissue equivalent propane gas mixture to provide an equivalent site diameter of 2 μm . These distributions, as shown in Figure 5.20, enable an approximate estimation of the expected contribution from different elements in the device construction to the silicon/converter microdosimeter dose distribution. Consider a hypothetical device with equal volumes of A150 and Si contributing to the detected secondary charged particle. This may be a reasonable first order approximation to the current device with an A150 converter located above the detector. At low neutron energies (<8 MeV) the contribution from silicon to the dose distribution is relatively small as reflected in the kerma ratio and visual comparison of the dose distributions. As the neutron energy increases above 8 MeV the contribution to the dose distribution increases particularly for lineal energies greater than 100 keV/ μm . Given that as much as 30% of the dose may be due to silicon interactions at these lineal energies then one may expect discernible spectral distortion at neutron energies between 8 and 34 MeV. Clearly, the minimisation of silicon versus converter volume is required, a point that will be discussed later in this section. Interestingly, at high neutron energies >34 MeV, the distortion will be less significant since the shape of the dose distribution for both a silicon and A150 device are quite similar, particularly at 66 MeV. This behaviour arises from the decrease in significance of hydrogen reactions and the increased importance of inelastic reactions with sufficiently similar secondary product deposition spectra in silicon and A150.

An alternative analysis based on cross-section data will now be presented. Given appropriate cross-section data the probability of an interaction with silicon and tissue may also be compared. For sufficiently thin volumes the probability P_c of an incident particle encountering a nuclear collision is estimated by:

$$P_c \cong r s t \quad (5.13)$$

where r is the atomic density (in atoms/cm³), s is the cross section (in cm²) and t is the thickness (in cm). Total interaction cross-section data for silicon and the important elements of tissue are shown in Figure 5.22 with the inelastic (reaction) cross-sections compared in Figure 5.23. The hydrogen total cross-section is relatively high for energies less than about 8 MeV. Above 10 MeV, the cross-section magnitudes are in order of atomic number with silicon having the highest cross section.

As an example, the relevant data for 14 MeV neutrons is provided in Table 5.9 along with the probability of interaction for a particle traversing 1 cm of material. The probability of an interaction is similar in both silicon and tissue. However, 40% of the interactions in silicon are elastic collisions with poor energy transfer to the recoiling nucleus. The tissue, with its high hydrogen content, obtains high energy transfer due to the efficient energy exchange in elastic reactions with hydrogen. The difference in the energy of the secondary particles accounts for the higher kerma in tissue than in silicon.

The probability of interaction in 1 cm was calculated for energy ranges from 0.1 to 150 MeV in tissue and silicon. The ratio of these interaction probabilities for both total interactions and inelastic events is displayed in Figure 5.24. The total interaction ratio ranges from 0.2 to just over 1 in the energy range of most interest to fast neutron therapy. At neutron energies greater than 20 MeV the ratio steadily rises. For inelastic reactions, the high silicon cross-section produces a probability of inelastic reaction which is 2-3 times higher in silicon than tissue.

Nuclear reactions (inelastic) and elastic reactions with oxygen and carbon are of particular interest in microdosimetry since they produce recoil products with high LET. The recoil products typically have ranges of the order of a few microns as shown in Figure 5.14 and Figure 5.15. The alpha and proton by-products of inelastic reactions have longer ranges typically in the tens of micron range. In comparing silicon to tissue, the volume in which secondary particles are generated is important in assessing the probability that an excessive number of events will be generated from within the silicon compared to the tissue converter. To reduce the probability of silicon events, the silicon volume should be as small as practicable subject to the shape requirements discussed in chapter 3 and the practical signal to noise issues discussed in chapter 4. Accurate estimation of the maximum allowable silicon detector dimensions requires considerable

work in completely simulating (Monte Carlo) the interaction of the required neutron and proton environments with various tissue converter, overlayer and silicon detector constructions. Such time consuming calculations were not pursued in this work although they are highly recommended for follow-on studies. As an interim solution to gauge the extent of this problem we consider existing calculations for A150 proportional counters.

Fidorra and Booz [106] performed Monte Carlo simulations of a 5.85 MeV neutron beam incident on an A150 proportional counter with a simulated site size of 1 μm diameter. As shown in Figure 5.21 the proportion of starters and insiders within the tissue volume is quite small. The same volume of silicon would have even fewer starters and insiders since the ratio of interaction probabilities is 0.9. Thus, a silicon microdosimeter and TE converter with silicon dimensions of around 1 μm should provide a microdosimetric spectrum which is reasonably free of silicon interactions. The microdosimetric spectrum produced by thicker silicon detectors in fast neutron environments will have silicon interaction products contributing to spectral components at high lineal energies.

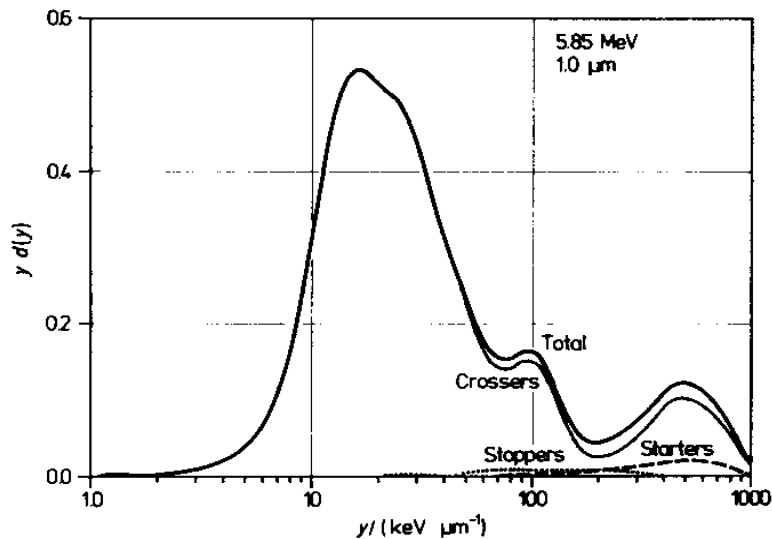


Figure 5.21. Monte Carlo calculations of $yd(y)$ of 5.85 MeV neutrons in TE A150 plastic and methane based TE gas (1 μm simulated diameter). Contributions of stoppers, starters, and crossers to the total. The contributions of insiders is insignificantly small. (from p42 ICR report 36 [11] and Fidorra and Booz [106]).

Complicating matters is the fact that the silicon microdosimeter is not completely surrounded by the TE converter, which is placed above the detector. For elastic collisions, this is not a significant problem since the neutron beam is generally normally incident to the detector. In this configuration, the majority of recoils will originate from

above the detector. Products from inelastic events are more isotropic and may originate from the silicon substrate. This may be a concern in silicon microdosimeters exposed to neutron energies around 10 MeV, where the inelastic cross-section is relatively high.

The implications of these results are significant for silicon microdosimetry design and experimentation. The silicon microdosimeter should be as small as practicable with the maximum amount of tissue equivalent material surrounding the detector.

Table 5.9. Nuclear interaction cross sections for 14 MeV neutrons incident on silicon and muscle.

Material	Nuclear Reactions		All Nuclear Interactions	
	$\sigma_{\text{inelastic}}$ at 14 MeV (barns)	P(interaction) in 1 cm	σ_{total} at 14 MeV (barns)	P(interaction) in 1 cm
Silicon	1.09	0.05	1.75	0.09
Muscle	Note 1	0.02	Note 1	0.1
Ratio(Si/Muscle)		2.5		0.9

Note 1: Muscle calculation uses elemental cross-sections from Figure 5.22 and Figure 5.23 and atomic density from Table 5.1.

Note 2: $\sigma_{\text{total}} = \sigma_{\text{elastic}} + \sigma_{\text{inelastic}}$

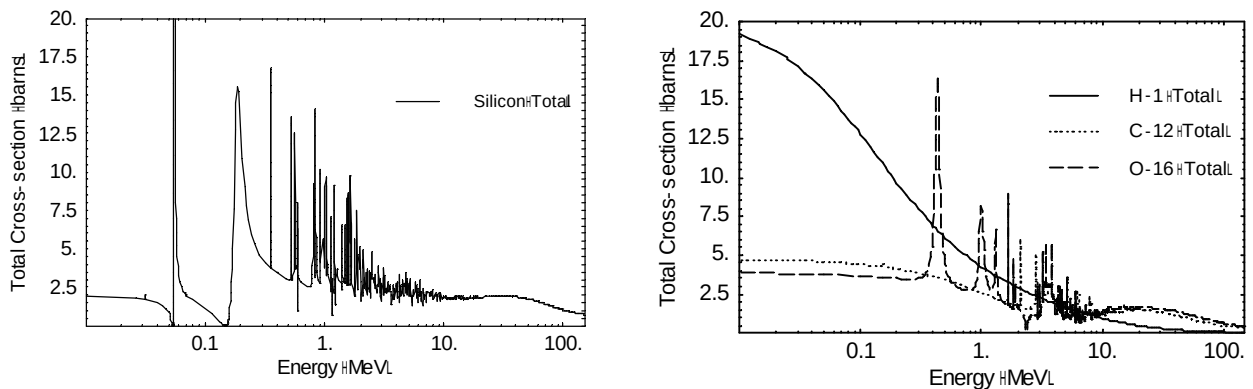


Figure 5.22. Total (elastic+inelastic) cross section for neutrons interacting with silicon (left side) and hydrogen, carbon and oxygen (right side). Si data is from Evaluated Neutron Data File, ENDF/B-VI library (>10MeV) and JENDL-3(300K) library (<10MeV). H,C and O data is from ENDF/B-VI library [193].

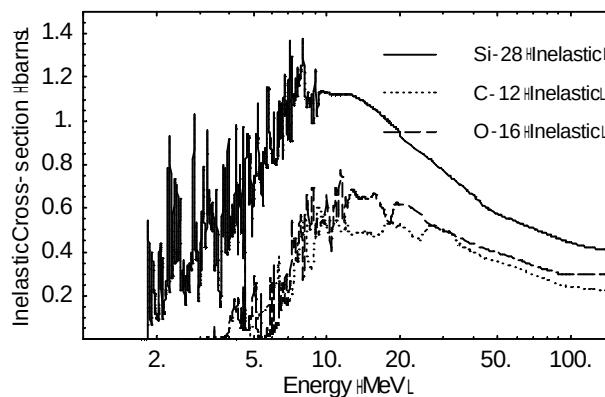


Figure 5.23. Inelastic (nuclear reaction) cross-section for neutrons interacting with silicon, carbon and oxygen. Data is from ENDF/B-VI library. [193]. Note that hydrogen, of course, does not undergo inelastic reactions.

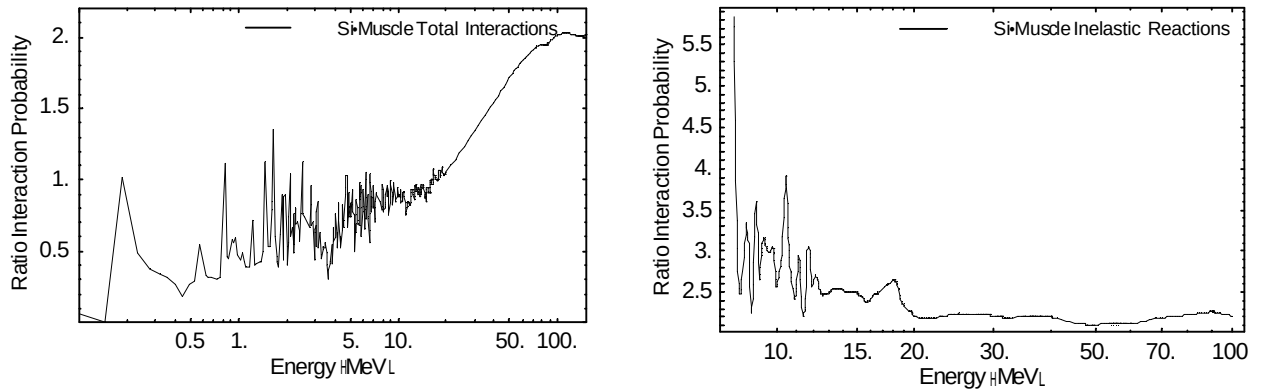


Figure 5.24. Ratio of probability of interaction between silicon and muscle as a function of neutron energy. Left side is for all interactions (elastic+inelastic); right side is for inelastic reactions only.

5.6 Summary of Tissue Equivalence Study

The requirements for TE were summarized in 5.3. To obtain tissue equivalence, the silicon detector/ TE converter should have a geometrical TE scaling factor of 0.63, a converter composition closely approximating tissue, and a minimum silicon detector size subject to shape and noise considerations (as discussed in chapters 3 and 4, respectively) in order to reduce the probability of nuclear interactions with silicon.

A Monte-Carlo program was developed to simulate the energy deposition of ions in a BNCT radiation field. The Monte Carlo results confirm that with appropriate geometrical scaling ($z = 0.63$) silicon detectors with well-known geometry will record energy deposition spectra representative of tissue cells of equivalent shape. That is, silicon is tissue equivalent for BNCT ion products under appropriate linear geometrical scaling. The scaling factor is also applicable to FNT but may need to be reduced for PT to around 0.56.

A brief study comparing tissue and silicon neutron interactions in FNT was described. For neutron energies less than 5 MeV, microdosimetric measurements in silicon volumes with dimensions of the order of 1 μm will not be significantly affected by events generated within the silicon. At higher neutron energies, the spectrum at high lineal energies may be affected by silicon elastic and inelastic reactions. A more detailed study is required to fully ascertain design requirements and the conditions under which satisfactory operation may be achieved. As a general guideline, the silicon microdosimeter should be as small as practicable with a maximum amount of tissue equivalent material surrounding the detector. The issue of tissue equivalence will be further discussed in the analysis of experimental results in subsequent chapters.