Monte Carlo investigation of energy response of various detector materials in $^{125}$I and $^{169}$Yb brachytherapy dosimetry

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Relative absorbed-dose energy response correction $R$ for different detector materials in water, PMMA and polystyrene phantoms are calculated using Monte Carlo-based EGSnrc code system for $^{125}$I and $^{169}$Yb brachytherapy sources. The values of $R$ obtained for $^{125}$I source are 1.41, 0.92, 3.97, 0.47, 8.32 and 1.10, respectively, for detector materials LiF, Li$_2$B$_4$O$_7$, Al$_2$O$_3$, diamond, silicon diode and air. These values are insensitive to source-to-detector distance and phantom material. For $^{169}$Yb source, $R$ is sensitive to source-to-detector distance for detector materials other than air and Li$_2$B$_4$O$_7$. For silicon, $R$ increases from 3 to 4.23 when depth in water is increased from 0.5 cm to 15 cm. For $^{169}$Yb source, the values of $R$ obtained for air and Li$_2$B$_4$O$_7$ in PMMA and polystyrene phantoms are comparable to that obtained in water. However, LiF, Si and Al$_2$O$_3$ show enhanced response and diamond shows decreased response in PMMA and polystyrene phantoms than in water.

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I. INTRODUCTION

Liquid water serves as reference medium for dosimetry of interstitial brachytherapy sources. Accurate dose measurements in the vicinity of individual brachytherapy sources are difficult. Ionization chambers are rarely used in brachytherapy dose measurements. This is because conventional ion chambers either lack the sensitivity to measure accurately the low levels of radiation emanating from individual sources, or are so large that dose gradients across the sensitive volume compromise spatial resolution. Solid state experimental dosimeters, such as silicon (Si) diodes, diamonds, Al$_2$O$_3$, LiF and Li$_2$B$_4$O$_7$ thermoluminescent dosimeters (TLD), provide the necessary sensitivity and spatial resolution but exhibit energy dependent responses which will vary with the source energy and with the position of the dosimeter within an absorbing medium.

Many studies are reported on dosimetric measurements around the $^{125}$I, $^{169}$Yb, $^{137}$Cs, $^{103}$Pd and $^{192}$Ir brachytherapy sources using the solid state detectors such as LiF, Li$_2$B$_4$O$_7$, Si diode, diamond, Al$_2$O$_3$. Relative diode response which is defined as the measured detector reading per measured unit of air kerma varies by 14%, 40%, and 75% with respect to source-detector distance in the presence of $^{137}$Cs, $^{169}$Yb and $^{192}$Ir sources, respectively. TLD-100 response has been shown to vary by 10% with respect to distance from a $^{192}$Ir source.

Monte Carlo techniques are widely used for calculating energy response corrections because of the ease with which the 3D geometry of the phantom, detector and source can be modeled. For example, Williamson et al., Piermattei et al. and Valicenti et al. employed the Monte Carlo-based MCPT code for calculating such correction factors. MacPherson et al.
used MCNP (version 4) Monte Carlo code to calculate absorbed-dose energy response correction for $^{169}$Yb source. Mobit et al.\textsuperscript{(22)} used the EGSnrc Monte Carlo system\textsuperscript{(24)} to calculate the corrections for LiF TLD rods for $^{125}$I sources.

According to TG43U1 update\textsuperscript{(1)} for brachytherapy dose measurements a well-characterized energy response function should be quantitatively accounted for. The objective of the present study is to calculate the relative absorbed-dose energy response correction as a function of depth in water for different solid-state detector materials for $^{125}$I (model selectSeed) and high-dose rate (HDR) $^{169}$Yb (model 4140) brachytherapy sources. The study also includes air as detector material representing an ionization chamber. Measurement of dose distribution is usually performed in ‘water-equivalent’ solid phantoms. The solid phantoms have the advantage that they can be precisely machined to accommodate sources and detectors, and distances can be accurately determined. Therefore, the study also includes calculation of relative absorbed-dose energy response correction for the investigated detector materials in solid phantoms polystyrene and polymethyl methacrylate (PMMA). We have employed the Monte Carlo-based DOSRZnrc and FLURZnrc user-codes\textsuperscript{(24)} of the EGSnrc code system\textsuperscript{(25)} in the present work.

\section{MATERIALS AND METHODS}

\subsection{Radioactive sources and detectors}

The geometric details and composition of the $^{125}$I selectSeed and $^{169}$Yb model 4140 are taken from the published studies.\textsuperscript{(26,27)} The photon energy spectra of the $^{125}$I and $^{169}$Yb sources needed for the Monte Carlo calculations are taken from literature.\textsuperscript{(1,27)} The detector materials investigated in the present study are LiF, Li$_2$B$_4$O$_7$, Al$_2$O$_3$, diamond, silicon and air. Table 1 presents the values of $Z_{\text{eff}}$ (effective atomic number), $\langle Z/A \rangle$ (electron density) and $\rho$ (mass density) of the investigated detector materials.

\begin{table}
\centering
\begin{tabular}{lccc}
\hline
Detector material & $Z_{\text{eff}}$ & $\langle Z/A \rangle$ & $\rho$ (g/cm$^3$) \\
\hline
Water & 7.5 & 0.555 & 1.0 \\
LiF & 8.27 & 0.462 & 2.635 \\
Li$_2$B$_4$O$_7$ & 7.4 & 0.485 & 2.44 \\
Al$_2$O$_3$ & 10.2 & 0.491 & 3.97 \\
Diamond & 6 & 0.496 & 3.5 \\
Silicon & 14 & 0.499 & 2.33 \\
Air & 7.6 & 0.499 & 0.0012 \\
\hline
\end{tabular}
\caption{Values of effective atomic number $Z_{\text{eff}}$, electron density $\langle Z/A \rangle$, and mass density $\rho$ of the detector materials studied.}
\end{table}

\subsection{Energy dependence of the detector}

The energy dependence of the detector may be separated into two components.\textsuperscript{(28,29,30)} One function, called the intrinsic energy-dependence, $k_{\text{in}}(Q)$, relates the detector output, $M_{\text{det}}(Q)$, to the average dose to the material of the sensitive detector element, $D_{\text{det}}(Q)$, as a function of beam quality, $Q$.

$$D_{\text{det}}(Q) = k_{\text{in}}(Q) M_{\text{det}}(Q)$$ \hspace{1cm} (1)

The other function, denoted the absorbed-dose energy dependence, $f(Q)$, relates $D_{\text{det}}(Q)$ to the dose to another medium, $D_{\text{med}}(Q)$, in the absence of the detector, as a function of $Q$.

$$D_{\text{med}}(Q) = f(Q) D_{\text{det}}(Q)$$ \hspace{1cm} (2)
For a cavity (detector) that is large in comparison to range of electrons,

\[ f(Q) = \left( \frac{\mu_{en}(Q)}{\rho} \right)_{\text{wat}}^{\text{det}} \]  

(3)

where \( \left( \frac{\mu_{en}(Q)}{\rho} \right)_{\text{wat}}^{\text{det}} \) is ratio of mean mass-energy absorption coefficient of medium-to-detector at \( Q \).

The above equation is applicable when there is charged particle equilibrium and the energy fluence spectrum of photon is not perturbed by the detector. For more details on \( f(Q) \), see DeWerd et al. \(^{(28)}\)

In brachytherapy, quantity of interest is dose to water. The detectors are generally calibrated against a reference beam, which is usually \( ^{60}\text{Co} \). The relative absorbed-dose energy response correction \( R \) is the largest single source of Type B (systematic) uncertainty for TLD and other secondary dosimeters used in brachytherapy dosimetry. For a given detector material and a beam quality \( Q \), \( R \) is defined as:

\[ R = \frac{D_{\text{det}} / D_{\text{wat}}^{\text{wat}}}{D_{\text{det}} / D_{\text{wat}}^{\text{Co}}} \]  

(4)

where the numerator represents detector-to-water dose ratio at \( Q \) (\( ^{125}\text{I} \) or \( ^{169}\text{Yb} \)), and the denominator represents the same dose ratio at \( ^{60}\text{Co} \).

In the presence of charged particle equilibrium and when the detector material does not alter the photon energy fluence spectrum (see Eq. (3)), the above equation can be written as:

\[ R = \left[ \frac{1}{f(Q)} \right] / \left[ \frac{1}{f(^{60}\text{Co})} \right] = \frac{\left[ \mu_{en} / \rho \right]_{\text{det}} / \left[ \mu_{en} / \rho \right]_{\text{wat}}^{\text{wat}}}{\left[ \mu_{en} / \rho \right]_{\text{det}} / \left[ \mu_{en} / \rho \right]_{\text{wat}}^{\text{Co}}} \]  

(5)

Here, the numerator represents ratio of mean mass-energy absorption coefficient of detector-to-water at \( Q \), and the denominator represents the same ratio at \( ^{60}\text{Co} \).

Figure 1 presents the values of \( R \) for the investigated detector materials shown as a function of photon energy in the range 10 keV to 1.5 MeV. The values are based on the mass energy absorption coefficients data by Hubbell and Selzter. \(^{(31)}\)
C. Monte Carlo calculations

C.1 DOSRZnrc simulations of dose ratios for 60Co beam

Calculation of dose ratios at 60Co is important to derive $R$ (see denominator of Eq. (4)). Dose ratios in water phantom for the investigated detector materials for the 60Co beam, $\left[\frac{D_{\text{det}}}{D_{\text{wat}}}\right]_{60Co}$, are calculated using the DOSRZnrc user code of EGSnrc code system. Here, $D_{\text{det}}$ and $D_{\text{wat}}$ represent dose to detector and dose to water, respectively. In the Monte Carlo calculations, a parallel 60Co beam is incident on a 20 cm radius by 40 cm height cylindrical water phantom. The beam has a radius of 5.64 cm at the front face of the phantom (field size is 100 cm$^2$). A realistic 60Co spectrum from a telecobalt unit distributed along with the EGSnrc code system is used in the calculations. Cylindrical detector materials of 0.5 cm diameter and varying thicknesses are positioned at a depth 0.5 cm along the central axis of the water phantom. The thicknesses of the detector material are varied from 0.1 cm to 0.5 cm to study the influence of the thickness of the detector on $\left[\frac{D_{\text{det}}}{D_{\text{wat}}}\right]_{60Co}$.

C.2 FLURZnrc simulations of collision kerma and mean energies for 125I and 169Yb sources

For the calculation of dose ratio of detector-to-water for the 125I and 169Yb sources (numerator of Eq. (4)), we used the FLURZnrc user-code. In the calculations, the photon fluence spectrum is scored in 0.5 mm thick and 0.5 mm high cylindrical shells, along the transverse axis of the sources (distances, 0.5 cm–15 cm) in the 20 cm radius by 40 cm high cylindrical phantoms. The fluence spectrum is converted to collision kerma to water and collision kerma to detector materials by using the mass-energy absorption coefficients of water and detector materials. Using the values of collision kerma to water and collision kerma to detector materials, the denominator of Eq. (1) is obtained for the 125I and 169Yb sources. In the calculation of collision kerma to detector materials, no detector material is present. We have assumed that the presence of the detector materials does not affect the photon fluence spectrum. At 125I and 169Yb photon energies, charged particle equilibrium exists and the collision kerma may be approximated to absorbed dose.

Fig. 1. Values of the relative absorbed-dose energy response correction $R$ presented for different detector materials shown as a function of photon energy in the range 10 keV–1.5 MeV. The values are calculated using the mass-energy-absorption coefficients of the detector materials and water.
The fluence weighted mean energy \( \overline{E}_f \) and the detector-kerma weighted mean energy \( \overline{E}_{k,m} \) (suffix \( m \) represents detector material) are calculated as a function of distance from the source along the transverse axis using the following expressions:

\[
\overline{E}_f = \frac{\int E \Phi(E) dE}{\int \Phi(E) dE} \tag{6}
\]

\[
\overline{E}_{k,m} = \frac{\int E^2 \Phi(E) \left( \frac{\mu_{en}(E)}{\rho} \right)_m dE}{\int E \Phi(E) \left( \frac{\mu_{en}(E)}{\rho} \right)_m dE} \tag{7}
\]

where \( E \) is the kinetic energy of photon in keV, \( \Phi(E) \) is the differential photon fluence spectrum at \( E \) about \( dE \) and \( \left( \frac{\mu_{en}(E)}{\rho} \right)_m \) is the mass energy absorption coefficient of the detector material \( m \) at the photon energy \( E \). The values of \( \overline{E}_f \) and \( \overline{E}_{k,m} \) are calculated for the \( ^{125}\text{I} \) (selectSeed) and \( ^{169}\text{Yb} \) (model 4140) as well as for the bare sources.

C.3 Monte Carlo parameters and statistical uncertainties

The PEGS4 dataset needed for Monte Carlo calculations described above is based on XCOM\(^{(32)}\) compilations. We set \( A_E = 0.521 \text{ MeV} \) (kinetic energy of the electron is 0.01 MeV) and \( A_P = 0.001 \text{ MeV} \) while generating the PEGS4 dataset, where the parameters \( A_E \) and \( A_P \) are the low-energy thresholds for the production of knock-on electrons and secondary bremsstrahlung photons, respectively. All the calculations utilized the PRESTA-II step length and EXACT boundary crossing algorithms. In all calculations, electron range rejection technique is used to save computation time. We set \( E_{SAVE} = 2 \text{ MeV} \) for this purpose.

The photon transport cut off energy \( PCUT \) is chosen at 1 keV in all calculations. In DOSRZnrc calculations, we set \( A_E = ECUT = 0.521 \text{ MeV} \) (10 keV kinetic energy). In the FLURZnrc calculations, electrons are not transported by setting electron transport cutoff parameter \( ECUT = 2 \text{ MeV} \) (kinetic energy). Up to \( 10^9 \) photon histories are simulated. The 1\( \sigma \) statistical uncertainties on the calculated DOSRZnrc-based dose values are generally within 0.3%. The 1\( \sigma \) statistical uncertainties on the calculated FLURZnrc-based collision kerma values are usually 0.1% and never exceeded 0.2%. The statistical uncertainties on the calculated values of \( R \) are less than 0.6%. Throughout the text, the number shown in parentheses following a value represents the absolute uncertainty on the last digit of the value with a coverage factor \( k = 1 \).

III. RESULTS & DISCUSSION

A. Mean energies

An analysis of XCOM data shows that the interaction mechanisms at 27 keV photons in water are 46.4% photo electric absorption, 41% Compton scattering and 12.6% coherent scattering. At this energy, even after multiple Compton scattering in water, the energy of the scattered photons does not change significantly. Hence, the mean energies of the \( ^{125}\text{I} \) source do not change with the depth in water. For example, \( \overline{E}_f \) for the selectSeed \( ^{125}\text{I} \) source and the bare \( ^{125}\text{I} \) line source is about 28 keV in water, independent of distance.

For the \( ^{169}\text{Yb} \) source, mean energies decrease with distance in water. This is due to substantial degradation in the photon energy after scattering. Figure 2 presents the values of \( \overline{E}_f \) and \( \overline{E}_{k,m} \) for the \( ^{169}\text{Yb} \) (model 4140) source at various transverse axis distances in water. The
energy degradation is significant in PMMA and polystyrene phantoms when compared to water phantom because scattering is high in PMMA and polystyrene.

B. Dose ratios for $^{60}$Co beam

Table 2 presents the values of $D_{det} / D_{wat}$ for different detector materials at 0.5 cm depth in water phantom for various detector thicknesses. Also shown in this Table are the values of $\mu_{en} / \rho_{det} / (\mu_{en} / \rho_{wat})_{Co}$ calculated at 1.25 MeV, and $< Z / A >_{det} / < Z / A >_{wat}$, for comparison purposes. For a given detector material, the dose ratio is independent of detector thickness. It is interesting to see that the values of $\mu_{en} / \rho_{det} / (\mu_{en} / \rho_{wat})_{Co}$ agree well with the values of $D_{det} / D_{wat}$ (a maximum difference of 1.8% is observed for the air material). This suggests that at the $^{60}$Co energies, the investigated detectors (thickness from 0.1 cm to 0.5 cm) behave like a photon detector, as Compton scattering is the predominant interaction in all the detector materials. This implies that dose to detector is related to dose to water by the relation $D_{det} / D_{wat} = [\mu_{en} / \rho]_{det} / (\mu_{en} / \rho)_{wat}$, as the difference between the values of $[\mu_{en} / \rho]_{det} / (\mu_{en} / \rho)_{wat}$ is small (see Table 2), we have used $[\mu_{en} / \rho]_{det} / (\mu_{en} / \rho)_{wat}$ values for calculating $R$. 

![Fig. 2. Monte Carlo-calculated collision kerma weighted mean energies of different detector materials shown as a function of distance along transverse axis of the $^{169}$Yb source (model 4140) in water. Also shown is the fluence weighted mean energy in water for comparison purpose.](image_url)
Table 2. Monte Carlo-calculated ratio of dose to detector and dose to water for different detector materials for $^{60}$Co beam presented for different detector thickness. The number shown in parentheses following a value represents the absolute uncertainty on the last digit of the value with a coverage factor $k=1$. The radius of the detector is 5 mm. These detectors are at a depth of 0.5 cm in a 20 cm radius by 40 cm height unit density water phantom. The $^{60}$Co beam has a radius of 5.64 cm at the phantom surface. Also shown in this table are the values of ratio of mass-energy-absorption coefficients of detector to water calculated at the $^{60}$Co energy (1.25 MeV) and the values of ratio of $<Z/A>$ of detector to water.

| Detector Material | Thickness of Detector (mm) | $\frac{\langle \mu_m / \rho \rangle_{det}}{\langle \mu_m / \rho \rangle_{water}}$ | $<Z/A>_{det}$ | $<Z/A>_{water}$ |
|-------------------|---------------------------|---------------------------------|----------------|----------------|
| LiF               | 0.828(5)                  | 0.830(4)                        | 0.833(3)       | 0.828(2)       |
| Li$_2$B$_4$O$_7$   | 0.862(5)                  | 0.862(3)                        | 0.865(2)       | 0.873           |
| Diamond           | 0.879(5)                  | 0.886(2)                        | 0.886(3)       | 0.906           |
| Silicon           | 0.906(5)                  | 0.907(4)                        | 0.906(3)       | 0.894           |
| Al$_2$O$_3$       | 0.883(5)                  | 0.884(4)                        | 0.884(3)       | 0.873(2)       |
| Air               | 0.883(4)                  | 0.883(4)                        | 0.874(4)       | 0.885           |

C. Relative absorbed-dose energy response correction

C.1 $^{125}$I source

For a given detector, $R$ is independent of distance for $^{125}$I source. Value of $R$ at a given distance for a given material is insensitive to source model and the phantom material. Table 3 presents the values of $R$ calculated for selectSeed $^{125}$I source in water and bare line $^{125}$I source in water, polystyrene and PMMA phantoms for silicon. The results suggest that $R$ is insensitive to source model, distance from source and phantom materials. The distance-independent values of $R$ obtained for $^{125}$I source (selectSeed or bare $^{125}$I source) in water, polystyrene and PMMA phantoms are 1.41, 0.92, 3.97, 0.47, 8.32 and 1.10, respectively, for LiF, Li$_2$B$_4$O$_7$, Al$_2$O$_3$, diamond, Si and air. This suggests that the investigated detectors are good for relative dose measurements in water for $^{125}$I sources.

Table 3. Monte Carlo-calculated values of relative absorbed-dose energy response correction $R$ for silicon detector material for selectSeed $^{125}$I source in water phantom and for bare $^{125}$I line source in water, polystyrene and PMMA phantoms.

| Distance (along transverse axis of the source) (cm) | SelectSeed $^{125}$I | Bare $^{125}$I Line Source |
|---------------------------------------------------|----------------------|----------------------------|
|                                                   | Water                | Water | Polystyrene | PMMA |
| 0.5                                               | 8.318                | 8.324 | 8.324       | 8.324 |
| 1.0                                               | 8.318                | 8.324 | 8.324       | 8.324 |
| 2.0                                               | 8.318                | 8.323 | 8.323       | 8.323 |
| 3.0                                               | 8.318                | 8.322 | 8.322       | 8.322 |
| 4.0                                               | 8.318                | 8.321 | 8.321       | 8.322 |
| 5.0                                               | 8.317                | 8.319 | 8.320       | 8.321 |
| 6.0                                               | 8.316                | 8.319 | 8.319       | 8.320 |
| 7.0                                               | 8.314                | 8.318 | 8.319       | 8.320 |
| 8.0                                               | 8.314                | 8.318 | 8.318       | 8.319 |
| 9.0                                               | 8.313                | 8.318 | 8.317       | 8.318 |
| 10.0                                              | 8.312                | 8.313 | 8.317       | 8.318 |
| 11.0                                              | 8.310                | 8.312 | 8.316       | 8.317 |
| 12.0                                              | 8.309                | 8.310 | 8.316       | 8.317 |
| 13.0                                              | 8.307                | 8.308 | 8.316       | 8.316 |
| 14.0                                              | 8.306                | 8.306 | 8.315       | 8.315 |
| 15.0                                              | 8.303                | 8.304 | 8.315       | 8.315 |
The value of \( R = 1.41 \) for LiF-TLD calculated in the present work is consistent with the published value of 1.41 by Meigooni et al.\(^8\) Mobit and Badragan\(^{22}\) have also studied the energy response correction for LiF-TLD micro rods of different diameters. Their study showed that \( R \) is sensitive to diameter of LiF rod (i.e. the values of \( R \) are 1.406 ± 0.2% and 1.323 ± 0.2% for rods of 0.1 cm and 0.5 cm diameter (0.6 cm length), respectively.)\(^{22}\) The authors also studied the angular and radial distance dependence of \( R \). For a LiF-TLD of diameter 1 mm calibrated at 1 cm on the transverse axis of the \(^{125}\)I source in water, \( R \) decreases by a maximum of 3.5% within the \( 6 \text{ cm} \times 6 \text{ cm} \times 6 \text{ cm} \) calculation region. For the 5 mm diameter LiF-TLD, \( R \) decreases by a maximum of 5% in the same region. Note that a 5% uncertainty is assigned to \( R \) for the LiF-TLD-100 based dosimetry.\(^{33}\)

### C.2 \(^{169}\)Yb source

Figure 3 presents the Monte Carlo-calculated values of \( R \) for the \(^{169}\)Yb (model 4140) source as a function of distance along the transverse axis of the source for different detector materials. For a given detector material, the \(^{169}\)Yb source (model 4140) shows \( R \) is distance-dependent (change in \( R \) is not significant for distances 5 cm and above). For example, \( R \) increases from 1.11 to 1.18, 1.80 to 2.28, 2.97 to 4.17 and 1.036 to 1.052, respectively for the LiF, \( \text{Al}_2\text{O}_3 \), silicon and air detector materials when the distance is varied from 0.5 cm to 15 cm. This is because mean energy decreases with depth in water (see Fig. 2), which results in increase in \( R \). Increase in \( R \) is substantial for the Si diode detector (up to 40%) as the atomic number is high (see Table I). For the air material, increase in \( R \) is within 6% as its \( Z_{\text{eff}} \) is comparable to that of water. The detector materials, \( \text{Li}_2\text{B}_4\text{O}_7 \) and diamond have \( Z_{\text{eff}} \) values smaller than that of water. Hence, the values of \( R \) decrease from 0.972 to 0.954 and 0.832 to 0.781, respectively, for the \( \text{Li}_2\text{B}_4\text{O}_7 \) and diamond detectors, for the above-mentioned distance range.

MacPherson and Battista\(^{15}\) studied the response for LiF for \(^{169}\)Yb point source as a function of distance in water using the Monte Carlo-based MCNP code. The values of \( R \) reported are 1.16 and 1.2 at 1 cm and 4 cm, respectively. In our study, we obtain the values of 1.12 and 1.15 at 1 cm and 4 cm for the bare \(^{169}\)Yb line source, respectively.

The Monte Carlo-calculated values of \( R \) for the investigated detectors for the bare \(^{169}\)Yb source are different from the \(^{169}\)Yb 4140 source model. This is shown in Fig. 4 for the LiF detector. The calculations with the bare \(^{169}\)Yb line source have resulted in overestimation.
of $R$ when compared to the encapsulated $^{169}$Yb source (model 4140) for LiF, $\text{Al}_2\text{O}_3$, and Si detectors. For example, depending upon the distance, the overestimation is about 3% for LiF (independent of distance), 15% to 22% for silicon, and 11% to 15% for $\text{Al}_2\text{O}_3$. For Li$_2$B$_4$O$_7$ and air, the variation in the values of $R$ between the bare $^{169}$Yb source and the encapsulated $^{169}$Yb source (model 4140) is only 1%. This difference is comparable to the statistical uncertainty of about 1%. For diamond, the bare $^{169}$Yb source has resulted in underestimation of $R$ by 5%, independent of distance.

C.3 Effect of detector thickness
In general, $R$ includes corrections for volume averaging (influence of dose gradients in the detector volume) and self-absorption by the detector. The energy response of a Scanditronix (IBA Dosimetry GmbH, Schuirenbruck, Germany) p-type diode detector (active volume is 2.5 mm diameter and 60 $\mu$m thick)$^{12,14,17}$ for $^{125}$I source has been established to be within $\pm$ 3.5% for diode-to-source distance range of 0.5 cm to 10 cm.$^{12}$ The published value of average absolute response with respect to dose in water for $^{125}$I source is 6.75.$^{12}$ This value includes self-attenuation of the diode, which is 0.911.$^{12}$ The present study gives absolute response value of 7.44 (independent of distance), which does not include self-attenuation of the diode, because we have not modeled the full diode. When including the published self-attenuation of 0.911, we obtain average absolute response value of 6.78, which agrees with (within 0.4%) the above-mentioned published value of 6.75.

In the calculations, we have not modeled the detectors due to limitations associated with the DOSRZnrc and FLURZnrc codes.$^{124}$ For example, simulation of a cylindrical detector whose axis is parallel to the source axis (cylindrical source) cannot be modeled using the above user-codes. To quantify approximately the influence of finite detector dimensions on the calculated dose values of $^{125}$I and $^{169}$Yb sources, we modeled the detectors as cylindrical shells using the DOSRZnrc user-code. In this study, the Scanditronix p-type diode detector (60 mm thick active volume of the diode embedded in a circular disk of silicon substrate of diameter 3.5 mm and thickness 0.45 mm)$^{12,14,17}$ is modeled as cylindrical shells of height 1 mm. The 60 $\mu$m thick sensitive silicon diode (cylindrical shell) material is embedded between 0.225 cm thick silicon substrate shells. Both kerma and absorbed dose are scored in the 60 $\mu$m thick sensitive diode region. The values of kerma to Si diode and absorbed dose to Si diode obtained from the
DOSRZnrc simulation are statistically indistinguishable. A comparison of dose results obtained from the DOSRZnrc simulations with the FLURZnrc simulation (detector is not modeled in collision kerma calculations) gives self-attenuation by the diode detector. The value of self-attenuation by the diode detector obtained for the $^{125}$I source is 0.889(1) (independent of distance), which compares reasonably well with the published value of 0.911.(12) For the $^{169}$Yb source, the values of self-attenuation by the diode detector obtained at depths of 1 cm, 5 cm, 10 cm and 15 cm in water are 0.992(5), 0.983(5), 0.973(7), and 0.970(7), respectively. A similar study using the 1 mm thick and 1 mm height LiF and $\text{Al}_2\text{O}_3$ detectors in water gives negligible self-attenuation for the $^{169}$Yb source at all distances. However, for $^{125}$I source, self-attenuation by LiF is 0.975(5) and by $\text{Al}_2\text{O}_3$ is as large as 0.850(1), independent of distance.

C.4 Influence of phantom materials on energy response

The relative absorbed-dose energy response corrections obtained in the solid phantom materials PMMA, polystyrene and water are designated as $R_{\text{PMMA}}$, $R_{\text{Poly}}$ and $R_{\text{Water}}$ respectively. The ratios $R_{\text{PMMA}}/R_{\text{Water}}$ and $R_{\text{Poly}}/R_{\text{Water}}$ would demonstrate the influence of solid phantoms on the energy response of the detectors compared to the water phantom. Except for the air and $\text{Li}_2\text{B}_4\text{O}_7$ detector materials, the FLURZnrc-based collision kerma ratios (numerator of Eq. (1)) obtained in the solid phantom materials are different from that in water phantom. Figures 5 and 6 present the values of $R_{\text{PMMA}}/R_{\text{Water}}$ and $R_{\text{Poly}}/R_{\text{Water}}$ for the $^{169}$Yb source (model 4140). These figures demonstrate that both PMMA and polystyrene materials produce similar energy response corrections for air and $\text{Li}_2\text{B}_4\text{O}_7$ detector materials, at all distances. Whereas, for the rest of the detector materials, the values of $R_{\text{PMMA}}/R_{\text{Water}}$ and $R_{\text{Poly}}/R_{\text{Water}}$ deviate from unity (larger than unity implies over-response and smaller than unity implies under-response) as the distance increases. For example, for the Si detector material, the values of $R_{\text{PMMA}}/R_{\text{Water}}$ and $R_{\text{Poly}}/R_{\text{Water}}$ are 1.18 and 1.30, respectively, at 15 cm depth. For the diamond detector, the values of $R_{\text{PMMA}}/R_{\text{Water}}$ and $R_{\text{Poly}}/R_{\text{Water}}$ are 0.93 and 0.88, respectively, at 15 cm depth.

![Figure 5](https://example.com/fig5.png)

**Fig. 5.** Ratio of the relative absorbed-dose energy response correction in PMMA phantom to water phantom $R_{\text{PMMA}}/R_{\text{Water}}$ presented for different detector materials as a function of distance along the transverse axis of the $^{169}$Yb source (model 4140).
IV. CONCLUSIONS

The Monte Carlo-based relative absorbed-dose energy response corrections as a function of depth in water, PMMA and polystyrene phantoms for detector materials such as LiF, Li$_2$B$_4$O$_7$, A1$_2$O$_3$, diamond, silicon and air are calculated for the $^{125}$I and $^{169}$Yb brachytherapy sources. For the $^{125}$I source, the relative absorbed-dose energy response correction for a given detector is independent of distance in the phantom materials, suggesting that all the detector materials are good for relative dose measurements. For the $^{169}$Yb source, the correction is distance-dependent. As opposed to $^{125}$I source, detailed modeling of actual $^{169}$Yb is important, as the bare line source-based response is significantly different from the encapsulated $^{169}$Yb source (model 4140). The relative absorbed-dose energy response of a given detector for the $^{125}$I source is insensitive to the phantoms investigated. Whereas, for the $^{169}$Yb source, PMMA and polystyrene phantoms demonstrate over-response for LiF, A1$_2$O$_3$ and Si diode and under-response for diamond when compared to that in water medium. The over- or under-response is significant at large distances. The corrections for the detector materials, air and Li$_2$B$_4$O$_7$ are almost identical in all the phantoms.

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REFERENCES

1. Rivard MJ, Coursey BM, DeWerd LA, et al. Update of AAPM Task Group No. 43 Report: A revised AAPM protocol for brachytherapy dose calculations. Med Phys. 2004;31(3):633–74.
2. Meisberger LL, Keller RJ, Shalek RJ. The effective attenuation in water of the gamma rays of gold 198, iridium 192, cesium 137, radium 226 and cobalt 60. Radiology. 1968;90(5):953–57.

3. Venselaar JLM, Van der Giesen PH, Dries WJF. Measurement and calculation of the dose at large distances from brachytherapy sources: Cs-137, Ir-192, and Co-60. Med Phys. 1996;23(4):537–43.

4. Mishra V, Waterman FM, Suntharalingam N. Anisotropy of an 192Ir high dose rate source measured with a miniature ionization chamber. Med Phys. 1997;24(5):751–55.

5. Baltas D, Kramer R, Löffler E. Measurements of the anisotropy of the new Ir-192 source for the microSelectron-HDR. In: International Brachytherapy: Programme and Abstracts. 7th International Brachytherapy Working Conference. Baltimore, MD, The Netherlands: Nucletron International B.V.; 1992. p. 290–306.

6. Weaver KA. Response of LiF powder to 125I photons. Med Phys. 1984;11(6):850–54.

7. Ling CC, Schell MC, Yorke ED, Palos BB, Kubiatowicz DO. Two-dimensional dose distribution of 125I seeds. Med Phys. 1985;12(5):652–55.

8. Meigooni AS, Milia JA, Nath RA. Comparison of the solid phantoms with water for dosimetry of 125I brachytherapy sources. Med Phys. 1988;15(5):695–701.

9. Meigooni AS, Milia JA, Nath RA. Influence of the variation of energy spectra with depth in the dosimetry of 192Ir using LiF TLD. Phys Med Biol. 1988;33(10):1159–70.

10. Chiu-Tsao S, Anderson LL, O’Brien K, Sanna R. Dose rate determination for 125I seeds. Med Phys. 1990;17(5):815–25.

11. Venselaar JLM, Van der Giesen PH, Dries WJF. Measurement and calculation of the dose at large distances from brachytherapy sources: Cs-137, Ir-192, and Co-60. Med Phys. 1996;23(4):537–43.

12. Williamson JF, Perera H, Li Z, Lutz WR. Comparison of calculated and measured heterogeneity correction factors for 125I, 137Cs and 192Ir brachytherapy sources near localized heterogeneities. Med Phys. 1993;20(1):209–22.

13. Fontenla DP, Ahmad M, Chiu-Tsao S, Anderson LL. Diode dosimetry of 103Pd model 200 seed in water phantom. Med Phys. 1994;21(6):817–20.

14. Perera H, Williamson JF, Li Z, Mishra V, Meigooni AS. Dosimetric characteristics, air-kerma strength calibration and verification of Monte Carlo simulation for a new Ytterbium-169 brachytherapy source. Int J Radiat Oncol Biol Phys. 1994;28(4):953–70.

15. MacPherson MS, Battista JJ. Dose distributions and dose rate constants for new ytterbium-169 brachytherapy seeds. Med Phys. 1995;22(1):89–96.

16. Valicenti RK, Kirov AS, Meigooni AS, Mishra V, Das RK, Williamson JF. Experimental validation of Monte Carlo dose calculations about a high-intensity Ir192 source for pulsed dose-rate brachytherapy. Med Phys. 1995;22(6):821–29.

17. Das RK, Li Z, Perera H, Williamson JF. Accuracy of Monte Carlo photon transport simulation in characterizing brachytherapy dosimeter energy-response artefacts. Phys Med Biol. 1996;41(6):995–1006.

18. Rustgi SN. Application of a diamond detector to brachytherapy dosimetry. Phys Med Biol. 1998;43(8):2085–94.

19. Li Z, Palta JR, Fan JJ. Monte Carlo calculations and experimental measurements of dose distribution for new 103Pd sources. Med Phys. 2000;27(5):1108–12.

20. Solberg TD, DMarco JJ, Hugo G, Wallace RE. Dosimetric parameters of three new solid core I-125 brachytherapy sources. J Appl Clin Med Phys. 2002;3(2):119–34.

21. Nakano T, SUCHOVERSNA N, BILEK MM, MCKENZIE DR, NG N, KRON T. High dose-rate brachytherapy localization: positional resolution using a diamond detector. Phys Med Biol. 2003;48(14):2133–46.

22. Mobit P, Badragan I. Response of LiF-TLD micro-rods around 125I radioactive seed. Phys Med Biol. 2003;48(19):3129–42.

23. Andersen CE, Nielsen SK, Greilich S, Helt-Hansen J, Lindegaard JC, Tanderup K. Characterization of a fiber-coupled Al2O3:C luminescence dosimetry system for online in vivo dose verification during 192Ir brachytherapy. Med Phys. 2009;36(3):708–18.

24. Rogers DWO, Kawrakow I, Seuntjens JP, Walters BRB, Mainegra-Hing E. NRC User Codes for EGSnrc. NRCC Report PIRS-701. Ottawa, ON: National Research Council of Canada; 2010. Available from: http://www.irs.inms.nrc.ca/EGSnrc/pirs701.pdf

25. Kawrakow I, Mainegra-Hing E, Rogers DWO, Tessler F, Walters BRB. The EGSnrc Code System: Monte Carlo simulation of electron and photon transport. NRCC Report PIRS-702 (revB). Ottawa, ON: National Research Council of Canada; 2010. Available from: http://www.irs.inms.nrc.ca/EGSnrc/pirs702.pdf

26. Solberg TD, DMARCO JJ, Hugo G, Wallace RE. Dosimetric parameters of three new solid core I-125 brachytherapy sources. J Appl Clin Med Phys. 2002;3(2):119–34.

27. Meigooni AS, Milia JA, Nath RA. Influence of the variation of energy spectra with depth in the dosimetry of 192Ir using LiF TLD. Phys Med Biol. 1988;33(10):1159–70.

28. Mobit P, Badragan I. Response of LiF-TLD micro-rods around 125I radioactive seed. Phys Med Biol. 2003;48(19):3129–42.

29. Andersen CE, Nielsen SK, Greilich S, Helt-Hansen J, Lindegaard JC, Tanderup K. Characterization of a fiber-coupled Al2O3:C luminescence dosimetry system for online in vivo dose verification during 192Ir brachytherapy. Med Phys. 2009;36(3):708–18.

30. Rogers DWO, Kawrakow I, Seuntjens JP, Walters BRB, Mainegra-Hing E. NRC User Codes for EGSnrc. NRCC Report PIRS-701. Ottawa, ON: National Research Council of Canada; 2010. Available from: http://www.irs.inms.nrc.ca/EGSnrc/pirs701.pdf

31. Karaiskos P, Papagiannis P, Sakelliou L, Anagnostopoulos G, Baltas D. Monte Carlo dosimetry of the selectSeed 125I interstitial brachytherapy seed. Med Phys. 2001;28(8):1753–60.

32. Medich DC, Tries MA, Munro JJ. Monte Carlo characterization of an ytterbium-169 high dose rate brachytherapy source with analysis of statistical uncertainty. Med Phys. 2006;33(1):163–72.

33. DeWerd LA, Bartol LJ, Davis SD. Thermoluminescence dosimetry. In: Clinical Dosimetry Measurements in Radiotherapy. Madison WI: Medical Physics Publishing; 2009.

34. Sutherland JCH, Rogers DWO. Monte Carlo calculated absorbed-dose energy dependence of EBT and EBT2 film. Med Phys. 2010;37(3):1110–16.

35. Hubbell JH, Selzter SM. Tables of x-ray mass attenuation coefficients and mass energy absorption coefficients. Gaithersburg, MD: National Institute of Standards and Technology; 1995. Available from: http://physics.nist.gov/physRefData/XrayMassCoef.
32. Berger MJ, Hubbell JH. XCOM, Photon cross sections on a personal computer. Report No. NBSIR87–3597. Gaithersburg, MD: NIST; 1987. Available from: http://physics.nist.gov/PhysRefData/Xcom/html/xcom1.html
33. Meigooni AS, Gearheart DM, Sowards K. Experimental determination of dosimetric characteristics of Best double-walled I-125 brachytherapy source [Data sheet]. Springfield, VA: Best Industries Inc.; 2000.