Neutron spectra and dose equivalents calculated in tissue for high-energy radiation therapy

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Neutrons are by-products of high-energy radiation therapy and a source of dose to normal tissues. Thus, the presence of neutrons increases a patient's risk of radiation-induced secondary cancer. Although neutrons have been thoroughly studied in air, little research has been focused on neutrons at depths in the patient where radiosensitive structures may exist, resulting in wide variations in neutron dose equivalents between studies. In this study, we characterized properties of neutrons produced during high-energy radiation therapy as a function of their depth in tissue and for different field sizes and different source-to-surface distances (SSD). We used a previously developed Monte Carlo model of an accelerator operated at 18 MV to calculate the neutron fluences, energy spectra, quality factors, and dose equivalents in air and in tissue at depths ranging from 0.1 to 25 cm. In conjunction with the sharply decreasing dose equivalent with increased depth in tissue, the authors found that the neutron energy spectrum changed drastically as a function of depth in tissue. The neutron fluence decreased gradually as the depth increased, while the average neutron energy decreased sharply with increasing depth until a depth of approximately 7.5 cm in tissue, after which it remained nearly constant. There was minimal variation in the quality factor as a function of depth. At a given depth in tissue, the neutron dose equivalent increased slightly with increasing field size and decreasing SSD; however, the percentage depth-dose equivalent curve remained constant outside the primary photon field. Because the neutron dose equivalent, fluence, and energy spectrum changed substantially with depth in tissue, we concluded that when the neutron dose equivalent is being determined at a depth within a patient, the spectrum and quality factor used should be appropriate for depth rather than for in-air conditions. Alternately, an appropriate percent depth-dose equivalent curve should be used to correct the dose equivalent at the patient surface. © 2009 American Association of Physicists in Medicine. [DOI: [10.1118/1.3089810](https://doi.org/10.1118/1.3089810)]

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I. INTRODUCTION

High-energy radiation therapy administered using medical accelerators operated at energies >8 MV results in the production of neutrons,^{1,2} mainly in the primary and secondary collimators, the jaws, and the x-ray target.^{3,4} These neutrons scatter throughout the treatment vault, transferring energy primarily through elastic scatter with low- Z materials until eventually the neutrons are thermalized ($E=0.024$ eV) and finally captured by a receptive nucleus. These neutrons pose a health risk for any patient undergoing high-energy radiation therapy because they increase the out-of-field radiation dose and, thus, the corresponding risk of secondary malignancies.⁵⁻⁹ Studies have measured or used Monte Carlo to calculate neutron fluence and energy spectrum, but the majority of these studies have evaluated such neutron properties in air or on the patient's surface rather than within a phantom or patient where organs at risk are actually found.^{5-7,10-13} This observation is not surprising because measurements within a patient or even a phantom are very difficult due to substantial challenges in neutron detection. In some instances, the neutron dose equivalent has been determined within a phantom through measurements with bubble detectors¹⁴ and using Monte Carlo.^{15,16} In both approaches, a

pronounced decrease in neutron dose equivalent with increasing depth in tissue was observed. Although depth in tissue has been found to be one of the factors that most substantially affects the neutron dose equivalent, there has been a paucity of studies on the actual neutron dose or other properties as a function of depth in tissue. Zanini *et al.*¹⁷ investigated the neutron spectrum at specific organ locations in a phantom from a specific irradiation field using bubble detectors. Such measurements offer valuable information. However, additional and more general information about the neutron spectrum and related properties at depths in tissue would provide more generalized results and allow more general patient-specific scenarios to be evaluated.

In the current study, we examined the neutron field produced by a medical accelerator operated at 18 MV as it exists within a tissue-equivalent phantom. In particular, we evaluated fluence, energy spectra, and quality factors as a function of depth and examined the importance of field size and source-to-surface distance (SSD) on the absolute neutron dose equivalent and the percentage depth-dose equivalent (PDDE) in tissue. This information will be useful for evaluating the neutron dose equivalent at locations within a patient.

FIG. 1. Monte Carlo model of accelerator, couch, vault, and tissue-equivalent phantom for scoring neutrons. Tally locations through the phantom are illustrated with arrows.

II. METHODS AND MATERIALS

We used a Monte Carlo model of a Varian 2100 Clinac (Varian Medical Systems, Palo Alto, CA) operated at 18 MV that was previously developed¹⁸ in MCNPX (version 2.5, Los Alamos National Laboratory, Los Alamos, NM).¹⁹ This model includes the entire accelerator head, including a 120-leaf millennium multileaf collimator (MLC), gantry, treatment couch, and treatment vault. An 18 MV photon beam was simulated with 18.0 MeV electrons incident on the bremsstrahlung target. Neutron production and scattering throughout the treatment vault were modeled with the ENDF/IV library, and neutrons were tracked at all energies until captured. Neutron fluence and energy spectra in air calculated by this model were previously benchmarked and found to agree well with measurements.¹⁸

The Monte Carlo model was used in the present study to evaluate neutron characteristics (such as dose equivalent) in a rectangular phantom of ICRU tissue.²⁰ The phantom measured 100 cm in length and width and 30 cm in thickness (in the beam-line direction) and was 11.1% carbon, 76.2% oxygen, 10.1% hydrogen, and 2.6% nitrogen by weight. Neutron fluence was tallied in the phantom as a function of energy (five bins per decade) using a track-length estimate of the cell flux (F4 tally¹⁹). The tally voxels were 2×2 cm² in the lateral dimensions and between 0.1 and 2 cm in the beam-line direction, with thinner voxels used near the surface.

Neutron characteristics were first studied as a function of depth in the phantom from a 10×10 cm² field (MLC: 10×10 , jaws: 10.2×10.2) with the phantom at 100 cm SSD. The neutron energy spectra were calculated in the phantom on the central axis and at 25 cm from the central axis (toward the gantry) throughout the depth of the phantom (Fig. 1). The spectra were used to calculate dose equivalent, average energy, and average quality factor as detailed below. To contrast neutron properties in tissue with those in air, the total

fluence was also calculated along the central axis with the phantom absent, beginning 70 cm from the bremsstrahlung target (30 cm above the phantom surface) and extending to 130 cm from the target (through to the bottom of the phantom).

In addition to tallies through the depth of the phantom, neutron fluence was also calculated at the surface of the tissue phantom (0.1 cm depth) as a function of distance from the central axis for different field sizes and SSDs. Specifically, a closed field (MLC and jaws closed), a 10×10 cm² field (MLC: 10×10 , jaws: 10.2×10.2), and a 15×15 cm² field (MLC: 15×15 , jaws: 15.2×15.2) were simulated at 100 cm SSD. The 10×10 cm² field was also simulated at 90 and 110 cm SSDs. For these field size and SSD configurations, PDDE curves were also calculated on and off the central axis (at 0, 25, and 45 cm toward the gantry), normalized to the dose equivalent at 0.1 cm depth at each distance.

Specific neutron properties were calculated in the following manner based on the neutron fluence. The neutron dose equivalent (H) was determined using the following equation:

$$H = \sum_E \phi(E) \cdot K(E) \cdot Q_n(E), \quad (1)$$

where $K(E)$ represents the neutron kerma factors²¹ and $Q_n(E)$ represents the neutron quality factors.²² The average neutron quality factor was also calculated according to the following equation:

$$\bar{Q}_n = \frac{\sum_E Q_n(E) \phi(E)}{\sum_E \phi(E)}. \quad (2)$$

The energy dependent neutron quality factors, $Q_n(E)$, were calculated in tissue for each neutron energy by Siebert and Schuhmacher²² based on the neutrons' linear energy transfer (LET) in tissue and the $Q(\text{LET})$ versus LET relationship defined by the ICRP.²³ The values of $Q_n(E)$ include only the LET of charged secondary particles generated by neutron interactions and do not include the biological weighting of emitted photons from neutron capture events. In the current study we used $Q_n(E)$, which excludes capture gammas, for two reasons. First, this approach of excluding the capture gammas is consistent with epidemiologic studies, such as atomic bomb survivor studies, which evaluate neutron dose solely based on charged secondary particles.²³ Second, out-of-field dose studies around a high-energy linear accelerator will almost invariably include separate and additional photon measurements to account for head leakage photons and photons scattered from the primary photon beam. Any such photon measurements will also measure capture gammas.

Because of their variety and complexity, it is worthwhile to distinguish the biological weighting factors used in this study from other weighting factors that are reported in the literature. For example, although not considered in this study, the biological weighting of capture gammas can be included in the quality factor, in which case the dose and dose equivalent are related by Q_{eff} . Q_n and Q_{eff} are conceptually related as follows:

FIG. 2. Neutron fluence in arbitrary units (A.U.) as a function of depth in tissue (d) with the phantom present or absent. Phantom surface was located at $d=0$ and negative depths refer to the fluence in air above the phantom.

$$H_{n+n_\gamma} = Q_{\text{eff}}(D_n + D_{n_\gamma}) = Q_n \cdot D_n + D_{n_\gamma}, \quad (3)$$

where H_{n+n_γ} is the dose equivalent from neutrons and capture gammas, D_n is the dose from neutrons and D_{n_γ} is the dose from capture gammas (which have a defined quality factor of 1).²³ Use of Q_{eff} is confounded by its dependence on the geometry of the patient and radiation field, as these parameters affect the relative importance of photons and neutrons. That is, while Q_n depends only on the neutron energy spectrum and the material composition of the medium in which those neutrons interact, Q_{eff} depends additionally on the size and shape of the phantom or patient and the directionality of the radiation field and is therefore difficult to define uniquely. Q_{eff} is most commonly considered in the context of the effective dose equivalent for a person, H_E , which, when divided by the organ-weighted absorbed physical dose D' , yields the specially defined Q_{eff} of q_E :

$$\frac{H_E}{D'} = q_E. \quad (4)$$

The value of q_E is of general interest because it is after this value that the most recent radiation weighting factors w_R have been patterned (although they are different by a scaling factor).²³ Numerically, the values of Q_n and q_E or w_R are similar but are most distinct for thermal neutrons, where Q_n is 16.2 (in tissue)²² while q_E and w_R are around 2.²³

III. RESULTS

Figure 2 shows the neutron fluence with the phantom present and absent. With the phantom absent (“air”), the total neutron fluence decreased with distance in an approximately inverse square manner. The presence of the phantom drastically affected the neutron fluence as neutrons are readily scattered. The presence of the full-sized phantom therefore produces a full-scatter scenario, analogous to full phantom scatter in photon beams. There was a marked increase in the neutron fluence near the surface of the phantom and in the air up to 20 cm above the phantom. The neutron fluence reached

FIG. 3. Neutron energy spectra in A.U. from a Varian 2100 accelerator calculated at the same location relative to the treatment head in air and at 0.1 cm depth in tissue.

a maximum at approximately 2 cm depth in the phantom, beyond which it decreased sharply with depth.

Figure 3 compares neutron spectra in air and at a 0.1 cm depth in tissue, calculated at the same location: 25 cm away from the central axis. Error bars based on the statistical uncertainty are included for both spectra and are relatively small. Both spectra showed a predominance of neutrons around 0.4 MeV that were primarily direct neutrons, that is, neutrons that traveled directly from the accelerator head to the patient. At energies below this peak, the spectra also showed the presence of epithermal neutrons, which were primarily scattered neutrons, that is, neutrons that scattered throughout the treatment vault. Finally, at energies of 10^{-7} – 10^{-8} MeV, both spectra showed a peak of thermal neutrons that originated largely from neutron interactions with hydrogen (either in tissue or in the concrete of the treatment vault). Although both spectra were calculated at the same position relative to the isocenter, the increased scatter from the phantom resulted in an increase in total fluence from 116×10^3 to 192×10^3 n/cm²/MU with the presence of the phantom, with a small statistical uncertainty of less than 1% in each value. There was also a decrease in the average neutron energy from 0.24 MeV in air to 0.14 MeV in tissue, with statistical uncertainties of 1.6% and 1.3%, respectively, in these values.

Figure 4 shows the neutron spectra in tissue at depths from 0.1 to 19.5 cm. At shallow depths (0.1–4.5 cm), there was a substantial decrease in the direct neutron peak at 0.4 MeV and an increase in the thermal neutron peak at 10^{-7} – 10^{-8} MeV with increasing depth. These changes occurred as fast neutrons lose energy and rapidly become thermalized, primarily because they have a relatively large cross section for elastic scatter with hydrogen.¹ The buildup of thermal neutrons occurred because the neutron cross sections are different for elastic scatter and neutron capture. Neutron absorption occurs only with thermal neutrons (primarily through absorption by nitrogen); however, the cross section for absorption is smaller than that for elastic scattering. At depths deeper than 4.5 cm, an increase in depth in tissue

FIG. 4. Neutron energy spectra in A.U. at depths of 0.1, 1.5, 4.5, 9.5, and 19.5 cm in tissue.

corresponded to a decrease in both the numbers of direct neutrons (0.4 MeV) and thermal neutrons (10^{-7} – 10^{-8} MeV). The remaining fast neutrons continued to be thermalized rapidly, further decreasing the direct neutron peak. However, because there were fewer fast neutrons remaining, only a few thermal neutrons were produced. Meanwhile, the existing thermal neutrons continued to be absorbed, decreasing the thermal neutron peak.

Figure 5 shows the average neutron energy as a function of depth in tissue at 100 cm SSD from a 10×10 cm² field at 25 cm from the central axis. Not surprisingly, considering Fig. 4, the average neutron energy decreased rapidly over the first 5 cm of tissue depth as the energy of fast neutrons was quickly degraded and thermal neutrons accumulated. However, at depths greater than 7–8 cm, the neutron energy became nearly constant because equilibrium was achieved between the absorption of thermal neutrons and the thermalization of the remaining fast neutrons. That is, the shape of the energy spectrum remained nearly constant at this and greater depths and only reduced in magnitude with increasing depth.

Figure 6 shows the average quality factor for neutrons as a function of depth in tissue. The quality factor in tissue is

FIG. 5. Average neutron energy as a function of depth in tissue.

FIG. 6. Neutron quality factor (Q_n) as a function of depth in tissue.

greatest for fast neutrons (where Q_n is between 16 and 20 over the energy range of 0.1–1 MeV) and for thermal neutrons (where $Q_n = 16.2$ due to proton recoil following capture by nitrogen).²² At all depths, the vast majority of neutrons were either fast or thermal, so the quality factor did not change as drastically as neutron fluence or average neutron energy as a function of depth in tissue. However, there was a slight decrease in the average quality factor with increasing depth due primarily to the decreased number of fast neutrons (for which the quality factor is the highest) relative to epithermal and thermal neutrons. Overall, the quality factor for neutrons decreased from 17.1 at a depth of 0.1 cm to 13.8 at a depth of 19.5 cm.

Figure 7 shows the neutron dose equivalent at the surface of the tissue phantom as a function of distance from the central axis for three field sizes. Statistical uncertainties in each calculated dose equivalent as estimated by the Monte Carlo code were small, averaging less than 1%. To avoid overcrowding the image, error bars, which were similar for each field size, are included only on the 10×10 cm² field to illustrate their size. Within the boundaries of the primary

FIG. 7. Dose equivalent (H) as a function of distance from the central axis calculated at the phantom surface for field sizes of 0×0 , 10×10 , and 15×15 cm².

FIG. 8. Dose equivalent (H) as a function of distance from the central axis calculated at the surface for SSDs of 90, 100, and 110 cm.

photon field, the neutron dose equivalent is elevated because there is no head shielding to degrade the neutron energy through scatter. This elevated dose equivalent decreases beyond the edge of the primary photon field with a penumbra width of approximately 3 cm. Beyond this penumbra, the neutron dose equivalent decreases monotonically but only slightly with increasing distance from the central axis. The dose equivalent also changed only slightly with field size. Compared to the closed field, the dose equivalents outside the primary photon field were approximately 14% and 18% larger for the 10×10 cm² field and the 15×15 cm² field, respectively. This difference was most pronounced close to the treatment field and was almost negligible by 50 cm from the central axis. Previous work by Mao *et al.*³ also demonstrated a small effect from field size (less than 10% variation, although decreasing with field size); however, this was evaluated based on total neutron production rather than dose equivalent in tissue. When evaluating neutron fluence for shielding considerations, the NCRP recommends measuring the neutron fluence with a closed jaw configuration to maximize neutron production.²⁴ Mao *et al.* found that the total number of neutrons produced in the accelerator head is a maximum when the jaws are closed, but the current study found that the dose equivalent in the patient plane was actually slightly higher for open fields because there is no attenuation from head shielding. Practically, for shielding considerations, the variations in the neutron fluence as the field size changes are small, except within the primary photon field.

Figure 8 shows the impact of SSD on neutron dose equivalent for a 10×10 cm² field. At each distance from the central axis, the dose equivalent increased with decreasing SSD and followed closely (within 2%) an inverse square relationship from the bremsstrahlung target. This relationship is not entirely expected because the bremsstrahlung target is not the major source of neutrons; rather, they come primarily from the primary collimator and jaws,^{3,4} which are at different distances from the calculation point than the target, and therefore should not vary exactly in an inverse square relationship from the target. Furthermore, scattered neutrons from the treatment vault also contribute, but these neutrons

FIG. 9. Neutron PDDE curves for neutrons incident on a phantom at 100 or 90 cm SSD from a treatment field either 10×10 or 0×0 cm² and measured either on the central axis (0), at 25 cm, or at 45 cm away from the central axis. Legend: SSD, field size, distance from the central axis.

have a uniform distribution¹ and are therefore also not expected to vary in an inverse square relationship from the bremsstrahlung target. The statistical uncertainty in each calculated dose equivalent was on average approximately 1%. As an example, error bars are shown for the 100 cm SSD scenario.

Neutron PDDE curves were generated by normalizing the neutron dose equivalent to that at a depth of 0.1 cm. Such curves are shown in Fig. 9 as a function of SSD, field size, and distance from the central axis. All of the curves follow one of two trends. The first and slightly higher trend was observed if the neutron PDDE was calculated within the primary photon field (90 or 100 cm SSD, 10×10 cm² field, and 0 cm from the central axis). The second trend, comprised of slightly lower but overlapping PDDE curves, was observed if the PDDE was calculated outside of the primary photon field regardless of field size, SSD, or distance from the field edge. Although factors such as SSD and field size affected the absolute neutron dose equivalent (as illustrated in Figs. 7 and 8), it did not affect the PDDE curve provided the PDDE was evaluated outside the primary photon field.

IV. DISCUSSION

The presence of tissue, as compared to in air, substantially perturbed the neutron field. In particular, it produced a large amount of scatter that increased the total fluence by over 60% and decreased the average energy. This increase in fluence will be detected during neutron measurements made *in vivo*. Bubble detectors, for example, are generally calibrated in air for the dose equivalent that the neutron field produces in a phantom,²⁵ that is, the additional scattered neutrons are accounted for in the calibration. If this detector is used *in vivo* with the standard calibration, it will both directly measure the scattered neutrons and simultaneously account for them through the calibration. That is, it will double count these scattered neutrons and will therefore overrespond. Caution and careful evaluation of the detector cali-

FIG. 10. Neutron PDDE curves interpolated from D'Errico *et al.* (Ref. 15) for 0.24 MeV monoenergetic neutrons normally incident on water ("0.24 MeV") and based on data from the current study for polyenergetic neutrons produced by a medical accelerator and scattered throughout a treatment vault incident at all angles on tissue ("Spectrum").

bration are therefore required when conducting *in vivo* neutron measurements.

Within the phantom, the neutron dose equivalent decreased sharply as a function of tissue depth as the neutron energy and fluence decreased. This sharp decrease in neutron dose equivalent emphasizes the importance of considering an appropriate depth in tissue when evaluating the neutron dose equivalent delivered to a particular point in the patient's body. The PDDE curve for neutrons is much steeper than that for clinical photon beams (dose equivalent at 10 cm depth $[H_{10}] = 5.3\%$ compared to 67% for a 6 MV beam or 56% for a Co-60 beam). This is particularly important for studies that evaluate the neutron dose equivalent to a patient based on measurements made in air or on the surface of a patient. Because the neutron dose equivalent decreases by approximately 15% per cm of depth in tissue, it is clear that appropriate attention to the depth of organs of interest is essential in estimating the neutron dose equivalent. The uniformity of PDDEs for neutrons, shown in Fig. 9, demonstrates that depth can be accounted for without undue burden as the PDDE does not depend on the treatment parameters of field size, distance from the edge of the field, or SSD. It should be noted that this work was based on a Varian accelerator, and for the same nominal photon beam energy, different accelerator manufacturers have slightly different accelerating potentials that result in different neutron fluences.²⁶ However, although there are differences in fluence, the spectra are similar, even for different nominal beam energies,²⁶ so the results of this study should remain generally true.

One of the complicating factors in neutron dosimetry is that the neutron spectrum is often not known in detail. Instead, the average neutron energy may be known or estimated. It is of interest to know how a neutron PDDE based on monoenergetic neutrons compares to a PDDE curve based on a detailed spectrum. To this end, PDDE curves are shown in Fig. 10 for both cases. The detailed spectrum was taken from the current study based on a polyenergetic distribution

of neutrons. A monoenergetic PDDE curve was calculated from the previously published work by D'Errico *et al.*,¹⁵ who calculated PDDE curves for several energies of monoenergetic neutrons normally incident on a water tank. These data were interpolated to yield a PDDE curve from monoenergetic 0.24 MeV neutrons (the average energy of the neutrons in air as calculated in the current study). At shallow depths (<4 cm), the PDDE from the current study was within 10% of that calculated for monoenergetic neutrons of the same average energy. At greater depths (>4 cm), the PDDE from the current study was higher than that calculated for monoenergetic neutrons. At 10 cm depth, the neutron PDDE was greater by a factor of 2, and at 20 cm depth, the neutron PDDE was greater by a factor of 10 as calculated in the current study as more accurate spectral information includes the presence of high-energy neutrons that deposit dose at deeper depths than 0.24 MeV monoenergetic neutrons. Although relatively large discrepancies were observed when using monoenergetic neutron PDDEs, these discrepancies occurred at relatively deep depths where the dose equivalent is small. Therefore, for many applications, the use of a monoenergetic PDDE curve based on the average neutron energy may be adequate.

The quality factors and dose equivalents calculated throughout this study address only the neutrons themselves, not secondary gammas that are generated during neutron capture events. In evaluating the total dose equivalent to a particular point in the patient, secondary gammas must be accounted for as they may deposit a comparable dose equivalent to the neutrons, particularly at deeper depths.⁹ Generally, in an evaluation of the total dose equivalent, these secondary gammas are accounted for through photon measurements carried out using a separate, photon-specific dosimeter.^{5,7,8} Because a photon-specific dosimeter does not distinguish between the secondary gammas arising from neutron capture and photons due to head leakage or patient scatter, all photon doses are generally bundled together. For such circumstances, it would be incorrect to bias the quality factor with photons that were already accounted for as part of the photon measurements. The need to know the "neutron-only" dose equivalent as presented in this work is a separate and important step in determining the total dose equivalent to a point within the patient's body.

V. CONCLUSIONS

The neutron spectrum, average energy, and dose equivalent all changed substantially as a function of depth in tissue. Therefore, our findings indicate that studies attempting to determine neutron dose equivalent within the patient need to examine neutrons as they exist at a depth in tissue of the organ of interest. In addition to illustrating the importance of considering an appropriate depth in the patient, the data in the current study may also facilitate correcting common neutron measurements so the neutron dose equivalent may be determined with more accuracy. For example, dose equivalent measurements at the surface of a patient (such as with the McCall method²) may be corrected by applying an ap-

propriate PDDE curve. Such steps would ensure greater accuracy in the determination of neutron dose equivalents in a patient.

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