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A review on photoneutrons characteristics in radiation therapy with high-energy photon beams

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ABSTRACT

In radiation therapy with high-energy photon beams ($E > 10$ MeV) neutrons are generated mainly in linacs head thorough (γ,n) interactions of photons with nuclei of high atomic number materials that constitute the linac head and the beam collimation system. These neutrons affect the shielding requirements in radiation therapy rooms and also increase the out-of-field radiation dose of patients undergoing radiation therapy with high-energy photon beams. In the current review, the authors describe the factors influencing the neutron production for different medical linacs based on the performed measurements and Monte Carlo studies in the literature.

1. Background

In spite of tremendous developments in cancer treatment methods, radiation therapy using medical electron linear accelerators (linac) still plays an unparalleled role in palliation and treatment of tumors. In radiation therapy with photon beams ($E > 10$ MeV) neutrons are generated mainly in linacs thorough (γ,n) interactions of photons with nuclei of high atomic number materials constituting the linac head and the beam collimation system.[1–3] On the other hand, high-energy photon interactions with patients and treatment room wall could be the other sources of photoneutrons in radiation therapy. Different reaction mechanisms like giant dipole resonance (GDR), quasi-deuteron (QD) delta resonance (DR), etc. are involved in the production of photoneutrons.[4] The neutrons emitted from the GDR mechanism are similar to the evaporation neutrons from a compound nucleus while the QD neutrons have been compared with the pre-equilibrium model. GDR neutrons are of low energy with an isotropic angular distribution.[4] In high-energy accelerators where the photon energy and intensity is high compared to the neutrons, it is difficult to experimentally measure the direct photoneutron component.[5,6]

The NCRP 116 recommends a quality factor of 20 for photoneutrons energy of 0.1–2 MeV which is produced in radiation therapy with photon beams.[7] They are highly penetrating particles with high radiobiological effectiveness (RBE). Their contribution in patient out-of-field dose is smaller than scattered photons but considering their quality factor of 20 gives a significant contribution in patient effective dose and consequently in radiation-induced fatal cancer risk.[8,9]

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2. Aim

In this article, we have tried to review the published studies on photoneutron production in different linacs, including Varian, Elekta and Siemens. Additionally, the effect of different factors on photoneutrons dosimetric features has been reviewed.

3. Materials and methods

3.1. Photoneutron production in linac head

The main sources for photoneutrons in a linac head are high atomic number components, including target, primary collimator, secondary collimators, wedges, blocks and multi-leaf collimators. The tungsten (W) and lead (Pb) with high cross sections for (γ,n) reaction are major sources of photoneutrons in medical linacs. Although, other elements such as iron, copper and aluminum are present, their probability for neutron production is negligible. For example, only ^{56}Fe atoms can produce neutrons among the iron atoms. For quantitative comparison, it can be said that the energy threshold of photoneutron production for W and Pb are 6.74 and 6.19 MeV, respectively. Whereas respective thresholds for Cu and Fe have been reported to be 9.91 and 7.65 MeV.[10] On the other hand, the probability of photoneutron interactions increases steeply with photon energy and its maximum value has been found in the therapy range of 13–18 MeV photons for the materials used in the linac head including W, Pb, Cu and Fe.[2,11]

The other noteworthy point is that the main neutron producing materials, tungsten and lead, have very low absorption cross sections for the energy range of neutron produced in linac head. Therefore, photoneutrons have a great chance to penetrate the shielding and reach the patient and bunker walls.

Over recent three decades, many studies have reported neutron dosimetry using different types of dosimeters.[8,12–22] However, the experimental methods have not been able to analyze the origin of neutrons reaching the dosimeter. The proposed method of choice to tackle this problem have been the Monte Carlo methods.[23] By modeling different components of a linac and initiating the primary electron striking on target and then following the history of all particles including photons, electrons and neutrons in different parts of a linac until its death, all information about the interactions and number of generated particles as well as deposited energies in different parts or any defined volume can be tallied and provided at the end of simulation.[24] In several MC studies the contribution of different parts of a linac in neutron fluence received by patient or at the isocenter has been calculated for some commercial linacs.[5,10,23,25,26]

MC methods have been employed extensively to evaluate the photoneutron characteristics in radiation therapy.[26–30] MC studies have shown that neutron source strength or Q value varies with linac model, location of scoring cell, field size and modeling geometry.

Using the MC methods, Pena et al. calculated the contribution of different components of a Primus linac operating in its 15 MeV photon beam for 10 cm × 10 cm field size.[26] The contribution of different components in neutron source strength were reported as primary collimator 52%, secondary collimator jaws 21%, target 12%, Multi-leaf collimator (MLC) 6.6%, shielding 5%, flattening filter 0.41%. A recent study by Becker et al.[13] on the same linac showed results consistent with the previous study of Pena et al. Comparing the results of both studies with the results obtained on a Varian 2100C/2300C linac by Mao et al., Zanini et al. and Howell et al. reveals that in both linacs the overall trend of contribution are identical but there are small discrepancies for different components.[30–32] The results were summarized in Table 1. As seen in Table 1, the target and flattening filter in Varian linac shows higher contribution of 25% and 75% relative to Primus. However, it can be concluded from different MC studies that the primary collimator which is made from tungsten alloys has the highest contribution among different components.[26,32,33] The second contributor is secondary collimator jaws and then target, multi-leaf collimator, shielding and flattening filter as other contributors.

The commonly used quantity for neutron production in different linacs is neutron source strength, Q, which is defined as the number of neutrons at the isocenter coming from linac head per X-ray dose delivered at the isocenter.[2] Neutron source strength of a linac is an important factor used in the neutron dose calculations for both shielding purposes and patient out-of-field dose calculations. A most complete data set of neutron source strength was provided by an experimental study of Followill et al.[17] The results of different studies on neutron source strength of commercially used linacs were summarized in Table 2. A review of Table 2 indicates that the neutron strength for different linacs depends on the photon energy and linac head structures as well as a model. Furthermore, it should be noted that a wide range of Q values reported for a specific model and photon energy result from large uncertainties in neutron measurement methods. Differences in MC modeling and application of different codes for MC calculations could also account for the observed discrepancies.

In the study of Mao et al. the neutron strength of Varian Clinac 2100C/2300C for different energies was estimated

Table 1 – MC calculation of component contribution in photoneutrons from Varian 2100C/2300C linac.[32]

Component	20 MeV	18 MeV	15 MeV	10 MeV
Target	17.2% (W,Cu)	16% (W,Cu)	9% (W,Cu)	0.01% (Cu)
Primary collimator	36% (W)	41% (W)	38% (W)	45% (W)
Flattening filter	10.4% (Fe,Ta)	9% (Fe,Ta)	22% (W)	0.03% (CU)
Jaws	36% (W)	35% (W)	29% (W)	56% (W)
Others (magnet, shielding, etc.)	1%	1.4%	1.2%	1%

Table 2 – Comparison of neutron source strength for various medical linacs.

Manufacturer	Model	Energy (MeV)	Q ($n\,Gy^{-1}$)	Study
Siemens	KD	20	0.92×10^{12}	McCall (1987)[2]
Siemens	Primus	15	0.20×10^{12}	Lin et al. (2001)[47]
Siemens	Primus	15	0.17×10^{12}	Pena et al. (2005)[26]
Siemens	Primus	15	0.136×10^{12}	Becker et al. (2007)[13]
Siemens	MD2	10	0.08×10^{12}	Followill et al. (2003)[17]
Siemens	MD	15	0.2×10^{12}	Followill et al. (2003)[17]
Siemens	KD	18	0.88×10^{12}	Followill et al. (2003)[17]
Siemens	Primus (with MiMIC)	10	0.02×10^{12}	Followill et al. (2003)[17]
Siemens	Primus (with MiMIC)	15	0.12×10^{12}	Followill et al. (2003)[17]
Siemens	Primus (with MLC)	15	0.21×10^{12}	Followill et al. (2003)[17]
Varian	1800C	18	2.9×10^{12}	McCall (1987)[2]
Varian	20C	15	0.93×10^{12}	McCall (1987)[2]
Varian	18C	10	0.059×10^{12}	McCall (1987)[2]
Varian	1800C	18	2.27×10^{12}	McGinley and Landry (1989)[56]
Varian	1800C	15	1.23×10^{12}	McGinley and Landry (1989)[56]
Varian	1800C	10	0.06×10^{12}	McGinley and Landry (1989)[56]
Varian	2100C/2300C	20	1.2×10^{12}	Mao et al. (1997)[32]
Varian	2100C/2300C	18	1.2×10^{12}	Mao et al. (1997)[32]
Varian	2100C/2300C	15	6.8×10^{11}	Mao et al. (1997)[32]
Varian	2100C/2300C	10	3.8×10^{10}	Mao et al. (1997)[32]
Varian	2100C	18	0.96×10^{12}	Followill et al. (2003)[17]
Varian	2100C (with MLC)	18	0.87×10^{12}	Followill et al. (2003)[17]
Varian	2300 CD	18	0.95×10^{12}	Followill et al. (2003)[17]
Varian	2500	24	0.77×10^{12}	Followill et al. (2003)[17]
GE	Saturne 43	25	2.4×10^{12}	Fenn and McGinley (1995)[58]
GE	Saturne 43	18	1.5×10^{12}	Fenn and McGinley (1995)[58]
GE	Saturne 41	15	0.47×10^{12}	Fenn and McGinley (1995)[58]
GE	Saturne 41	12	0.24×10^{12}	Fenn and McGinley (1995)[58]
Elekta	SL-20	17	0.69×10^{12}	McGinley et al. (1993)[57]
Elekta	SL-25	22	2.37×10^{12}	McGinley et al. (1993)[57]
Elekta	SL-20	18	0.46×10^{12}	Followill et al. (2003)[17]
Elekta	SL-25	18	0.46×10^{12}	Followill et al. (2003)[17]

for closed jaws.[32] The neutron strength was increased significantly from 10 to 18 MeV and it reached from 3.8×10^{10} to 1.2×10^{12} neutrons per Gy at the isocenter. While in the other study on Primus linac for the 15 MeV photon beam, the neutron source strength was calculated at 0.17×10^{12} n Gy^{-1}.[26]

Flattening filters are often made of medium atomic number materials such as iron and copper which do not have considerable cross sections for photoneutron production.[34,35] Removal of flattening filters from linacs has been proposed and studied in several investigations as a way to increase the dose rate for radiosurgery and intensity-modulated radiation therapy (IMRT) treatments.[25,36–40] It was shown that neutron fluence for a 18 MeV beam of Varian linac was about 69% lower for a flattening filter free beam measured by gold foil activation in neutron moderators.[39] Another MC study by Mesbahi on Elekta SL-25 showed results very close to those of a Varian linac. It was found that removing a flattening filter decreases the number of photons produced in the target which is required for a given dose at the isocenter. So, the neutron fluence for a flattening filter free beam was on average 54% lower compared to the original beam. It is believed that by decreasing the photon produced in the target, the interactions between photons and upper stream components are diminished. Consequently, the number of neutrons per monitor unit (MU) is reduced because of the decrease in the number of (γ,n) interactions in the primary collimator and other contributing components.[25]

3.2. *Photoneutron fluence and dose equivalent around a medical linac*

Radiation protection calculations and measurements for photoneutrons around medical linacs are aimed to protect both patients and staff from unwanted radiation. In addition to photoneutrons generated in linac head, when the photon beam of a linac is irradiating the patient, primary, scattered and leakage photons inside the therapy room have the chance for photoneutron production through interactions with patient and concrete wall of the room. The energy spectra of photoneutrons are characterized by two components: a peak around 1 MeV, due to the nucleus evaporation, and a bump in higher energy region due to direct reaction. The mean energy of photoneutron is around several MeV. Meanwhile, neutron spectra shape is changed into more complex distribution due to transmission through the head components and shielding at the patient plane.[5]

According to IAEA 47, the neutron fluence at the isocenter Φ (n cm^{-2}) per 1 Gy of X-ray is as follows:[41,42]

$$\Phi = \Phi_{\rm dir} + \Phi_{\rm sc} + \Phi_{\rm th} = \frac{aQ}{4\pi d^2} + \frac{5.4aQ}{S} + \frac{1.26Q}{S} \qquad (1)$$

The $\Phi_{\rm dir}$ denotes the direct neutron fluence, $\Phi_{\rm sc}$,the scatter neutron fluence, $\Phi_{\rm th}$, the thermal neutron fluence, Q, the neutron source strength per Gy of X-ray at the isocenter, a, the transmission factor for linac head which is 1 for Pb and 0.85 for W. In addition, d is the distance (cm) between the measured point and the target and S is the area of the treatment room (cm^2).

The early measurements using indium activation foil showed that for a 24 MeV of a Clinac 2500 photon beam, the effective energy of neutrons were from 0.28 MeV at the isocenter under closed collimators to 0.016–0.024 MeV in the maze entrance.[43]

In a study on the neutron dose equivalent at the maze entrance on different rooms for 15 and 18 MeV photon beams by indium foil measurements, it was reported that the neutron dose equivalent ranged from 0.79 to 1.6 (mSv n/Gy x) for Varian linacs, 0.17 to 1.24 for Siemens linacs. It was also found that for 13 studied linacs the neutron dose rose by 13% with decreasing the field size from maximum to zero.[1]

Different studies have reported controversial results concerning the effect of a field size on neutron doses around medical linacs. Garnica-Garza pointed out in his study that neutron yield increases with field size and the difference between the smallest and largest field size was in the order of 25%. In another study, Kim et al. reported that the neutron dose decreases with field size for field sizes greater than 20 cm × 20 cm. They compared their calculated fluence and mean energy with the early work of Kase et al.[44] with EGS4 code. Their calculated fluence and mean energies were almost 2 times higher. It can be attributed to the cross section files used in EGS4 code and renewed cross section files (LA150U) used in MCNPX code. According to Kim et al. neutron yield was higher in 20 cm × 20 cm and they suggested using neutron yield of this field size for design and calculation of door shielding for medical accelerator facilities. The study of Chibani et al. showed that the neutron fluence increased with increasing field size for Primus 18 MeV, Varian 15 and 18 MeV photon beams.[45] However, the field sizes greater than 20 cm × 20 cm were not studied and it seems that the statistical uncertainty of more than 5% have influenced their final conclusion.

A study on photoneutrons of 10 and 15 MeV photon beams of the Varian 2100C/2300C linac, showed that for the middle size of the irradiation fields, a 20 cm × 20 cm, maximum dose equivalent was seen and neuron dose decreased with field sizes higher than 20 cm × 20 cm.[46] Additionally, another study found that the neutron yield increases linearly as the field size decreases.[32]

Chibani et al. showed that for a 18-MeV Siemens linac the maximum neutron-to-photon ratio was 1.4×10^{-4} at the surface. They concluded that the neutron flux and dose at the given depth increases with increasing field size. Their reason was that the bulk of neutrons come from the upper part of a linac not from jaws and the neutrons produced within jaws are likely to be stopped locally because of jaw thickness. The neutron dose equivalent (Sv/Gy) of 2.82×10^{-3} and 6.96×10^{-3} were calculated for 5 × 5 and 20 × 20 field sizes, respectively. The neutron dose equivalent for Varian was 4 times higher than linacs for 18 MeV photons. They related these differences mainly to primary electron energy, which was 14 MeV for 18 MeV photon beam of a Siemens linac, and to a smaller extent to differences in linac geometries and materials.

In a study by Lin et al.[47] on a Primus linac the neutron source strength was measured with cone size of 25 × 25 for 15, 18 and 21 MeV electron beams. The neutron dose for 12 MeV electrons was not detected and neutron strength of 100, 262, 349 μSv Gy^{-1} electron dose was measured for 15, 18 and 21 MeV beam, respectively. Comparing the neutron source strength for photon and electron beam revealed that for the 15 MeV energy the neutron strength for a photon beam is almost 18 times higher than that of electron beams. In the study, neutron production from a mobile linac used for intraoperative radiation therapy was evaluated.[48] The neutron dose equivalent was measured using bubble detector for 12 MeV electron beam of a Mobetron linac. It was reported to be 0.33 μSv Gy^{-1} at the accelerator head, 0.18 μSv Gy^{-1} in the patient plane and 0.31 μSv Gy^{-1} at the floor plane, on the beam axis and under the beam stopper. They concluded that neutron dose equivalent from a Mobetron linac was at least one order of magnitude lower than that produced by a conventional linac operated at the same energy in electron mode.

3.3. *Photoneutron spectra*

Photoneutron spectra in different points around the linac can be calculated by MC methods. But the neutron spectra measurements are sophisticated and vulnerable to different measurement inaccuracies. In study by Ongaro et al., a commercial passive neutron spectrometer was used to obtain the neutron energy spectra in patient plane.[5] The spectrometer consisted of several bubble detectors with different energy thresholds ranging from 10 keV to 10 MeV. Each detector included a polycarbonate vial filled with elastic tissue-equivalent polymer and superheated freon drops were dispersed inside the gel. Because of metastable state of freon, neutron interactions with dosimeter generates bubbles in which the number of bubbles trapped in polymer is proportional to the neutron fluence. The neutron spectrum can be obtained by the unfolding process.

In the study of Garnica-Garza, it was found that the spectra and mean energy were insensitive to field size and it differed within 1%.[49] They concluded that the photoneutron production takes place in the components above the movable jaws and opening the jaws enables more neutrons to reach the isocenter.

Some studies stated that linac elements which do not take part in therapeutic beams can be important sources in neutron production.[26,46,47] In a Mc study on photoneutron production for a 15 MeV Primus linac, the neutron spectra was calculated at different locations in the treatment room (Fig. 1).[26] The detailed geometry of linac head including the shielding and other components which was not routinely simulated in photon beams in other studies was considered in their study; they also simulated a conventional treatment room to study the effect of walls on the neutron spectra. The approximate results can be seen in Fig. 1. Neutron spectra at the isocenter and room corner represent two peaks. There is a peak from 0.1 to 5–10 MeV with its maximum on 0.7 MeV for fast neutrons. Another peak also exists below 1 eV and is centered on 0.05 eV. It is seen that the epithermal neuron fluence does not vary significantly inside the treatment room as well as maze. While for fast neutrons the neutron fluence is more than 7 times higher for isocenter relative to the other locations in treatment room.

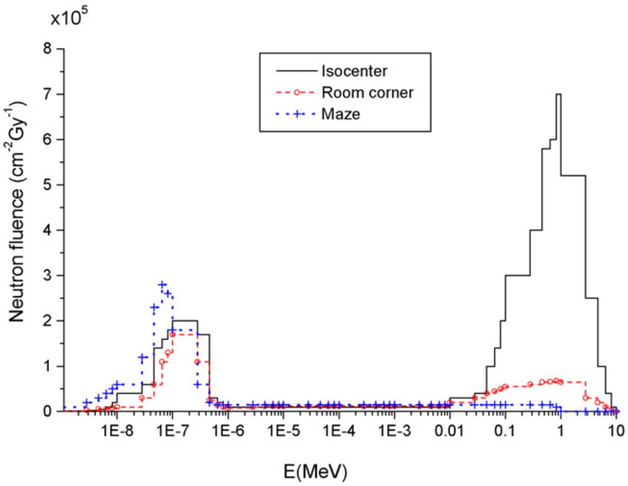

Fig. 1 – Neutron spectra at different points of a typical treatment room for 15 MeV photon beam of Primus linac (the spectra was derived from the study of Pena et al.).[26]

Several studies on photoneutron spectra in a treatment room have shown that the thermal neutron spatial distribution is homogenous, while epithermal neutron fluence reaches its maximum value in the vicinity of the target, primary collimator and magnet. Additionally, fast neutrons are predominant part of neutron spectra at the isocenter and are decreased with the inverse square of distance from the linac head.[17,50]

3.4. *Depth dose equivalent of neutrons produced in radiotherapy*

The neutron attenuation is faster outside the photon beam. For photon beams from 10 to 18 MeV the depth of 50% maximum dose equivalent,$d_{H50,}$ range is from 7.5 to 8.5 cm. If the d_{H50} values are compared to those calculated for monoenergetic neutrons, the effective neutron energy can be calculated. d'Errico et al. found the effective neutron energy of 1.8–2.1 MeV for their studied photon energy.[15] When the neutrons transmit through head shielding they degraded and their mean energy dropped to the range of 0.3–0.8 MeV.[5,14,15,25,26,46] Kim et al. worked on a 2100C/2300C Varian linac and found that mean energy of neutrons are from 0.38 to 0.45 MeV for 10 and 15 MeV beams considering 0 cm × 0 cm field size.

3.5. *Effect of multi-leaf collimator in IMRT*

Recent studies have revealed that using newly developed collimation systems including MLC and other beam collimating accessories influences the neutron fluence in photon beams depending on their position in the beam.[30,51] For MLC-based treatments to account for the increase in MU relative to conventional treatments, modulation scaling factor (MSF) has been used.[45,52] Neutron increase in these techniques is proportional to increase in MU and so in MSF. Also it is related to neutron leakage dose in comparison with conventional treatments assuming that both treatment modalities deliver the same dose to the target.

It is assumed that for both IMRT and conventional treatments the tumor dose is almost the same. But for IMRT, it is required that the dose be delivered in small segments which are formed by MLCs and in spite of the conventional treatments, many segments were used to deliver dose to a large treatment volume. So the number of MUs in IMRT is inevitably increased. It has been shown that higher MU results in increased photoneutron compared with conventional treatments. In the measurements by TLD and Bonner sphere system for a prostate treatment, neutron dose equivalent for IMRT was approximately 6 times higher than that of four field conventional beams. This can be related directly to the increased MU of 6 times for the IMRT treatment.[31]

3.6. *Effect of wedge filter on photoneutron fluence*

Recently, there have been several studies on the effect of wedge filter on the photoneutron contamination of photon beams and its spatial distribution around a linac head.[53–55] In the study of Hashemi et al. using polycarbonate film dosimeters it was found that for 18 MeV photon beam of an Elekta linac, the neutron dose equivalent (NDE) of wedged beam was 3–5 times higher than open beams depending on field size. Also they showed that NDE increases with field size for both open and wedged beams.[54] In the MC study of Mesbahi et al. on the same linac, the results revealed that the NDE is on average 6.5 times higher for wedged beams.[55] The effect of field size on neutron dose was different for open and wedged beams. The neutron dose decreased with field size for open beams while it increased with field size for wedged beams at the patient plane. For the point on the central axis, the NDE decreased from 1 to 0.6 mSv Gy^{-1} X-ray with field size variation from 5 cm × 5 cm to 30 cm × 40 cm. But for a wedged beam, it raised from 4 to 7 mSv Gy^{-1} X-ray for the same field sizes. It was pointed out that introducing the high Z wedge filter into the pathway of high-energy photons would lead to an increase in the number of photoneutrons. On the other hand, using the wedge filter, the photon fluence reaching the d_{max} is decreased by a rate which equals to the wedge factor. So, to compensate for the attenuation effect of the wedge filter, the MUs required to produce a constant dose at the d_{max} are increased and cause more photoneutron production for wedged beams. Another effect could be the increase of backscattered photons and their interactions with head shielding and components, which causes more leakage of photon and neutron through the head shielding. It is recommended that the presence of higher photoneutron fluence for wedged beams should be taken in account when the patient received dose and secondary cancer risk from radiation therapy are calculated. Moreover, it is required to establish more conservative design of accelerator room and door shielding for neutrons to meet radiation protection guidelines.[53,55]

4. Conclusion

In the current study the neutron production in high-energy photon beams used in radiation therapy was reviewed. A number of studies have reported different neutron source strength for different linacs and in some cases for the same model and photon energy. It can be accounted for by uncertainties associated with measurement methods and difference in MC modeling of medical linacs with different MC codes. Studies also showed that application of MLC and wedge filter increases neutron production in high-energy photon beams. Moreover, removing the flattening filter from photon beams caused a significant decrease in neutron production.

International guidelines on shielding against photoneutrons are based on the developing knowledge on the photoneutrons characteristics. However, the current knowledge on neutrons produced in linacs is in increasingly using new neutron dosimetry techniques which will overcome the uncertainties associated with neutron dosimetry. On the other hand, application of Monte Carlo methods provides more precise information about the photoneutron properties in radiation therapy. In the current review, we have considered the new published information on photoneutron characteristics which might have direct impact on calculating the required shielding barriers and patient out-of-field dose in radiation therapy.

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