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A COMPARATIVE STUDY ON 6 MeV PHOTON BEAM PERCENTAGE DEPTH DOSE OF VARIAN CLINAC 2300 C/D, ELEKTA SYNERGY PLATFORM, AND SIEMENS PRIMUS LINACS*

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Abstract. This work presents a comparative study of the 6 MeV percentage depth dose curves for photon beams of VARIAN Clinac 2300 C/D, ELEKTA Synergy Platform, and SIEMENS Primus Linacs. The measurements were performed for a set of 7 field sizes (5×5, 10×10, 15×15, 20×20, 25×25, 30×30, 40×40) cm^2, keeping the same conditions (*e.g.*, geometry, ion chamber voltage and polarity, incremental step and direction). During our study we used the Mephysto MC^2 software and two semiflex ion chambers from Physikalisch-Technische Werkstätten GmbH, calibrated in water under ^{60}Co gamma ray beam.

Key words: 3D conformal planning systems, percentage depth dose, treatment quality.

1. INTRODUCTION

Linear accelerators, available with a wide range of capabilities, have become the common treatment machines in most radiation oncology departments. In Romania, in the last few years, the radiotherapy services have been significantly improved due to installations of new equipments provided with 3D conformal planning system from VARIAN, ELEKTA, and SIEMENS. From the physicist point of view, a very important issue is the homogeneity of the beam parameters. Regarding this matter, we are bringing into discussion the possibility of obtaining a similar treatment quality in all radiotherapy centres by providing a very useful database for the commissioning process and a comparison between beam parameters for all megavoltage medical units already mentioned.

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2. GENERAL CONSIDERATION

Linear accelerators may produce photon beams, electron beams or both depending on the model used. Photons are produced with a broad spectrum of energies, ranging up to the maximum energy of electrons that strike the target, with a peak at approximately half this maximum value. In principle, the full energy spectrum should be known, in order to derive the energy dependent factors necessary for the determination of the absorbed dose. The beam quality is dependent also on the target design. All accelerators are provided flattening filters, which have the secondary effect of altering the energy spectrum. The magnitude of this effect depends on the material used and its thickness. In addition to these factors there are also other factors that alter the spectrum and magnitude of the secondary radiation beam, as is the design of the collimation system. For these reasons, the most modern protocols specify energy directly, in terms of the attenuation properties of the beam [15].

The parameter applied, in most protocols, sometimes referred to as the quality index, Q_i, is the ratio of the chamber reading, at a depth of 20 cm, in a water phantom, to the reading at a depth of 10 cm, for the same fixed source to chamber distance, SCD, for a field size of 10×10 cm^2 at the chamber level. This is actually a tissue-phantom ratio (TRP_{10}^{20}). An alternative quality index, referred to in some older protocols, is defined for a fixed source-to-skin distance, SSD, of 100 cm, with the surface field size defined at 10×10 cm^2. The quality index for this definition is the ratio of depth doses on the central axis, at 20 cm and 10 cm, respectively (D_{20}/D_{10}) [9,10].

The design of the collimation systems is different for all three linacs. For example, the VARIAN multileaf collimator, MLC, is positioned as a tertiary system below the standard lower jaws (Fig. 1). Placing the multileaf collimator in a position closer to the patient, then either the SIEMENS or ELEKTA collimators provides both positive and negative results. A negative aspect of this approach is that a collimator nearer the patient will have a greater overall bulk, especially, in the direction of leaf motion. This is because beam divergence requires a larger system to cover the same (40×40) cm^2 field size. A simple calculation, using the geometry shown in Figure 1, demonstrates the difference. A comparison can be made for a hypothetical design that allows the leaves of a multileaf collimator to reach a total of 15 cm across a field midline and retract to the outer edge of a 40 cm wide field.

For the VARIAN geometry where the collimator is mounted in a tertiary position with the bottom of the collimator at l_1 = 53.5 cm, the total width of the collimator system must be about 60 cm. For ELEKTA l_2 = 50.9 cm and for SIEMENS l_3 = 37.9 cm. Using the geometry shown in the figure for the ELEKTA collimator (lower surface of the multileaf collimator at 37.3 cm), this number is reduced by almost 20 cm. In fact, as discussed below, the VARIAN design is modified to bring the dimension to a size that is roughly equivalent to the ELEKTA number. One positive result of mounting the multileaf collimator further from the photons target and nearer the patient is that leaf width is larger. This offers distinct advantages in terms of manufacturing because it simplifies issues like the machining of the tongue and the groove into each leaf. The slight separation between the side surfaces of neighbouring leaves, needed to avoid friction, is easier to achieve when the multileaf collimator is mounted lower in the treatment head.

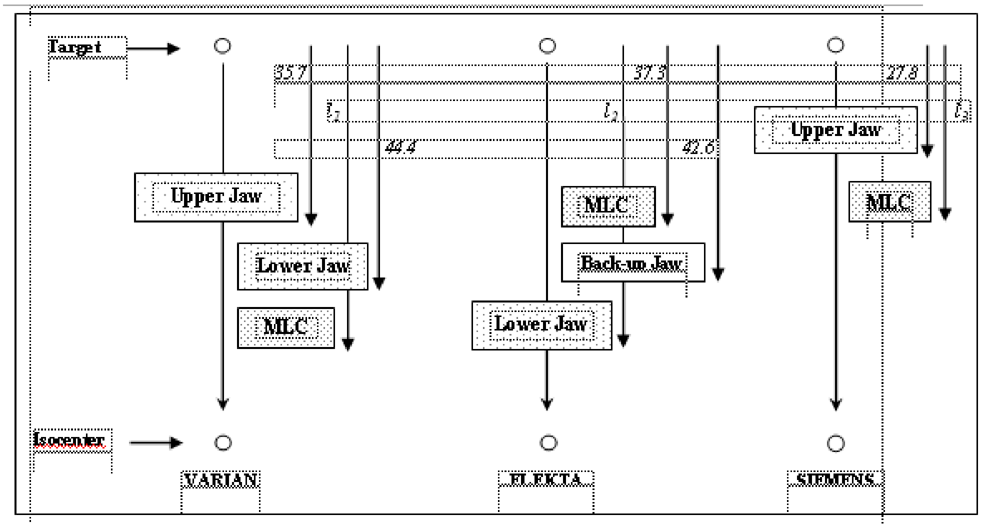

Fig. 1 – Schematic geometry for VARIAN, ELEKTA and SIEMENS collimator systems (Modified, after, [5]). MLC = multileaf collimator.

Also, maintaining a 1.0 mm position accuracy for the leading edge of a leaf at isocentre translates to positioning accuracy of 0.37 mm at the multileaf collimator position for the ELEKTA geometry, while a 0.54 mm accuracy is needed for the VARIAN placement of the collimator. A disadvantage of mounting the leaves nearer to the photon target brings the between-leaf leakage regions closer together and can cause the transmission peaks to join so that average radiation leakage for the MLC is increased. This is seen for the apparent increased through-the-leaf leakage for the ELEKTA collimator compared to the VARIAN one. This is not the expected result due to significantly greater thickness of the ELEKTA collimator (7.5 cm compared to 5.5 cm for VARIAN) in the beam direction. It should be pointed out that this effect is also dependent on the spot size for the particular radiation beam and is not entirely due to the position of the collimator [5].

Considering all this facts, we present a comparative study of the 6 MeV PDD curves for photon beams of VARIAN Clinac 2300 C/D, ELEKTA Synergy Platform and SIEMENS Primus linacs, and how PDD is reflected in clinical outcome.

3. MATERIALS AND METHODS

In order to characterise the central axis dose distribution it has to normalize the dose at depth with respect to dose at a reference depth. The quantity, PDD, may be defined as the quotient, expressed as a percentage, of the absorbed dose at any depth, d, to the absorbed dose at a fixed reference depth, d_0, along the central axis of the beam (Fig. 2).

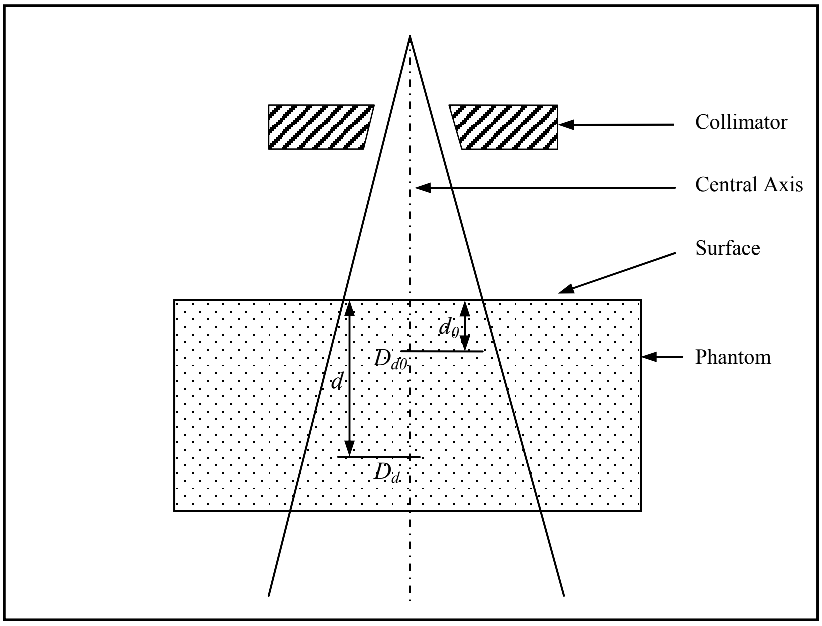

Fig. 2 – Percentage depth dose is $(D_d/D_{d0}) \times 100$ %, where d is any depth and d_0 is reference depth of maximum dose [11].

For high energies of linacs beams, the reference depth is taken at the position of the peak absorbed dose ($R_{100} = d_0 = d_{max}$). In clinical practice, the peak absorbed dose on the central axis is sometimes called the *maximum dose*, the *dose maximum*, or simply, D_{max}. The percentage depth dose, PDD (beyond the depth of maximum dose) increases with beam energy. Higher energy beams have greater penetrating power and thus deliver a higher PDD.

As seen in Fig. 2, the PDD decreases with depth beyond the depth of maximum dose. However, there is an initial build-up of dose which becomes more and more pronounced as the energy is increased. The region between the surface and the point of maximum dose is called the *dose build-up region*.

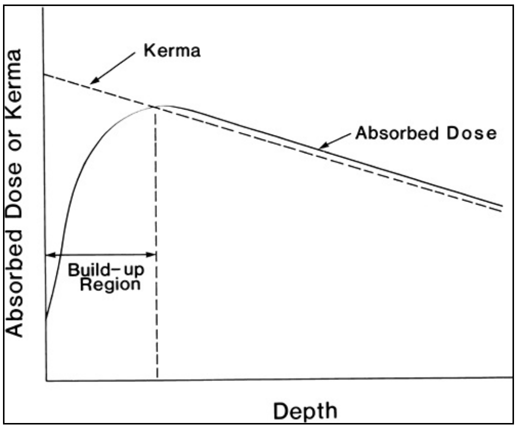

Fig. 3 – Schematic plot of absorbed dose, in a phantom, and kerma as functions of depth [11].

The dose build-up effect of the higher energy beams gives rise to what is clinically known as the skin-sparing effect. For megavoltage beams, the surface dose is much smaller than the D_{max}. This offers a distinct advantage over the lower energy beams for which the D_{max} occurs at the skin surface. Thus, in the case of the higher energy photon beams, higher doses can be delivered to deep tumours without exceeding the tolerance of the skin. As the high energy, photon beam enters the patient or the phantom, while high speed electrons are ejected from the surface and the subsequent layers. These electrons deposit their energy at a significant distance away from their site of origin. Due to these two reasons, the electron fluence and the absorbed dose increase with depth until they reach a maximum. However, the photon energy fluence continuously decreases with depth.

Concerning the field size, this may be specified either by the geometry or dosimetry point of view. The geometrical field size is defined as the projection, on a plane, perpendicular to the beam axis, of the distal end of the collimator as seen from the front centre of the source. The dosimetric or the physical field size is the distance intercepted by a given isodose curve (usually, 50 % isodose) on a plane perpendicular to the beam axis at a stated distance from the source. The field size is also defined at a predetermined distance such as the *source-surface distance* (SSD) or the *source-axis distance* (SAD). The latter term is the distance from the source to gantry rotation axis known as the isocentre.

For a sufficiently small field one may assume that the depth dose at a point is effectively the result of the primary radiation, *i.e.*, the photons which have traversed the overlying medium without interacting. In this case, the contribution of the scattered photons to the depth dose is negligibly small or even zero. But as the field size is increased, the contribution of the scattered radiation to the absorbed dose increases, too. Since this increase in scattered dose is greater at larger depths than at the depth, D_{max}, the PDD increases with increasing field size.

The increase in PDD caused by the increase in field size depends on beam quality. Since the scattering probability or cross-section decreases with energy increase and the higher energy photons are scattered predominantly in the forward direction, the field size dependence of PDD is less pronounced for the higher energy than for the lower energy beams.

Photon fluence emitted by a point source of radiation varies inversely with square of the distance from the source. PDD increases with SSD due to the effects of the inverse square law. Although the actual dose rate at a point decreases with increase in distance from the source, the PDD, which is a relative dose with respect to a reference point, increases with SSD. In clinical radiotherapy, SSD is a very important parameter. Since PDD determines how much dose can be delivered at depth, relative to the surface dose or D_{max}, the SSD needs to be as large as possible. However, in practice, since the dose rate decreases with distance, the SSD is set at a distance which provides a compromise between dose rate and PDD [9].

Our comparative study is based on the 6 MeV PDD curves for photon beams in VARIAN Linac 2300 C/D, ELEKTA Synergy Platform, and SIEMENS Primus Linacs for a set of 7 field sizes (5×5, 10×10, 15×15, 20×20, 25×25, 30×30, 40×40) cm^2, keeping the same conditions (*e.g.*, geometry, ion chamber voltage and polarity, incremental step and direction).

The measuring equipment consisted of a 3D water phantom, Unidos radiation dosimeter, Mephysto MC^2 software, one reference Semiflex ionization chambers and one Semiflex ionization chamber for field measurements, calibrated in water under ^{60}Co gamma ray beam, all this from Physikalisch-Technische Werkstätten GmbH Germany.

4. RESULTS AND DISCUSSIONS

A sinoptic view of PDD parameters highlighted in IAEA TRS 398 protocol (R_{100}, R_{80}, R_{50}, Ds, D_{100}, D_{200}, Q_{i}, NAP at SSD = 100 cm) is presented in Table 1, after dose normalisation procedure at $d_{\max}$ (R_{100}), where: R_{100} is the depth of maximum dose; R_{80} is the depth of 80 % dose value; R_{50}, the depth of half maximum dose value; D_{S}, dose value at surface (0.5 mm beneath surface); D_{100}, dose value at 100 mm depth; D_{20}, the dose value at 20 mm depth; Q_{i}, the quality index; NAP, the Nominal Accelerating Potential [9].

Table 1

The measured values of PDD parameters for VARIAN, ELEKTA, and SIEMENS linacs, based on IAEA TRS 398 protocol [9]

Field Size [cm^2]	R_{100} [mm]	R_{80} [mm]	R_{50} [mm]	Ds [%]	D_{100} [%]	D_{200} [%]	Q_{i}	NAP [MeV]	SSD [cm]	Manufacturer
5×5	16.01	60.53	138.89	49.90	63.45	34.73	-	4.43	100	VARIAN
10×10	16.04	66.83	153.88	50.74	67.29	38.72	0.6691	5.65	100	
15×15	16.03	69.81	162.79	55.94	69.10	41.12	-	6.91	100	
20×20	15.05	72.10	169.82	59.93	70.34	42.92	-	8.28	100	
25×25	14.99	73.52	174.86	63.16	71.02	44.10	-	9.53	100	

Table 1 (continued)

30×30	14.03	74.18	178.23	66.39	71.40	44.89	-	10.66	100	
40×40	15.00	76.29	183.42	71.03	72.40	46.23	-	12.46	100	
5×5	17.01	63.51	144.09	42.76	64.90	36.01	-	4.71	100	ELEKTA
10×10	17.01	68.82	158.09	46.78	68.24	39.77	0.6784	6.07	100	
15×15	17.00	71.79	167.36	50.73	69.98	42.21	-	7.59	100	
20×20	15.99	72.96	172.58	54.82	70.83	43.74	-	9.11	100	
25×25	15.97	74.96	177.70	58.31	71.70	44.99	-	10.48	100	
30×30	15.92	75.60	180.62	61.17	71.99	45.74	-	11.81	100	
40×40	15.94	77.80	186.33	64.70	72.93	47.06	-	13.96	100	
5×5	16.97	61.45	141.12	47.22	63.73	35.20	-	4.61	100	SIEMENS
10×10	16.00	67.32	154.95	51.31	67.37	39.01	0.6735	5.84	100	
15×15	16.04	70.68	164.40	55.01	69.54	41.53	-	7.08	100	
20×20	15.01	71.98	170.35	58.03	70.35	43.18	-	8.66	100	
25×25	14.95	73.71	175.46	61.14	71.25	44.24	-	9.54	100	
30×30	15.00	74.97	178.94	63.30	71.97	45.28	-	10.74	100	
40×40	15.03	77.38	183.93	66.22	72.78	46.43	-	12.33	100	

The first step of the study was to introduce a normalization procedure for PDD curves to create an appropriate method of comparing PDDs for various accelerators, but at the same energy and field size. Figure 4 shows the central axis depth dose for 6 MeV photon beam achieved in VARIAN, ELEKTA, and SIEMENS accelerators for a (10×10) cm^2 field size. These doses are expressed as a percentage of the maximum dose on the beam axis. The shapes of the PDD curves and dose distributions depend on a specific linac as well as on the field size. Usually, for a given machine, the depth of the maximum dose slightly changes with field size.

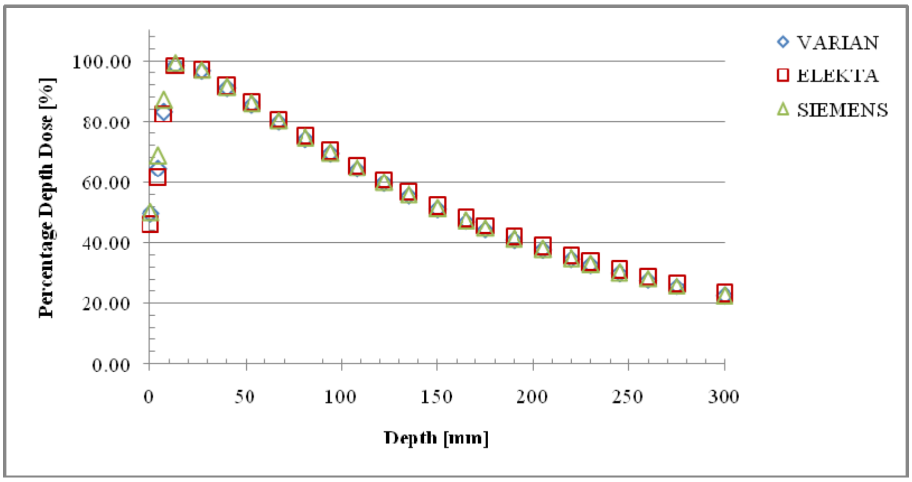

Fig. 4 – Brute PDDs as a function of depth in water, for a (10×10) cm^2 field size, in the case of the linear accelerator, VARIAN, ELEKTA, and SIEMENS.

In order to compare the depth dose curves, obtained with the same field size and energy, in different accelerators, a normalization procedure is introduced. Then, the normalization depth, N_D, is chosen to be 80 % of the maximum dose for each linac, at the same depth, according to the formula:

$$N_D = R/R_{80} , \quad (1)$$

where: R is the depth for a specified depth dose and R_{80} is the depth of the 80 % of maximum dose.

The results of depth normalization procedure are presented in the Table 2, for a 10×10 cm^2 field size.

Table 2

PDD as a function of depth and normalization depth for energy 6 MeV, field size (10×10) cm^2, SSD = 100 cm, for VARIAN, ELEKTA, and SIEMENS linacs

Normalisation depth N_D	Field size: 10×10 cm^2					
	VARIAN		ELEKTA		SIEMENS	
	Depth [mm]	PDD [%]	Depth [mm]	PDD [%]	Depth [mm]	PDD [%]
0.01	0.7	51.1	0.7	47.1	0.7	51.7
0.05	3.3	60.0	3.4	56.8	3.4	64.0
0.1	6.7	82.2	6.9	82.6	6.7	86.2
0.2	13.4	99.2	13.8	99.1	13.5	99.4
0.4	26.7	96.9	27.5	97.0	26.9	96.9
0.6	40.1	91.1	41.3	91.1	40.4	91.0
0.8	53.4	85.5	55.0	85.6	53.8	85.6
1.0	66.8	80.0	68.8	80.0	67.3	80.0
1.2	80.2	74.6	82.6	74.5	80.8	74.7
1.4	93.5	69.7	96.3	69.6	94.2	69.6
1.6	106.9	64.8	110.1	64.8	107.7	64.9
1.8	120.2	60.3	123.8	60.1	121.1	60.3
2.0	1336	56.0	137.6	55.9	134.6	56.1
2.2	147.0	51.9	151.4	51.8	148.1	52.0
2.4	160.3	48.3	165.1	48.1	161.5	48.3
2.6	173.7	44.8	178.9	44.6	175.0	44.9
2.8	187.0	41.6	192.6	41.4	188.4	41.6
3.0	200.4	38.6	206.4	38.4	201.9	38.6
3.2	213.8	35.9	220.2	35.7	215.4	35.9
3.4	227.1	33.3	233.9	33.1	228.8	33.4
3.6	240.5	30.8	247.7	30.7	242.3	30.9
3.8	253.8	28.6	261.4	28.6	255.7	28.7

Figure 5 shows the PDD plotted, as a function of the normalization depth, for all 3 linear accelerators (Table 2). However, after the normalization procedure, the differences between PDD values do not appear to be substantial.

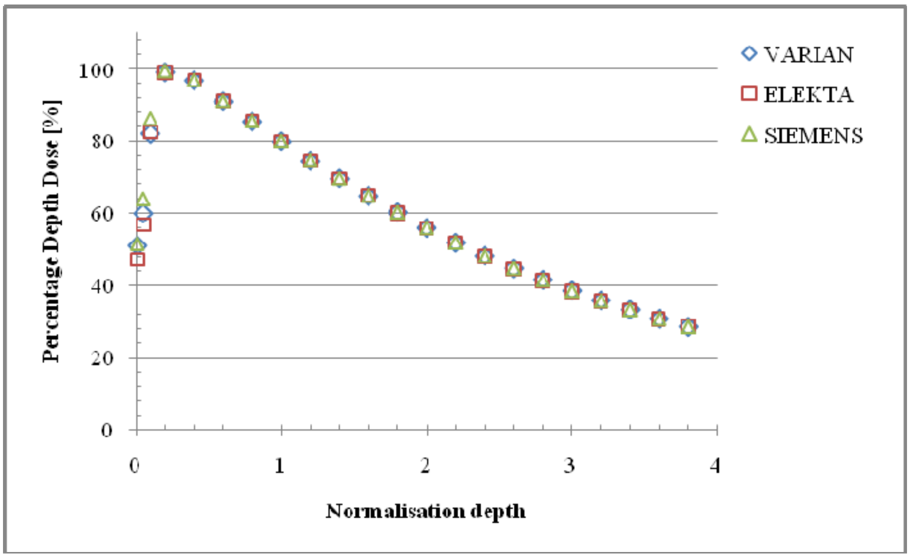

Fig. 5 – Normalization doses as a function of normalization depth, in water, field size (10×10) cm^2, for linear accelerators,VARIAN, ELEKTA, and SIEMENS.

The second step of the study was to analyze the differences between the normalized depth curves, for each field size. An appropriate way to do it, is to calculate the mean of the PDD values for all 3 accelerators (at the same energy and field size) on every normalisation depth value and to compute the relative deviation, for every case (Table 3 and Fig. 5).

Relative deviation, R_d, is given by the formula:

$$R_d = \frac{(PDD - \overline{PDD})}{\overline{PDD}} \times 100\% . \quad (2)$$

Table 3

Relative deviation as a function of normalization depth for energy 6 MeV, field size (10×10) cm^2, SSD = 100 cm, for VARIAN, ELEKTA, and SIEMENS linacs

Normalisation depth N_D	$\overline{PDD}$ [%]	Relative Deviation [%]		
		VARIAN	ELEKTA	SIEMENS
0.01	50.0	2.27	-5.74	3.47
0.05	60.3	-0.44	-5.75	6.19
0.1	83.7	-1.75	-1.27	3.03
0.2	99.2	-0.03	-0.13	0.17
0.4	96.9	-0.03	0.07	-0.03
0.6	91.1	0.04	0.04	-0.07
0.8	85.6	-0.08	0.04	0.04
1.0	80.0	0.00	0.00	0.00

Table 3 (continued)

1.2	74.6	0.00	-0.13	0.13
1.4	69.6	0.10	-0.05	-0.05
1.6	64.8	-0.05	-0.05	0.10
1.8	60.2	0.11	-0.22	0.11
2.0	56.0	0.00	-0.18	0.18
2.2	51.9	0.00	-0.19	0.19
2.4	48.2	0.14	-0.28	0.14
2.6	44.8	0.07	-0.37	0.30
2.8	41.5	0.16	-0.32	0.16
3.0	38.5	0.17	-0.35	0.17
3.2	35.8	0.19	-0.37	0.19
3.4	33.3	0.10	-0.50	0.40
3.6	30.8	0.00	-0.32	0.32
3.8	28.6	-0.12	-0.12	0.23

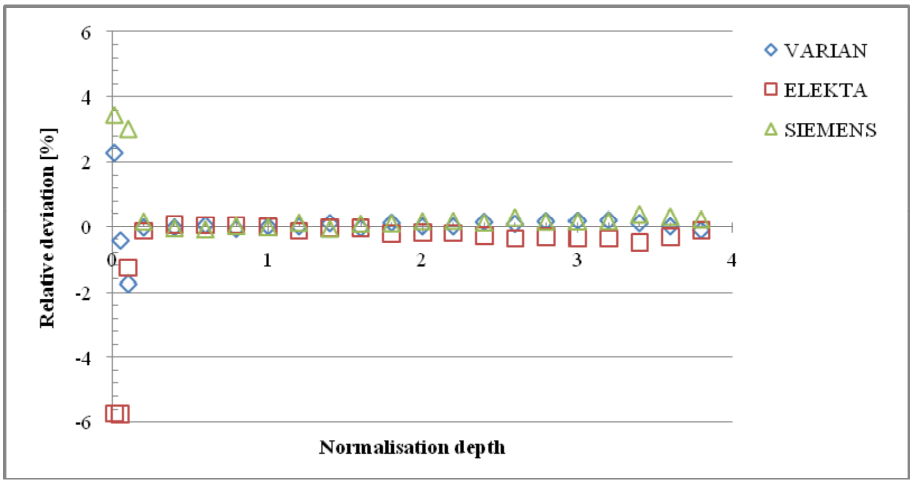

Fig. 6 – Relative deviation as a function of normalization depth for energy of 6 MeV, field size (10×10) cm^2, SSD = 100 cm, for VARIAN, ELEKTA and SIEMENS linacs.

The results shows that, the maximum relative deviation as a function of normalisation depth for a 10×10 cm^2 field size and a 6 MeV photon beam was around 0.5 % for all measured depth. Also, there was good agreement between PDD curves in build up region and maximum difference was observed for SIEMENS machine (around 6 %) (Fig. 6).

It is well known that, for commercial medical linear accelerators, the sources of radiation that determine dosimetric characteristics of clinical photon beams are:

– Direct radiation (focal radiation), that is, photon radiation generated at the target that reaches patient without any intermediate interactions;

– Indirect radiation (extrafocal radiation), which means photon radiation with a history of interaction/scattering in the head of the treatment unit filter with the flattening filter, collimators, or other structures in the treatment head;

– Contaminant electrons/positrons, that is, the secondary electrons and positrons released from interactions with either the treatment head or the air column.

Using the same procedure to analyze the data obtain from linacs, it can be seen that, for the entire range of field sizes, the VARIAN accelerator keeps relatively constant tendency towards the mean with small deviation (around 1 %), unlike the SIEMENS and ELEKTA systems that swap the trend from field size (5×5) cm^2 to (40×40) cm^2 with the deviation from the mean of 2 %.

Based on our PDD measurements, we can say that, for field sizes (10×10, 15×15, 20×20, 25×25, 30×30) cm^2, the maximum deviation was below 1 %, for all 3 accelerators. For field sizes (5×5, 40×40) cm^2 the deviation was no more than 2 %, for SIEMENS and ELEKTA linacs (Figs.7, 8).

A Monte Carlo study [4] for a 6 MeV beam for a Siemens linac, showed that an important contribution for the photon fluence has the target (83 %) and the window (7.6 %) which generate the direct radiation components and the primary collimator (3.7 %), the flattening filter (2.5 %) and the rest of the head (3.2 %) which generate the indirect radiation components. These values are slightly different for VARIAN and ELEKTA head architecture (for example, the ELEKTA target is thicker than VARIAN which cause more photon absorption. This could be a superiority of VARIAN linac to ELEKTA which need about 13 % more electrons for the same direct photon fluence [12]).

In the above situations, the overall results are very similar, concerning the photon beam of 6 MeV.

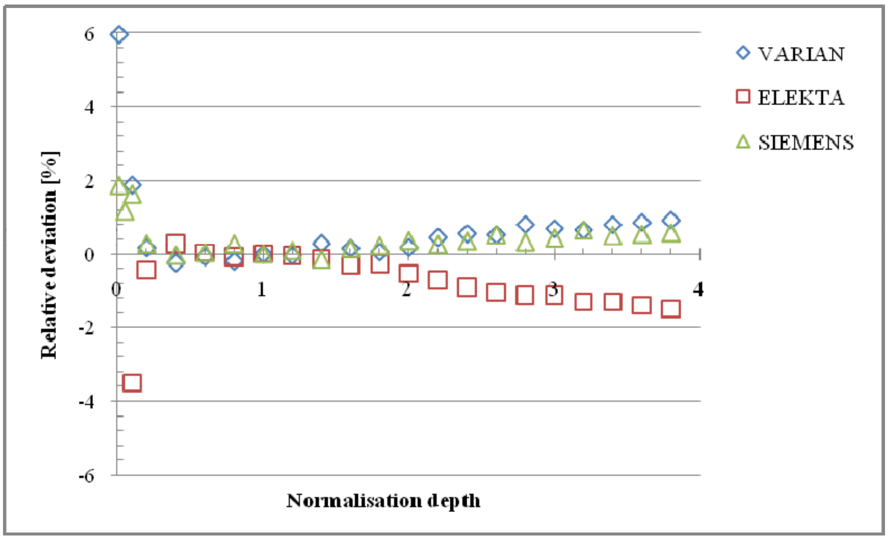

Fig. 7 – Relative deviation as a function of normalization depth for energy 6 MeV, field size 5×5 cm^2, SSD = 100 cm, for VARIAN, ELEKTA, and SIEMENS linacs.

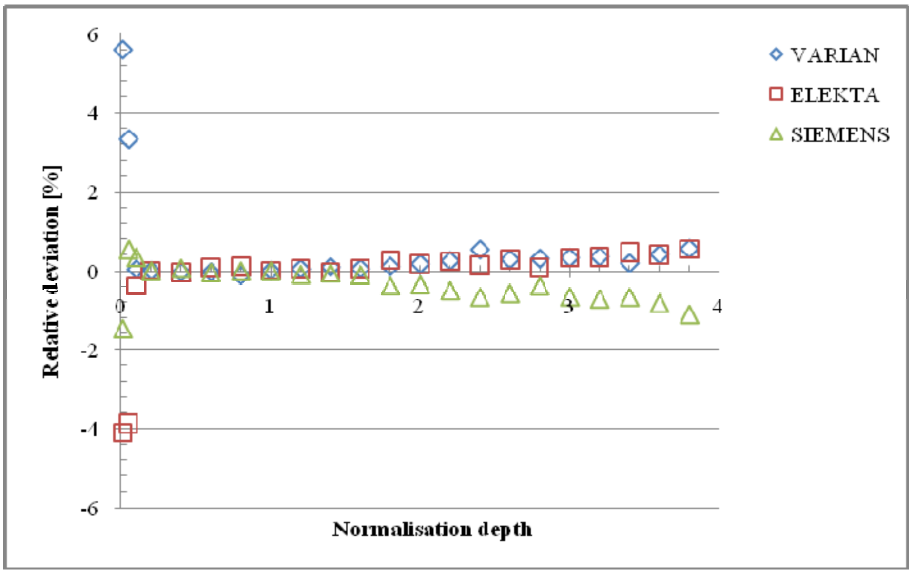

Fig. 8 – Relative deviation as a function of normalization depth for energy 6 MeV, field size (40×40) cm^2, SSD = 100 cm, for VARIAN, ELEKTA, and SIEMENS linacs.

The goal of treatment planning is to produce an appropriate isodose plan for the irradiated area. This plan is usually calculated for the use of treatment planning computers from the data obtained by the physicist. PDD normalization involves the selection of one of the values, the so called basic value, and then after normalization, all other values are presented as percentage of the basic value. The comparison of normalized PDDs is easier than the comparison of absolute values. In the clinical practice, various electron linear accelerators are used to produce photon beams. They differ in beam energy, construction, etc. Thus PDD diagrams obtained for medium irradiation differ, depending on the type of the medium, photon beam energy spectrum, and the field size. The comparison of various PDDs is possible due to the dose normalization procedure. In this paper, the dose is normalized to 80 % isodose line, this meaning that 80 % of the maximum dose rate (D_{80}) is at the same depth, for all the field sizes analyzed.

Another value, for example, the maximum range (R_{100}) could also be taken into consideration, although it is not so precise as R_{80} or R_{50}. The normalization makes PDD independent of beam energy or field size.

Thus, in clinical practice, there is a need for dose calculations, whenever irradiation of the patient follows the same treatment plan, but a different accelerator. Except for comparing of dose distributions for various combinations of energy, field size and accelerator, the normalization procedure is a convenient way as a part of accelerator check-up. Changing of energy or energetic spectrum in accelerator could cause each change in PDD shape. These parameters are very important for planning, because they constitute the base for computer calculations. They must be verified routinely. Dose normalization enables an easy and fast way of checking the accelerator standard parameters loaded into treatment planning systems. Summarizing, the normalization procedure, when dose is normalized to the depth selected by physicist, gives us advantages as follows:

– it provides the possibility of comparing the dose distributions for various energies, achieved in linear accelerators, both in scattering and scanning mode;

– it is a convenient way of comparison of dose distributions;

– it enables to check and control beam energy generated by a linear accelerator (during routine measurements).

5. CONCLUSION

Comparing PDD curves, obtained for the three linacs, one notices that the relative depth dose values were very close together and their differences were less than 2 % for all depths and all field sizes. There is a good agreement between PDD curves in build up region and differences observed around 6 % which can be explained due to contamination spectra.

The two main contributions to the dose, at the phantom surface and build up region, are electrons generated both by primary and secondary photon interactions within the water.

The photon energy spectra of the three linacs have very similar patterns and if normalized to maximum value, they almost overlapp each other. This small difference could have caused a negligible discrepancy between PDD curves of linacs.

Our results are consistent with those of Mesbahi *et al.* 2007 [12]. Despite of the fact that the architectures of the VARIAN, ELEKTA, and SIEMENS Linac treatment headsare quite different, the comparison between normalized percentage depth dose curves showed that the clinical results should be the same, provided that the treatment plan and biological response of the tumour are the same.

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