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ORIGINAL ARTICLE

A comparative study between flattening filter-free beams and flattening filter beams in radiotherapy treatment

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Abstract

Objective The aim of current study was to evaluation the physical parameters of FFF (Flattening Filter Free) energies in comparison with the standard energies used (flattening filter) in addition for clinical application like breast cancer treatment and palliative case for 6MV FFF and 10MV FFF respectively.
Methods The main characteristics of the photon beams have been analyzed using specific data generated by a Varian True Beam linear accelerator(linac), (Varian Medical Systems, Inc., Palo Alto, CA) linac having both FFF and FF beams of 6 and 10 MV energy, respectively, Eclipse treatment system for comparison, dosimetric system for relative and absolute dosimetry. We Compared all physical parameter for FFF versus FF and some clinical application for both energies 6 and 10 were included in current study for two sites cancer patients as example for planning to evaluation some parameter in each plan with different energy for example coverage of target volume, skin dose and organs at risk sparing (OAR).
Results The magnitude of contaminating electrons of FFF beam is relatively small in compared with standard for both energies and as a consequence the depth of dose maximum shows weak dependence on field size variation. Lateral dose profiles of FFF beam differ significantly from the FF beam for both energies 6 and 10 MV ($P < 0.05$). The central peak in the lateral profiles of FFF beam is pronounced only for medium to large field sizes. The higher the energy the more pronounced is the central peak. The shape of the lateral beam profile of an FFF beam changes slightly with depth due to a significantly reduced off-axis softening effect and hence the depth--dose characteristic remains almost constant across the field even for large field sizes. In clinical application we shown a significant difference between dose distribution for 6MV (FFF) in target coverage and doses for organs at risk versus 6 MV (FF) due to difference in depth of maximum dose and applicability for good distribution with FFF beam in case for breast cancer patient and good coverage for palliative treatment for bone Metastasis Using 10 MV(FFF) in compared with standard energy 10MV (FF).
Conclusion There are significant difference between dose distribution for 6MV (FFF) and 10MV(FFF) in target coverage and doses for organs at risk versus flattening energy due to difference in depth of maximum dose and applicability for good distribution with FFF beam, sparing for OARs and reduction in the time of treatment due to highest dose rate for FFF more than standard energies.
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Keywords: flattening filter-free beam; flattening filter beam; breast cancer; treatment planning; beam characteristics; multileaf collimator

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Traditionally, the flattening filter (FF) in the X-ray beam path of a linear accelerator produces an almost uniform fluence over a collimated field. This is particularly advantageous for 3D conformal radiation therapy (CRT) for practical reasons.

The removal of the FF leads to a rapidly decreasing fluence distribution, and thus to inhomogeneous dose distributions. The advantage of this is its positive influence on the peripheral dose through reduced head scatter and MLC leakage [1], as well as a considerable increase in the dose rate, which has a beneficial effect on modern therapy methods.

In addition to improved shielding in the treatment head, Ponisch *et al* [2] suggest the use of secondary jaws to track the MLC and removal of the FF as a source of scattered radiation with fluence-modulated radiation therapy (RT). The disadvantage of a non-uniform, conical fluence distribution can be considered with intensity-modulated radiation therapy (IMRT) in an optimization algorithm. Recent studies show the feasibility of using flattening filter-free (FFF) beams for IMRT and stereotactic body radiation therapy (SBRT) [3–5]. In addition, it has been concluded that the decreased variation in scatter factors and beam quality along the field simplifies the dose calculation [6]. It is often necessary to resort to field-in-field (FIF) techniques, also referred to as forward IMRT techniques, to achieve better conformity for the PTV in 3D CRT planning. Additional fields in one angle of incidence (multistatic field) can be used to adapt dose distribution optimally to the anatomy of the patient without the need for a wedge. Several studies for various RT locations show that a beneficial dose distribution can be achieved with this method with regard to homogeneity and conformity [7–9]. It is also possible to use FFF beams in 3D CRT through this FIF technique.

Materials and methods

Materials

Linear accelerator—TrueBeam system

The linear accelerator used in this study is a TrueBeam linac developed by Varian Medical Systems. It is designed to deliver FF as well as FFF photon beams. It represents a new platform of Varian linacs, where many key elements, including the waveguide system, carousel assembly, beam generation, and monitoring control system, differ from the preceding CLINAC series. Further, it contains a multiport X-ray filter management system (carousel) that accommodates field-flattening filters and open ports. The dosimetry systems of these linacs (monitor chamber) are capable of accurately processing a wide range of ionizations per pulse. The maximum dose rates of the TrueBeam system are 1400 and 2400 MU/min for 6 MV (labeled as 6XFFF) and 10 MV (labeled as 10XFFF) X-rays, respectively. The accelerator is equipped with asymmetric collimation jaws and an MLC consisting of 120 leaves on each side, allowing a maximum field size of 40×40 cm^2.

Dosimetry system

A PTW MP3 water phantom (PTW Freiburg GmbH) with inner tank dimensions of $694.0 \times 596.0 \times 502.5$ mm^3 is used together with a cylindrical semiflex ion-chamber (PTW, type 21010) with an inner cavity volume of 0.125 cm^3. Further, to compensate for beam output variations, a cylindrical ion-chamber (PTW, type 31010) was used as a reference in the present and all the following relative dose measurements.

PTW UNIDOS electrometer

For all measurements with the water tank scanning system, a PTW UNIDOS Electrometer is used, and the data collection is performed using the PTW MEPHYSTO software.

Vascular slices showed sharply cut edges with blood vessels. Outside the cut edge, the vessel wall and surrounding tissue were thick, and had long endovascular thrombosed segments.

Portal Dosimetry

During this work, Electronic Portal Imaging Devices (EPID) are used for acquiring megavoltage images during patient treatment. The larger area of the Digital Megavolt Imagers (DMI) is a square (43×43 cm^2 for single images).

Methods

Linear accelerator—TrueBeam system

For each photon energy, percentage depth dose curves (PDDs) are acquired for 13 square field sizes: 2, 3, 4, 6, 8, 10, 12, 15, 20, 25, 30, 35, and 40 cm. Field size is defined by jaws, not MLC, due to the standard data collection measured for jaws only, and after all data commissioning, then scatter from MLCs measured and add for data transfer for treatment planning system for creation different shapes for field sizes.

The water level is checked periodically using the front pointer – always before the first scan – for X6 and 6FFF additionally at mid-field size. Evaporation makes it necessary to fill the water tank approximately every 30 minutes, depending on room temperature and humidity. The front pointer method can detect changes of SSD in the order of 0.2 mm.

On one hand, the depths for maximum dose (dmax) serve as reference depths for the linac output calibration. Following the Varian recommendation, we calibrate our TrueBeams to deliver 1.0 Gy per 100 MU at an SSD of 100 cm at a depth of dmax for the 10×10 cm^2 reference field.

Results

D_{max} results

The following dmax depths for the 10 × 10 cm^2 field were determined:

16 mm (X6)
14 mm (6FFF)
26 mm (X10)
24 mm (10FFF)

Fig. 1 and 2 show the PDDs for all field sizes (from 2.0 cm^2 to 40 cm^2) for 6 MV (FF and FFF). Figs. 3 and 4 show the beam profiles for all field sizes (from 2.0 cm^2 to 40 cm^2) for 6 MV (FF and FFF).

Figs. 3 and 4 show the beam profiles for field size 40 × 40 cm^2 for 6 MV (FF and FFF), from which the following observations can be made:

Sm: in the plateau region, all profiles were smoothed once.

For the two FFF energies only, a single data point was added to the 40 × 40 cm^2 profile at 300 mm depth, at the location –294.5 mm. The data point value was guessed visually.

Sym: all profiles were mirrored (symmetries).

The saturation correction of transverse profiles is shown in analogy to the PDD curves. However, we find that it does not make sense to correct the profiles for saturation. While for a PDD at the lowest energy (6FFF), the dose per pulse ratio between d_{max} and 300 mm depth can be up to 6.7 (the PDD value of 6FFF, field size 1 × 1 cm, at 300 mm depth, is 14.9 %), the dose per pulse at the shoulder point of a FFF cross-plane profile is at least 40% (10FFF, 40 × 40 cm^2, 22 mm depth). As the transverse profiles are always normalized to the central axis (CAX) during TPS formatting, a saturation correction of the profiles would have no effect between the CAX and the shoulder point, from which it can be concluded that the effect is smaller than 0.1 %.

The magnitude of contaminating electrons from the FFF beam is relatively small and, therefore, the depth of the dose maximum shows a weak dependence on field-size variation. Lateral dose profiles of the FFF beam differ significantly from those of the FF beam. The central peak in the lateral profiles of the FFF beam is pronounced only for medium to large field sizes. The higher the energy, the more pronounced the central peak is. The shape of the lateral beam profile of a FFF beam changes slightly with depth due to a significantly reduced off-axis softening effect, and hence the depth-dose characteristic remains almost constant across the field, even for large field sizes.

Optimization of dose distribution

The larger area of the Digital Megavolt Imager (DMI) is square (43 × 43 cm^2 for single images). This offers the possibility to image larger field lengths at the same imaging distance.

Resolution is also slightly improved. The image size is 1280 × 1280 pixels for single images.

Case study – 6 MV (FFF)

For the second FFF energy, 6FFF, portal dosimetry results also improved dramatically. Outside the Complete Irradiation Area Outline (CIAO), profiles measured with a Varian aS1000 imager were often very low compared to calculations using EPIQA (EPIQA is non-transit commercial software that can convert a dosimetric image acquired by an EPID into a dose map, and compares the dose map with a reference dose distribution). This problem does not currently exist.

Fig. 5 shows an example of IMRT plan 5 fields of the breast, 6FFF, 1400 MU/min, again measured at isocenter distance. With the 2%/2 mm criterion, the gamma agreement index for both arcs is approximately 100% over the whole detector area.

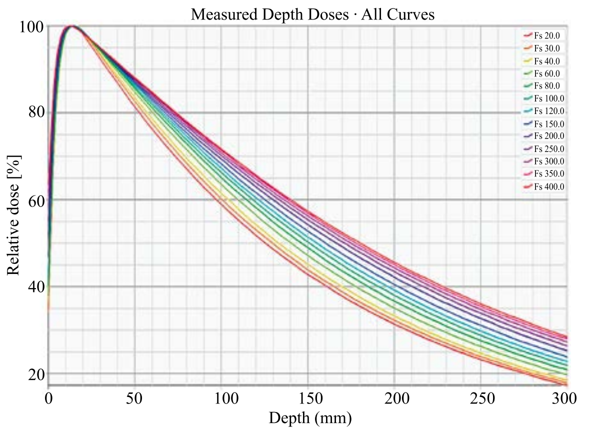

Fig. 1 Percentage depths dose for all field sizes (from 2.0 cm^2 to 40 cm^2) for 6 MV (FF)

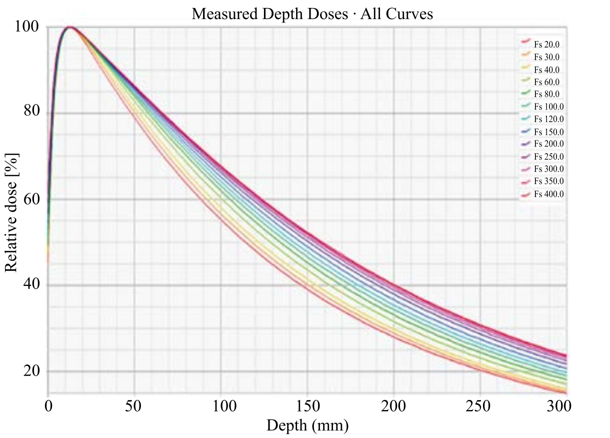

Fig. 2 Percentage depths dose for all field size (from 2.0 cm^2 to 40 cm^2) for 6 MV (FFF)

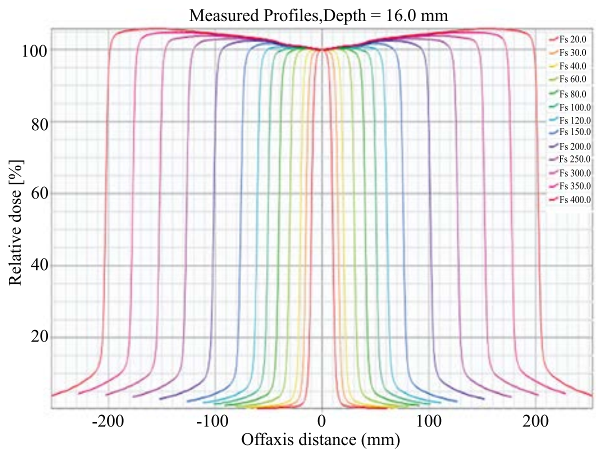

Fig. 3 Beam profiles for all field sizes (from 2.0 cm² to 40 cm²) for 6 MV (FF)

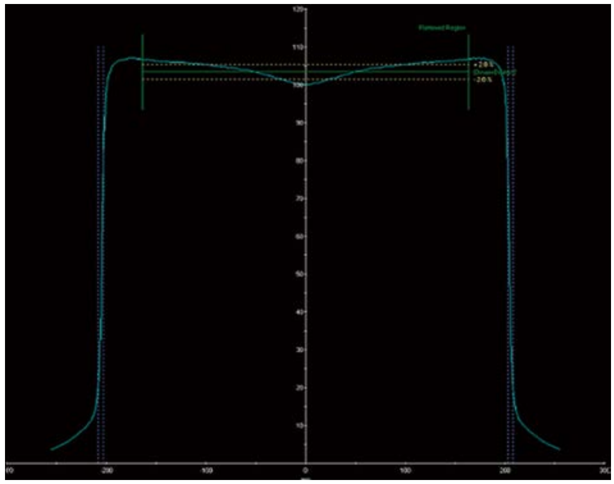

Fig. 5 Beam profile for field size 40 × 40 cm² for 6 MV (FF)

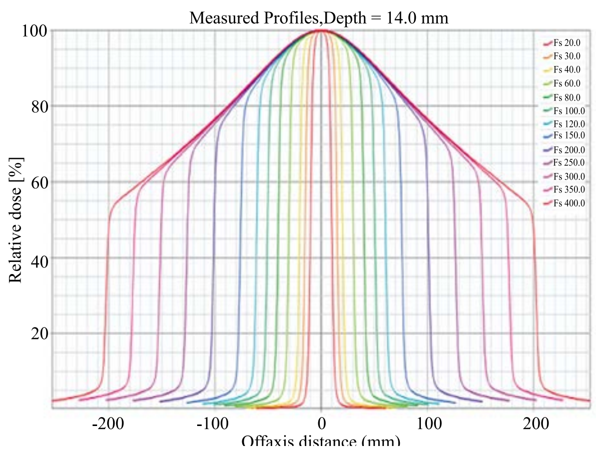

Fig. 4 Beam profiles for all field sizes (from 2.0 cm² to 40 cm²) for 6 MV (FFF)

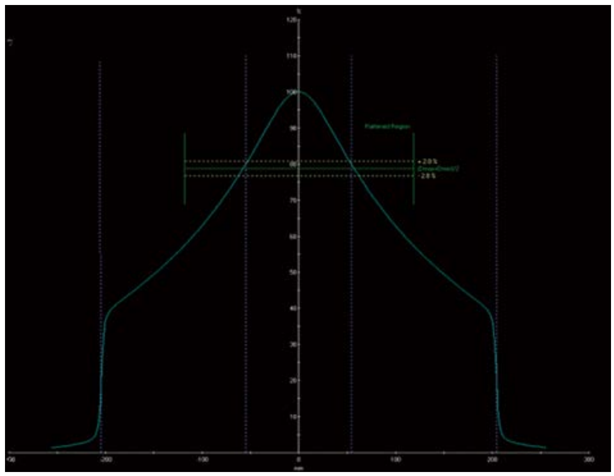

Fig. 6 Beam profile for field size 40 × 40 cm² for 6 MV (FFF)

As shown in Fig. 5, there is a significant difference between dose distribution for 6 MV (FFF) in target coverage and doses for organs at risk versus 6 MV (FF), due to the difference in depth of the maximum dose and applicability for good distribution with the FFF beam.

The photo-neutron fluence per monitor unit (MU) produced by the high-energy FFF beam is relatively less in comparison to that produced by the FF beam (Fig. 8). Hence, operating the accelerator in the FFF mode benefits both the patient and the radiation therapy technologist. However, the benefit of decreased neutron dose for FFF beams at high X-ray energies (15, 18 MV) needs to be critically examined, giving due consideration to their clinical use over low X-ray energies (6, 10 MV). Due to reduced average energy, treatment head leakage, and fractional neutron dose, the concrete thickness required for the FFF linac vault is also relatively less. Thus, the existing linac vault can safely be used for operating the FFF linac at reduced occupational exposure, and while constructing a new shielded vault there will be a saving of space and cost.

In addition, the phosphor screen of the EPID shows increased sensitivity to low-energy photons present in the spectra of the FFF beam. It was also reported that the EPID-measured profile changes minimally with increasing phantom thickness due to small energy variation across the profile. Portal dosimetry using existing EPID of standard linac is therefore a possible option for patient-specific quality assurance in the FFF beam.

Case study – 10 MV (FFF)

The following example shows palliative treatment of single are, 10FFF energy, 2400 MU/min and 2357.2 MU

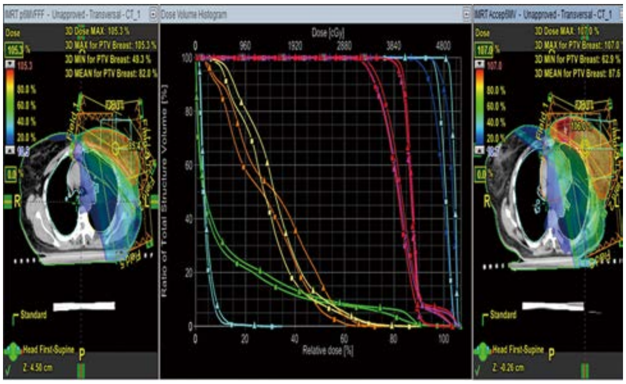

Fig. 7 Comparison for two-dose distribution for breast cancer, plotted at 6 MV (FFF) and 6 MV (FF)

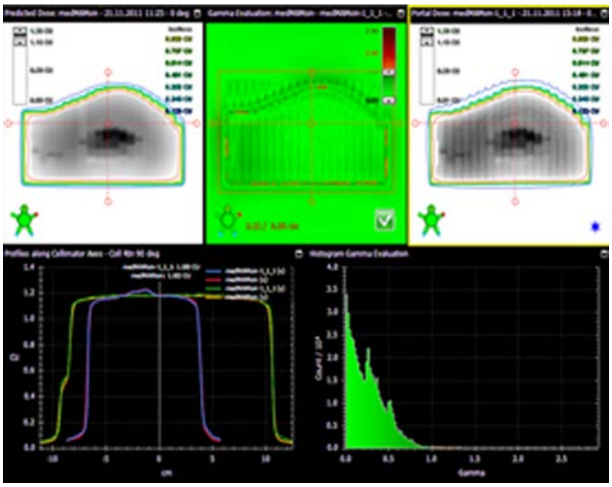

Fig. 8 EPID evaluation of two tangents for a breast cancer patient using FFF, as an example of a portal dosimetry tool for a TrueBeam linear accelerator

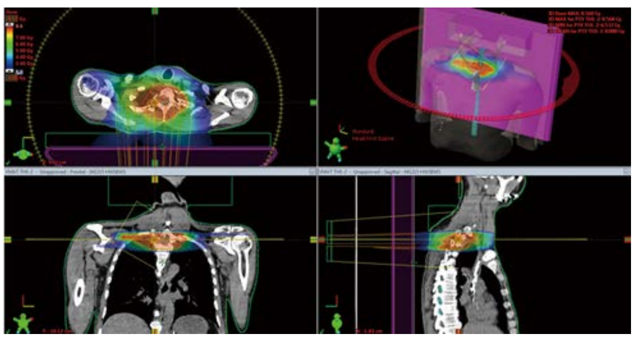

Fig. 9 Example of dorsal lesion for 10 MV FFF beam and dose distribution using the VMAT technique (single arc)

total, planned with Eclipse 13.6 (AAA 13.6.23). This plan was verified with portal dosimetry for a single fraction in palliative cases (Fig. 7).

A verification plan was calculated using a portal dose image prediction (PDIP) algorithm. The arc was measured with the DMI imager at isocenter distance and analyzed using the 3%/3 mm DTA gamma criterion:

Volumetric modulated arc therapy (VMAT), or Varian RapidArc® Radiotherapy Technology, is an advanced form of IMRT that delivers a precisely sculpted 3D dose distribution with 360° as the maximum for angle rotation of the gantry in a single or multi-arc treatment. RapidArc uses a dynamic MLC, variable dose rate, and variable gantry speed to generate IMRT-quality dose distributions.

The current dosimetry protocols that are followed for output measurement of photon beams from medical linear accelerators require a beam quality correction factor. This beam quality correction factor is related to the quality index [%DD(10) or TPR1020] of the photon beam. As the reference conditions for measuring the quality index of the photon beam is given with reference to the FF beam, it cannot be directly applied for the FFF beam. Therefore, there is a need to revise the existing dosimetry protocols for the FFF beam. The conventional definition of beam penumbra is not applicable to the FFF beam, and requires a modification to the definition. The primary electrons have been reported to penetrate through the high Z thin targets used for generating bremsstrahlung photons, posing a potential risk for producing high surface doses if not removed.

In the case of a standard photon beam (FF), the electrons penetrating through the thin bremsstrahlung targets are efficiently removed by the FF. In a FFF linac, an additional thin metal plate in front of the monitor chamber is used to remove the primary electrons penetrating through the bremsstrahlung target. The material and the thickness of this plate need to be optimized, maintaining the advantage of the FFF beam, and giving due consideration to the incidence of bremsstrahlung target breaks.

Discussion

Advanced beam therapy techniques, such as RT, where inhomogeneous dose distributions are applied, and IMRT, where varying fluence patterns across the beam are delivered, have stimulated interest in operating a standard linac in the FFF mode. A standard linac can be used for generating photon beams with either an FF beam or a FFF beam. Several Monte Carlo and experimental studies dealing with characteristics, dosimetric aspects, and radiation protection issues of FFF photon beams, generated by mechanically removing the flattening filter of existing standard linacs of different makes and models, have been recently reported. Studies related to treatment planning and dose delivery of various clinical cases using FFF beams demonstrate their clinical suitability and superiority over FF photon beams. A review of the properties of FFF photon beams summarizing the findings of different investigators has also been recently published.

A recent study by Hrbacek et al [15] reports the measured dosimetric characteristics of unflattened photon beams generated using a new model of a standard linac (Varian TrueBeam STx), capable of generating both flattened and unflattened clinical photon beams. It is well known that the FF in a standard linear accelerator acts as an attenuator, a beam hardener, and a scatterer. Obviously, the removal of the FF results in an increase in dose rate, softening of the X-ray spectra, and a reduction in head scattered radiation and the non-uniform beam profile. The reported dose rate of FFF beams is about 2–4 times higher than that of the FF beams, that is, FFF linear accelerators can typically be operated at a dose rate higher than 10 Gy/min under the normal operating conditions applied for FF linear accelerators. The increased dose rate decreases the dose delivery time, especially for hypo-fractionated stereotactic radiotherapy (SRT), and is thought to be useful in managing the intrafractional target motion.

The softening of the X-ray spectra affects the depth as well as the lateral dose distribution at all depths, and results in increased surface dose and slight shifting of the depth of maximum dose toward the surface. The lateral transport is reduced, which may result in greater control over gradients within the field and at target boundaries. The head scatter variation for an unflattened beam is typically about 1.5 % as against about 8 % of the flattened beam for the field sizes in the range from 3×3 cm^2 to 40×40 cm^2.

As a result, a simple model for dose calculation of irregular treatment fields is sufficient for the FFF beam. Moreover, due to the absence of the collimator exchange effect, it is not necessary to account for whether the upper or lower secondary collimator is defining the long side of the rectangular beam. The decreased head scatter, and hence the reduced head leakage, also results in decreased far field peripheral dose (PD) to the patient.

The near field PD is also due less to the combined effects of softer photon beam spectra, increased dose per pulse, and reduced collimator transmission. While treating the patients by radiotherapy (IMRT) with a 6 MV FFF beam, the integral dose to nearby healthy tissue and the whole-body integral dose, respectively, were found to be significantly higher than the FF beam, and the use of higher FFF beam energy is suggested as the remedy for the problem (e.g., using 10 MV instead of 6 MV) [1]. This is because the 10 MV unflattened depth dose characteristics are similar to those for a 6 MV flattened beam. The use of a FFF beam over a FF beam is a natural choice for IMRT treatments. However, the leaf travel time for creating a large number of optimized segments for static IMRT and the leaf speed for the dynamic and rotational IMRT are the limiting factors in the dose delivery efficiency of IMRT.

Hence, for effective and efficient use of the FFF beam, the technology of current MLCs needs to be modified. Further, the intensity of the FFF beam abruptly decreases with the off-axis distance for large open fields ($\geq 10 \times 10$ cm^2), which necessitates the off-axis distance-dependent modulation for delivering uniform dose to the tumor. While executing the off-axis distance-dependent modulation by dynamic MLC, larger monitor units are required, which increase the gross head leakage and lessen the advantage of using the FFF beam. This effect is significant in dynamic IMRT of off-axis targets and large volume targets; while dealing with such clinical cases, a modified FFF beam is required [16].

Conclusions

Although there are a number of advantages of using a FFF beam, especially for advanced radiotherapy techniques, there are a few limitations (e.g., using a relatively higher energy photon beam for SRT, limited speed of current MLCs, and off-axis distance-dependent modulation in IMRT) as well as challenges (e.g., criteria for beam quality evaluation and penumbra, establishment of dosimetry methods, and consequences of photon target burn-up) that need to be addressed for establishing the FFF beam as a viable alternative to the FF beam.

Conflict of interest

The authors indicated no potential conflicts of interest.

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