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IPEM topical report 1: guidance on implementing flattening filter free (FFF) radiotherapy

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Abstract

Flattening filter free (FFF) beams are now commonly available with new standard linear accelerators. These beams have recognised clinical advantages in certain circumstances, most notably the reduced beam-on times for high dose per fraction stereotactic treatments. Therefore FFF techniques are quickly being introduced into clinical use. The purpose of this report is to provide practical implementation advice and references for centres implementing FFF beams clinically. In particular UK-specific guidance is given for reference dosimetry and radiation protection.

Keywords: FFF, guidance, UK

(Some figures may appear in colour only in the online journal)

1. Introduction

In radiotherapy, beams of high energy x-rays are generated for therapeutic use by bombarding a high-Z metal target with high-energy electrons. The bremsstrahlung x-rays emitted in this process have a strongly forward-peaked intensity profile (which becomes more peaked as the incident electron energy increases) but a relatively constant angular energy spectrum. To counteract this triangular shaped beam profile and provide a nearer uniform fluence distribution it is customary to insert a conical shaped flattening filter (FF) into the beam. Historically, radiotherapy has been based on the delivery of flat (or wedged) beams to treat 'box-like' volumes to a uniform dose and with limited computing power these calculations were aided by near-uniform beam profiles. From here on this approach will be referred to as conventional flattening filter (cFF) delivery. Modern radiotherapy is no longer reliant on these methods, instead utilising fluence modifying techniques such as intensity-modulated radiotherapy (IMRT).

At this point in time IMRT techniques are set to overtake more traditional 3D conformal radiotherapy (3DCRT) across much of the world. In 2012 the percentage of radically irradiated patients treated with IMRT in the UK was reported as 15.3% (Mayles *et al* 2012). By 2014 this figure had increased to 35% and is predicted to increase to over 50% (Taylor *et al* 2015). The use of stereotactic ablative radiotherapy (SABR) techniques to treat early stage non-small cell lung cancer (NSCLC) has also gained in popularity and the introduction of a National Commissioning through Evaluation programme is set to double access to these techniques, opening the door for other sites such as liver, spine and oligometastases. The proportion of indications requiring either small field or intensity modulated treatments is therefore set to dominate the radiotherapy scene and for these techniques the flattening filter is not actually necessary, but more importantly, is no longer appropriate.

In flattening the beam profile a large proportion of the beam intensity is removed, the majority of which is then converted into scattered radiation. The flattening filter therefore acts as a secondary source of radiation (for both photons and electrons). It is estimated that extra-focal radiation contributes 11–16% of the photon fluence at the isocentre and that roughly 70% of this comes from the flattening filter (Petti *et al* 1983, Liu *et al* 1997); exact figures are energy and vendor specific. In producing a flat beam the filter causes a series of negative effects, such as:

- decreased primary beam intensity, leading to reduced dose rate;
- differential absorption across the field (changes in beam spectrum) causing problems for dose calculation and beam modelling;
- the need for the introduction of 'horns' in the particle fluence to compensate for this angular variation of the spectrum;
- the creation of a significant source of extra-focal scattered radiation;
- electron contamination in the primary beam;
- increased leakage radiation from the treatment head, increasing head shielding requirements;
- amplification of beam steering errors, necessitating the use of active beam monitoring and servo control.

The use of unflattened beams has come to be known by the acronym FFF, for flattening-filter-free. FFF beams were first studied by O'Brien *et al* (1991) who investigated their use on a Therac-6 linear accelerator (linac) with the intention of increasing the dose rate to reduce treatment times for intracranial stereotactic treatments. At this time there was already an accelerator operating without a flattening filter: the Racetrack Microtron MM50 (Brahme

et al 1980, Karlsson *et al* 1993) which utilised a scanning electron beam incident on the target to build up dose in the patient in a similar manner to spot-scanning proton beams. In the modern era the TomoTherapy and CyberKnife systems are examples of accelerators designed specifically for IMRT and stereotactic radiotherapy respectively. In these units flattening filters were deemed unnecessary, and manufactures of standard accelerators have followed suit with Elekta, Siemens and Varian now offering FFF beam energies.

FFF research has expanded with the increased use of IMRT and stereotactic radiosurgery (SRS)/SABR techniques, and also with the number of clinical accelerators now in use offering these beams. However, despite this increase in use there is still a lack of clear guidance on the use of FFF beams, particularly for conventional C-arm accelerators where definitions of field size, penumbra and quality assurance (QA) requirements are sparse.

At the present time manufacturers still require FFF linacs to also have cFF beams activated in order to define standard beam properties and QA requirements. However, many clinics already operate their FFF linacs exclusively in FFF mode clinically and reliance on the presence of cFF beams adds an additional QA burden. These dependencies need to be addressed in future releases.

Numerous research articles have been published on the dosimetric properties of FFF beams from Varian and Elekta accelerators (Fu *et al* 2004, Titt *et al* 2006, Pönisch *et al* 2006, Vassiliev *et al* 2006a, 2006b, Cashmore 2008) demonstrating that filter free operation results in reduced head leakage, higher dose rates and simplified beam modelling whilst maintaining plan quality.

2. Technical comparison

Filter free beams are now commercially available on a range of conventional C-arm linear accelerators from Varian, Elekta and Siemens. However, since Siemens left the linac market in 2012 this report will only discuss the Varian and Elekta implementations of this technique. Removal of the flattening filter from the beam-line in both systems is a straightforward process because of its location on a rotating carousel within the treatment head. In practice though, the filter cannot simply be removed but needs to be replaced by a thin metal 'enhancer' plate in the same position as the FF. This plate generates electrons which provide build-up dose to the ionisation chamber to give sufficient signal (and position-dependant information) to the servo plates. They, in turn, can then operate correctly to control the beam quality and steering (Vassiliev *et al* 2006a, Cashmore 2008). Material is also necessary to remove those contamination electrons generated in the primary collimator and target from the beam, which do not provide useful position information but do increase surface dose to the patient. Some material is also necessary so that the electron beam is never directly incident on the patient in the unlikely event of a target failure. Several studies have been undertaken to determine the most appropriate material, and thickness of material to fulfil these functions, and these have been reviewed by Georg *et al* (2011). Whilst these plates allow clinical operation of the accelerator with no FF without adding significantly to scatter, the absence of the FF causes significant changes to the energy spectrum. The beam hardening effect of the FF removes photons below about 1 MeV almost completely on the central axis (CAX), with decreasing effect towards the field edge. Without this material most of the low-energy photons pass through, resulting in a lower average beam energy and altering the penetration of the beam.

This softening of the beam then raises the question of whether the electron energy incident on the target should be raised in an attempt to try and match the original cFF beam, or whether a softer beam spectrum should be accepted. An FFF beam can never be completely matched

Table 1. Beam characteristics for 6 and 10 MV FFF and cFF beams from Varian and Elekta. See Xiao *et al* (2015) for Siemens data.

		Nominal energy (MV)	Filtration	Effective energy (MV)[a]	d_{max} (cm)	D_{10} (%)	$TPR_{20/10}$	Max dose rate (MU min^{-1})	Dose (mGy) per pulse[b]
Varian	FFF	6	0.8 mm	4	1.3	64.2	0.630	1400	0.8
		10	Brass plate	8	2.2	71.7	0.705	2400	1.3
	cFF	6	6/10 MV	6	1.4	66.4	0.666	600	0.3
		10	flattening filter	10	2.3	73.6	0.738	600	0.3
Elekta	FFF	6	2.0 mm stainless	6	1.7	67.5	0.684	1400	0.6
		10	steel plate	10	2.4	73.0	0.734	2200	0.9
	cFF	6	6/10 MV	6	1.5	67.5	0.678	600	0.2
		10	flattening filter	10	2.1	73.0	0.721	600	0.4

[a] Clinical effective energy, based on $TPR_{20/10}$ and percentage depth dose falloff.
[b] Measured at d_{max} on beam central axis for standard reference conditions.
Note: D_{max} refers to depth of maximum dose. MU are monitor units.

to a conventional beam as there are irreconcilable differences in the beam spectrum and in the relative contributions of low-energy photons and electrons to make this possible. The conical shape of the FF also leads to differential beam hardening, most prominent at the CAX, and gradually reducing with distance towards the field edge. Therefore any 'matching' can only exist in one position in the beam; the central axis. To do this a fixed parameter and field size must therefore be chosen for matching the beam to. The most common specifiers of beam energy used are D_{10}, the relative dose at 10 cm depth or the quality index (tissue phantom ratio 20/10 ($TPR_{20/10}$)), usually for a $10 \times 10\,cm^2$ radiation field size.

The key difference between the manufacturers is how they approach this issue of beam energy. In simple terms the Varian implementation utilises the same electron beam to create both cFF and FFF beams, resulting in a less penetrating beam for FFF compared to a cFF beam of the same nominal energy. In the Elekta system the FFF beams are independent of the cFF beams and energy-matching is undertaken to maintain central axis depth dose under nominal conditions (90 cm source to surface distance (SSD), 10 cm deep, $10 \times 10\,cm^2$). A 6 MV FFF beam on a Varian linac therefore has depth dose characteristics similar to a 4 MV cFF beam (Vassiliev *et al* 2006b), whereas an Elekta 6 MV FFF beam PDD remains similar to an Elekta 6 MV cFF PDD. These differences in beam energy between accelerator manufacturers must be kept in mind when comparing not just beam properties, but also room shielding and treatment plans. A summary of the main features of the Varian and Elekta FFF implementations, including D_{10} and quality index ($TPR_{20/10}$) is given in table 1, with cFF beam data for comparison.

Despite the differences in beam energy, the overall effects of filter removal in both implementations are broadly similar i.e. increased dose rate, reduced scatter, forward-peaked beam profile, reduced energy variation. These characteristics are examined in more detail in the next section.

2.1. Characteristics of FFF beams

This section briefly outlines the main differences between cFF beams and FFF beams in terms of dose rates, percentage depth doses (PDDs), profiles and scatter factors.

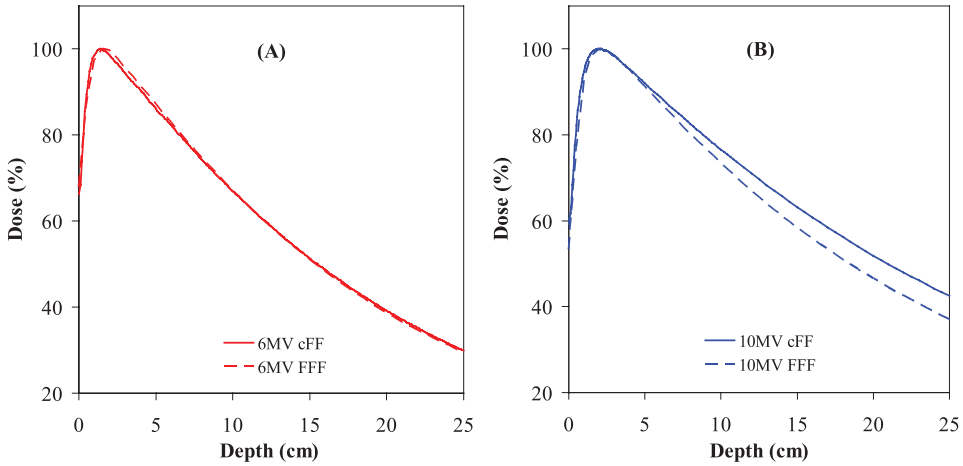

Figure 1. 10 × 10 cm cFF and FFF PDD for (A) 6 MV Elekta and (B) 10 MV Varian.

2.1.1. Dose rate. FFF beams can be run at higher MU min^{-1} rates than cFF beams. Elekta 6 MV and 10 MV FFF beams have maximum dose rates of 1400 MU min^{-1} and 2200 MU min^{-1} respectively. The Varian 6 MV FFF beam runs at a maximum of 1400 MU min^{-1} and the 10 MV FFF beam at a maximum of 2400 MU min^{-1} (Fu *et al* 2004, Pönisch *et al* 2006, Titt *et al* 2006, Vassiliev *et al* 2006a, 2006b, Cashmore 2008). For comparison, TomoTherapy 6 MV FFF beams run at a maximum of 850 MU min^{-1} and CyberKnife 6 MV FFF beams have a maximum dose rate of 1000 MU min^{-1}.

2.1.2. PDDs. Figure 1 shows the PDDs for the most commonly clinically used FFF beams for each manufacturer along with their cFF counterparts; as explained in section 2. The measured $TPR_{20/10}$ for an Elekta linac should be close to that of a cFF beam of the same nominal energy, whereas for Varian machines the values of $TPR_{20/10}$ measured for 6 MV FFF and 10 MV FFF will be closer to those of 4 MV and 8 MV cFF beams respectively.

2.1.3. Profiles. The difference in profile shape is the single greatest change associated with removal of the flattening filter, common to both Elekta and Varian linacs. Figure 2 shows a side by side comparison of a typical profile obtained during commissioning for both a cFF and FFF beam. This change in profile shape leads to some difficulties in the assessment of beam profiles using the standard definitions employed for analysis of beam profiles.

(1) Beam symmetry and flatness—the symmetry of the beam is still an important parameter for beam steering and energy. However, the calculation of 'flatness' for an FFF beam is no longer of use outside of a consistency check of profile shape, since with the removal of the flattening filter it provides no information on beam centring.
(2) Field size—for cFF beam profiles the field size is defined as the distance between the 50% dose points (full width half maximum) for a given profile. In figure 2, it can be seen this is no longer valid as a field size specifier for FFF beams given the changes to profile shape. See section 4.2 for further discussion of this issue.
(3) Penumbra—similarly to field size, the 80–20% definition for field penumbra is no longer valid for FFF beams.

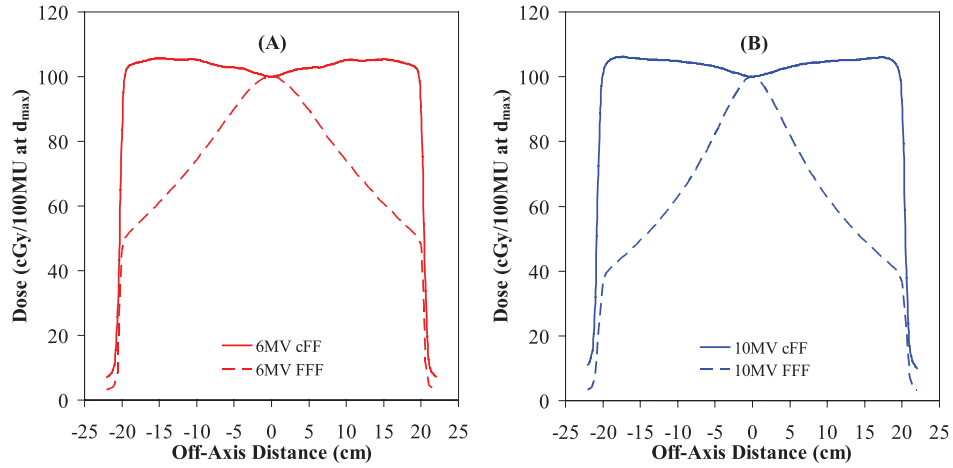

Figure 2. 40 × 40 cm cFF and FFF profiles for (A) 6 MV Elekta and (B) 10 MV Varian at d_{max}.

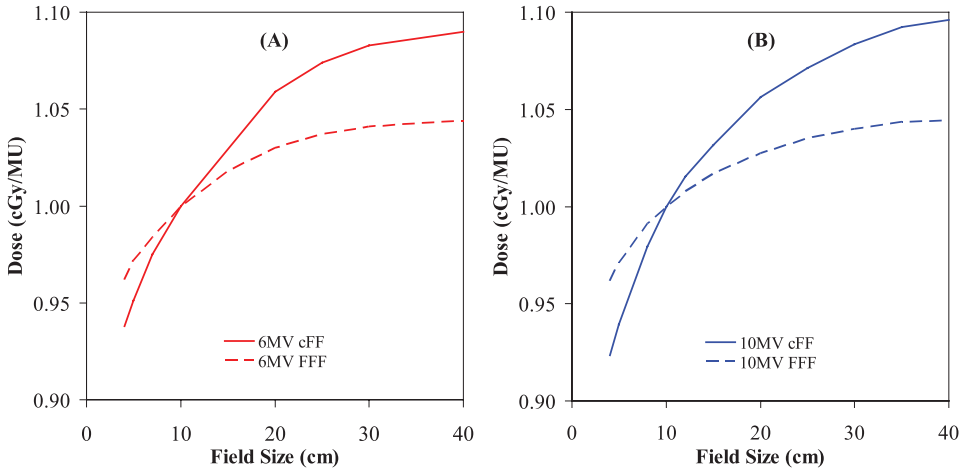

Figure 3. cFF and FFF relative output factors for (A) 6 MV Elekta and (B) 10 MV Varian at d_{max}.

2.1.4. Scatter factors. Both collimator scatter and phantom scatter (S_c and S_p respectively) will be affected by the change in energy spectrum of the FFF beam. Figure 3 is a comparison of total output factors for both cFF and FFF beams. As can be seen, due to the reduction in scatter, the variation of total output factor with field size is much less pronounced than that for a flattened field.

3. Radiation protection considerations

Adequate primary shielding of a linac bunker depends on the radiation workload, x-ray beam energies and dose rates. Secondary shielding will depend also on head leakage and scatter. Scatter from the walls and patient giving rise to the dose rate at the maze is dependent on energy and field size. Bunker design will also be influenced by occupancy of adjacent areas.

FFF beams are delivered at a higher dose rate (currently 1200 MU min^{-1}–2400 MU min^{-1}) compared with cFF beams (typically up to 600 MU min^{-1}). Introduction of FFF beams in an existing linac bunker will therefore increase the maximum dose rates (measured with the beam on) substantially. Annual doses however may be largely unaffected and are likely to be more dependent upon the clinical use of the linac than the higher dose rates specifically.

3.1. Primary beam

The energy spectra of the primary beam will change due to the removal of the flattening filter, and it is important to understand if the linac adjusts the accelerating potential for FFF mode (see section 2). It is also therefore important to understand the FFF nomenclature used when considering radiation protection for FFF.

If the accelerating potential is not adjusted (i.e. Varian linacs), the beam energy spectrum of the FFF beam will be softer than the flattened beam, the TVLs shorter and shielding requirements may be reduced; Kry *et al* (2009) report a reduction of 12% in TVLs and 10–20% in shielding thickness assuming all treatments are in FFF mode.

If the accelerating potential is adjusted to match the $TPR_{20/10}$ of the flattened beam (i.e. Elekta linacs), there could be increased penetration due to the higher maximum energy required. However, this is offset by the change in beam spectrum. The literature suggests the overall outcome is energy dependent, with Kry *et al* (2009) reporting a reduction of 7–11% for 6 MV. Paynter *et al* (2014) also report reduced penetration for 6 MV matched FFF beams, but report an increased penetration for 10 MV matched FFF beams when compared to cFF beams.

3.2. Legislative considerations

The Ionising Radiation Regulations 1999 (IRR 1999) are designed to protect workers and members of the public from work involving ionising radiation. They include annual dose limits in Regulation 11 which must not be exceeded. Regulation 8 describes many ways in which exposures should be restricted and doses kept to levels as low as reasonably achievable i.e. optimisation of exposure. The use of dose constraints when planning facilities can be used to meet this requirement. The latest guidance on the use of dose constraints in planning describes the use of a constraint for members of the public from a single source to be a maximum of 0.3 mSv per annum (HPA 2009). This figure is the accepted value for design of linear accelerator bunkers in the UK.

Calculations are carried out to determine the primary and secondary barrier thickness to reduce the exposure outside the bunker to the planned annual dose constraint, taking into account expected use of the linac and adjacent areas. It is often prudent to attempt to build contingency into the shielding for foreseeable future developments in, for example linac use, changes in patient selection/dose and use of adjacent areas, however a decision must be made at a local level regarding the cost of contingency shielding versus the future flexibility of the facility.

Many centres also review the predicted instantaneous dose rate at critical points around the installation resulting from the barrier thicknesses deduced from those calculations. These can indicate numerical values of some 10s of $\mu Sv\ h^{-1}$. Restricting this value to for example 7.5 $\mu Sv\ h^{-1}$ will lead to more shielding being installed than is required.

Controlled areas are defined in Regulation 16(1) as being areas where special procedures are required to restrict significant exposure to an individual in that area or to limit the probability of a radiation accident and to limit its magnitude; or if it is likely that an individual in that area would receive an effective dose in excess of 6 mSv yr^{-1} or three-tenths of any other dose limit for a radiation worker aged 18 years or over.

Provided the annual dose in areas outside the bunker meets the constraint then there is no need to designate areas as controlled under the regulations. This is in line with the views expressed in the BIR report *Radiation Shielding for Diagnostic Radiology* (Sutton *et al* 2012a) which uses only a design constraint of 0.3 mSv yr^{-1} (three tenths of 1 mSv yr^{-1}), with no reference to time averaged dose rates over a minute. This view has been further clarified in a letter in the Journal of Radiological Protection (Sutton *et al* 2012b) that states that the 0.3 mSv yr^{-1} constraint should be adhered to but that a 7.5 μSv h^{-1} instantaneous dose rate averaged over a minute (IDR) constraint is not considered valid for diagnostic radiology. It is the view of this working party that using this IDR constraint in shielding calculations for linear accelerator bunkers using FFF beams is also not valid.

Prior risk assessment is also required under Regulation 7. Consideration of the workload, case mix and modes of operation will determine whether additional measures are required. It is recommended that a detailed risk assessment is carried out for both commissioning use and clinical use, which is very clear about the scope of the assessment in terms of delivered dose to isocentre or MU, and pays particular attention to higher energies. Regular audit or measurements can be used to confirm that the risk assessment is still valid.

3.3. Beam profile

The FFF beam profile gives a much greater dose at the central axis than off axis, which has a number of effects that should be considered in radiation protection.

(1) Within any shielding material, more photons are scattered away from the central axis than towards it, reducing the effect of the increased dose rate.
(2) The maximum dose rate will only be realised at the centre of the beam. This is an important consideration when measuring dose rates of primary beams. Therefore measurement using an FFF beam alone may not be adequate for checking that the lateral extent of a primary barrier is adequate. In addition, any extra shielding required for FFF may not need to cover the full lateral extent of the cFF primary barrier.
(3) Due to the beam profile, IMRT is generally considered essential for larger FFF fields, bringing the potential for increased use of IMRT, or the possibility of increased modulation compared to IMRT with flattened beams, and with it an increase in MU Gy^{-1} required for treatment. This will primarily affect secondary barrier and maze entrance dose rates

3.4. Patient scatter

The change in energy spectrum affects patient scatter considerations. Whilst one would expect patient scatter to be increased due to a decrease in average energy, Kry *et al* (2009) report that the reduction in beam energy of their Varian FFF beams led to greater patient attenuation and reduction in collimator scatter, which affected the scatter dose to a greater degree, hence reducing patient scattered dose. In addition, if smaller fields are used for FFF in IMRT and VMAT this will result in a lower patient and wall scatter contribution per MU to the maze entrance dose than with cFF.

3.5. Leakage

Due to the removal of the flattening filter from the beam, the current required per MU has been reported to decrease by 57% at 6 MV FFF (Vassiliev *et al* 2006b) and therefore significant reductions in leakage have been reported (50% Kry *et al* 2009, 58% Cashmore 2008). It is also likely that there is a reduction in head scatter due to there being less material in the beam. The increased use of IMRT for large FFF beams is however likely to increase, negating some of the reduction in leakage. In clinical use, secondary barriers and mazes are likely to be adequate for FFF beams of the same nominal energy as the bunker was designed for. However any substantial increases in MU due to sliding window IMRT (IMRT factor) or volumetric arc therapy (VMAT)/RapidArc (and to a lesser extent step and shoot IMRT) will need to be assessed with secondary shielding and maze scatter in mind. As outlined by Jank *et al* (2014), the IEC guidelines (IEC 2007) state a maximum leakage of 0.1% of isocentre dose. If in the future a linac produces FFF only, it may be that a linac manufacturer will reduce head shielding; this should be specified by the manufacturer. Leakage measurements should therefore be carried out on each individual installation as part of the critical examination as required by the legislation.

3.6. Neutron production

Neutrons are not considered to be generated in significant fluences below 10 MV hence will not be an issue for 6 MV FFF beams. Consideration should be given to neutron production for FFF beams matched to 10 MV flattened beams, as the maximum energy will be capable of neutron production. Kry *et al* (2008) found that removal of the flattening filter for a non-corrected 18 MV beam reduced the number of neutrons produced per MU by a factor of 3.7. However, these high energies are not commercially available for FFF.

3.7. Clinical use

Annual dose surrounding a linac bunker is directly related to its clinical use. It is primarily the total isocentre dose delivered per year that will determine how the measurable dose rate relates to annual dose. Increase patient throughput from reduced treatment times using FFF alone is unlikely to significantly increase annual dose, as beam on time constitutes a small proportion of the clinical day (found to be 4–6% with flattened beams at the Christie NHS Foundation Trust in 2012—author's own data).

However, FFF beams are suited to high dose per fraction treatments, which could significantly affect the annual clinical workload and dose, particularly if one linac is used for all high dose per fraction treatments in a department. One factor that may negate this to some extent is that due to the short fractionation regimes, the imaging requirements for these treatments may be more time consuming than for standard treatments, however as experience is gained, this time may reduce. Consideration should also be given to the potential for a future change in dose per fraction, patient numbers (especially those receiving large doses per fraction), manufacturer or method of FFF application when designing bunker shielding. A risk assessment should be done if any of these variables change significantly.

As the adequacy of shielding may be heavily dependent upon FFF use, it is important to assess the effect on annual dose rate of any changes in use, including unintentional changes. There are two methods recommended for this purpose.

(1) Environmental monitoring to measure dose over long time periods (e.g. months) with passive detectors (e.g. film badges), undertaken at a suitable frequency or after known changes.
(2) Regular usage audits, such as methods using RT management systems, which can be queried to look at details such as the proportion of treatments pointing at a particular barrier. This data can then be used to compare actual use to intended use, to use during periods of environmental monitoring & subsequently adjust predictions of annual dose beyond a particular barrier.

3.8. Commissioning

The total dose over the course of acceptance testing and commissioning is likely to be substantially higher than for commissioning of flattened beams, therefore a detailed radiation risk assessment for individual commissioning periods should be carried out. As a guide, commissioning of FFF beams on one linac required similar beam on time to standard commissioning despite the higher dose rate (personal communication G Budgell). As such, a scaling of the monitor units used for standard commissioning in line with the increase in dose rate is likely to give an appropriate estimate of MU required for FFF commissioning. A knowledge of expected MU along with dose to personnel per MU can be used within a prior radiation risk assessment to identify expected doses to personnel during commissioning. Keeping a record of MU used during commissioning could be a useful radiation protection measure to estimate doses and inform future risk assessments.

3.9. Summary

If the only expected change from an FFF beam is an increase in instantaneous dose rate, not an increase in patient dose or throughput, and if the shielding is sufficient for the energy of the machine being installed, then no further increase in primary shielding is likely to be needed for FFF. Due to the reduction in required current per MU, the secondary shielding present is also likely to be sufficient provided there is no large change in the IMRT factor. However use of FFF for high dose per fraction treatments may lead to a higher annual dose rate. Special measures may be required for acceptance and commissioning periods. Radiation surveys should usually be carried out soon after installation and environmental monitoring with passive detectors carried out once FFF is in clinical use, to validate both assumptions and calculations.

This guidance does not constitute radiation protection advice, and employers should consult their own Radiation Protection Advisor for advice on compliance with the legislation for their individual situation on those aspects of work detailed in schedule 5 and ACOP13 (1)–(3) of IRR99. For FFF upgrades, this should include advice on performing appropriate radiation risk assessments, environmental monitoring and review of designation of areas.

4. Commissioning

The process of commissioning an FFF beam for clinical use should, in essence, be identical to that of a cFF beam. However, due to the changes in some of the physical characteristics of FFF beams, extra care must be taken to ensure that accurate data for the commissioning and validation of the TPS dose algorithm is obtained during beam data acquisition. The following section is intended to provide guidance on the potential issues that are particular to the commissioning of FFF beams.

4.1. Beam data acquisition for FFF beams

There are numerous studies in the literature of the physical characteristics of FFF beams (Georg *et al* 2011, Hrbacek *et al* 2011, Fogliata *et al* 2012), with the principal differences from conventional flattened beams being as follows.

(1) A change in beam profile shape from centrally flattened with steep penumbral gradients, to centrally peaked with dose gradients in both the penumbral region and what would traditionally be the 'flattened' region of the beam.
(2) An increase in dose rate (and a corresponding increased dose per pulse) due to reduced beam filtration.
(3) A softer beam energy spectrum due to reduction in beam hardening.

These physical differences, however, do not introduce any additional measurements to beam data acquisition when commissioning an FFF enabled linac, as current commercially available TPS systems do not require any measurements specific to FFF to produce a functional beam model. However, these new physical characteristics have consequences for the selection of measuring equipment, and lead to quantitative differences in the data acquired when compared to a cFF linac.

4.1.1. Choice of radiation detectors. The choice of measurement detector for beam data acquisition is heavily influenced by the dose per pulse (DPP) and dose gradients that are found in an FFF beam. DPP has a large effect on recombination losses (Boag and Currant 1980) when measuring with an ionisation chamber; an effect which generally speaking is usually only considered of consequence in absolute dosimetry and not the relative dosimetry measurements that make up the bulk of beam data acquisition. FFF beams, however, have a much higher DPP than a conventional flattened beam (for example, the DPP for 10 MV FFF in a Varian TrueBeam is approximately 4 times higher than that for the flattened 10 MV beam). As a result, ionisation chambers operating in an FFF beam may display significant differences in ion collection efficiency than when operating in a filtered beam (Lang *et al* 2012).

To compound this problem, the DPP varies not only with depth (so has consequences for PDD measurements) but also across the beam profile, given the dose gradients inherent to an FFF beam. These recombination loss effects may have a serious impact on the accuracy of measured data during commissioning, depending on their severity for a given ionisation chamber (Corns *et al* 2015). To illustrate this, figure 4 shows the potential effects that these recombination losses may have on PDD measurements in an FFF beam (determined from ion recombination and PDD measurements from a Varian TrueBeam). As can be seen, the recombination loss effect can be considered to be almost negligible for the cFF beam (hence similar measurements will be obtained regardless of detector choice), but has significant variation amongst the ionisation chambers when placed in an FFF beam. This recombination loss effect is also influenced by the chamber size—smaller chambers tend to have lower recombination losses as recombination varies as a function of electrode separation (Boag and Currant 1980).

The dose gradients present in the FFF beam also influence the choice of detector. Unlike the flattened beam, the FFF beam has dose gradients in the central beam, not just the penumbral region, which has the following significance.

(1) For profile measurements, the physical size of the detector must be small enough to accurately measure the position of the central axis peak, and to not be unduly influenced by dose averaging in the volume.

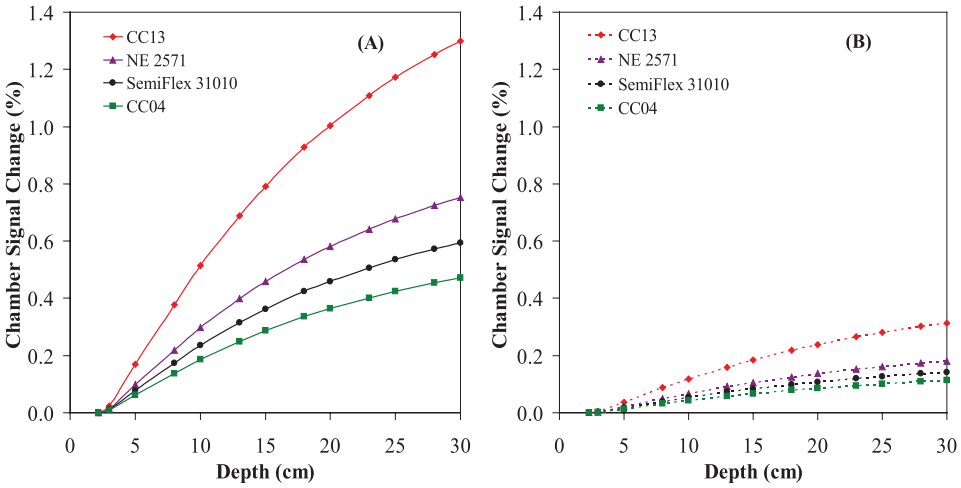

Figure 4. Effect on ionisation chamber signal at depth due to recombination losses for various detectors for (A) 10 MV FFF photons and (B) 10 MV flattened photons for a Varian TrueBeam. Measurements are for a 10×10 cm field at 100 cm SSD, using an operating potential of −350 V for all chambers. The ion chamber volumes are respectively: IBA CC13 0.13 cm^3, NE 2571 0.6 cm^3, PTW 31010 0.125 cm^3, IBA CC04 0.04 cm^3.

(2) For depth dose measurements, the physical size of the detector must be small enough that the change in dose gradient with depth does not unduly effect dose averaging within the detector volume.

A small volume chamber is therefore required for accurately measuring the shape of the profile of an FFF beam.

Based on these ion recombination and dose gradient effects, it is recommended that beam data acquisition for FFF beams be made with small-volume, 'spherical' ionisation chambers. These chambers typically show a low variation in ion recombination across the DPP range, and possess adequate spatial resolution to accurately scan profiles without being influenced by dose volume averaging effects. These chambers are typically used in small field dosimetry and, as such, care must be taken to ensure that any measurements made are within the recommended field size range stated by the manufacturer (or the performance is benchmarked against another detector at cross-over points between small and large fields) to limit any potential issues with stem effects on the acquired data. Axial irradiation of the detector may assist in minimising such effects. The use of semiconductor diode detectors for beam data acquisition may be feasible, but due to the inherent variation in diode signal response with beam energy, and therefore a tendency to over-respond to lower energy scatter radiation at increasing depth and with increasing field size, it is advisable to benchmark detector performance against a small-volume ionisation chamber under 'worst case' conditions (i.e. maximum field size and depth combination for profile measurements, and maximum field size for PDD comparisons) prior to acquisition of commissioning data.

4.1.2. Practical aspects of beam data acquisition in FFF beams. There are a few practical considerations to take into account when acquiring beam data in an FFF beam, primarily related to the dose rate involved. Prior to acquiring beam data in a scanning water tank, it is advisable to check that the electrometer is capable of handling the maximum dose rate

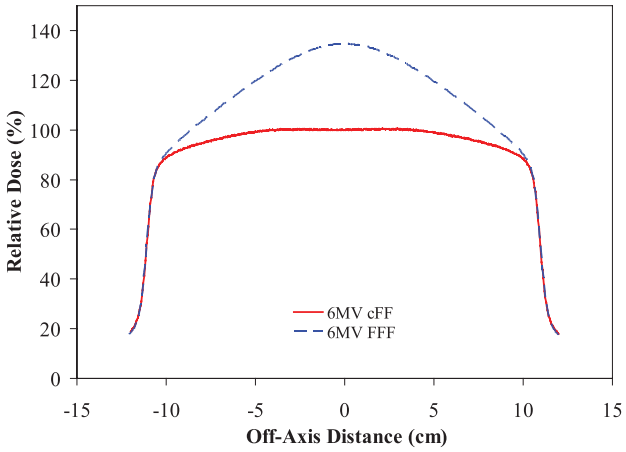

Figure 5. 6 MV FFF beam profile renormalised to give the same field size as the 6 MV cFF beam profile for the same linac.

provided by the machine without saturating—this can easily be tested by checking the dose linearity for the detector/electrometer combination used for the largest field size. Such saturation issues may be more prevalent if attempting to use semiconductor diodes, due to the inherently higher signal produced by such detectors. In addition, the high dose rate means that when measuring profiles and depth doses, a very large number of monitor units is required (as mentioned in section 3). This can potentially be reduced by acquiring data with a lower dose rate, but only if the DPP has been verified as being constant with dose rate and hence the recombination losses are identical.

4.2. Field size and penumbra definitions

The peaked profile of FFF beams causes issues with standard field size and penumbra definitions as the 50% and 80% dose levels are often no longer located at the beam edge. In order to apply standard definitions to FFF beams the field edges of the cFF and FFF beams must be made to coincide, for example by renormalising the FFF beam profile (figure 5).

A number of renormalisation methods have been proposed in the literature to achieve this. The inflection points of the curve in the penumbra (point of highest dose gradient) can be used to define the position of the field edge, however, this can lead to high uncertainties if the profile has been measured using a low resolution ($\geqslant$1 mm). Alternatively, Fogliata *et al* (2012) suggested renormalising the FFF beam to the 'shoulder point' of the profile (the point of maximum dose curvature), which can be calculated from the third derivative of the profile. This is often impractical due to inherent noise in scanned or film data. A simpler approach is to scan the FFF beam and cFF beam at the same time, identify the shoulder point by eye and renormalise the FFF profile to the value of the cFF profile at that point. When this has been done once, the value required for renormalising the FFF beam is known and can be re-applied for further measurements. For routine measurements this will only require establishing for a single depth and the limited number of field sizes used for QC checks, and values are expected to be similar for linacs of the same type. These values have been tabulated by Fogliata *et al* (2012, 2016) for both Varian and Elekta linacs.

These methods, however, require the user to have a conventional beam of the same energy for setup, calibration and QA of the FFF beam. This is not ideal as it requires the maintenance

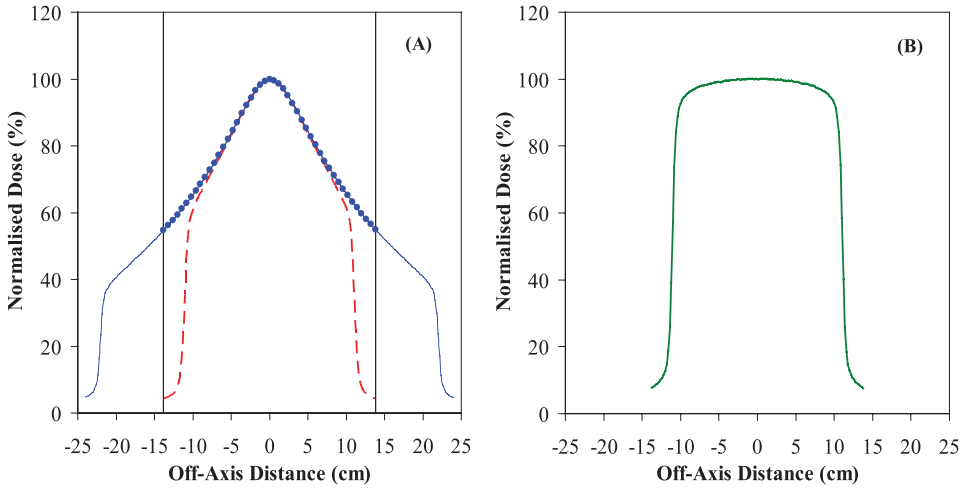

Figure 6. Graphical representation of the method for defining FFF field sizes. In (A), the 22 cm FFF beam profile (dashed) is divided by the 40 cm FFF beam profile (dotted) to give (B), a flat beam profile for which a conventional 50% field edge definition can be used.

of a beam energy that may not be required for clinical use. Some UK centres already operate their FFF capable linacs exclusively in FFF mode and it is assumed that this will become a more common practice as SRS/SBRT and IMRT techniques become more commonplace. A method is therefore required that does not rely on the use of the cFF beam, allowing FFF to operate independently.

This report recommends a simple but previously unpublished approach in which the profile of an FFF field is divided by the profile of the fully open FFF field, and the conventional definitions are applied to this ratio of measured profiles. The profile of the fully open FFF field is used, in a virtual sense, to flatten the FFF beam. The steps are as follows.

(1) Measure the profile of the FFF field of interest, in the radial or transverse direction.
(2) Measure the profile of the fully open FFF field, in the same direction and with the same scan resolution as for the profile of interest.
(3) Normalise the profiles to their respective central axis values.
(4) Take the ratio profile (1)/profile (2), and apply the conventional definitions of field size and penumbra to this profile instead of to the original profile (1).

This is illustrated graphically in figure 6.

The FFF beam profile (in this example a 22 cm wide 10 MV FFF profile shown in red dash) is divided through point by point by the profile of the maximum field size for the same beam (in this example a 40 cm wide 10 MV FFF profile shown in blue—part used for the division is dotted). This yields a virtually flattened beam profile (shown in green) to which the conventional field size and penumbra definitions can be applied. Figure 7 shows this derived beam profile overlaid onto a cFF profile from the same linac and shows that the 50% points match closely. Note that this approach can even be applied in the case of maximum square field size if the profile in step (2) is measured using a collimator angle of 45° or 315°.

This method has been tested by the authors using both water tank and film measurements and has been shown to give agreement within ± 0.5 mm of cFF fields for field sizes up to a 30×30 cm field on both Elekta and Varian linacs.

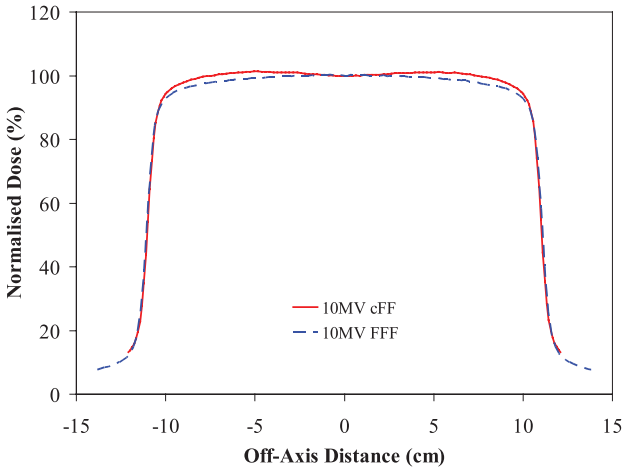

Figure 7. The derived FFF beam profile overlaid onto 10 MV cFF beam profile from the same linac with the same collimator settings.

When using these approaches, all of which ensure the field edges of cFF and FFF beams superimpose and match at 50%, then the standard definition of beam penumbra (20–80%) can be used.

4.3. Treatment planning system

Removal of the filter invalidates any of the standard assumptions of energy fluence and scattered radiation in the beam, so a treatment planning system (TPS) must have sufficient flexibility to model these changes. It is therefore important to be able to handle the incident particle fluence either by means of a sufficiently general calculation algorithm or through modelling of changes in the beam spectrum with off-axis position.

TPS commissioning for FFF beams will follow the standard procedures as for conventional beams, and most manufacturers now offer systems capable of handling FFF beam modelling and planning (Stathakis *et al* 2009, Cashmore *et al* 2011, 2012, Hrbacek *et al* 2011, Kragl *et al* 2011a, Muralidhar 2015).

To create the FFF beam model the existing 6 MV cFF model for a linac can generally be used as a starting point since the basic machine parameters for the model (machine limits etc) obviously remain the same. A new beam spectrum, depth doses and profiles along with total scatter ($S_{c,p}$) and phantom scatter (S_p) factors are required. In practice many users have traditionally referred to published S_p data, but these are not valid for FFF beams (Satherberg *et al* 1996, Kragl *et al* 2009), therefore $S_{c,p}$ and S_c should be measured explicitly.

As yet there is little data published on spectra for FFF beams, but conventional central axis spectra are simple to derive from standard published spectra (Mohan *et al* 1985). Some off-axis spectra have also been published for standard beams (Sheikh-Bagheri and Rogers 2002) and are used to modify attenuation towards the field edge, but none are currently available for FFF beams. However, although there is still a variation in beam spectrum across the field with the flattening filter removed it is much reduced, and a separate spectrum is often not required to model the off-axis component and fit to measured profiles. Spectral changes will also affect transmission factors for MLCs, collimators, shadow trays and treatment couches (Gardner *et al* 2015) etc, all of which should be investigated to provide accurate data for the TPS.

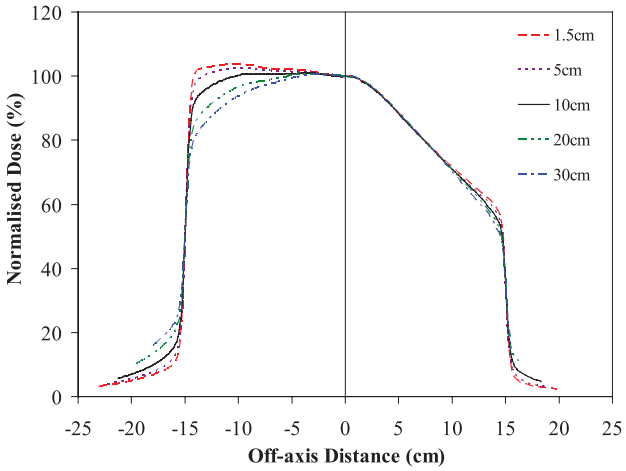

Figure 8. A half-beam profile plot comparing a 6 MV cFF beam (left of plot) against the energy matched FFF equivalent (right) for a $30 \times 30\,cm^2$ beam at varying depths. Profiles are normalised to 100% on the central axis and scaled to remove divergence.

It is important to remember that FF removal, with or without beam energy matching, changes not only the primary beam spectrum, but also the scattered beam spectrum within the patient. It is currently unclear to what extent this has been studied, and whether scatter kernels utilised for convolution/superposition algorithms within the TPS need to be modified (Azcona *et al* 2016).

At small field sizes the filter has no observable effect on the beam shape, but as the size increases the forward-peaked nature of the beam profile becomes more apparent. Figure 8 shows a selection of beam profiles for a $30 \times 30\,cm^2$ field normalised to remove beam divergence and output. In the shoulder region the beam profiles for the conventional beam show considerable variation due to the changing beam spectrum. This is why flattening filter can only be designed to give a 'flat' field at one particular depth. The FFF plot shows the reduction in this variation due to the similarity of the energy spectrum across the portal.

Due to the reduction in scattered radiation from the treatment head the beam penumbra can potentially be sharper in an FFF beam (Pönisch *et al* 2006, Cashmore 2008, Kragl *et al* 2009, Yarahmadi *et al* 2013) although differences reported are small and likely to be of little significance clinically.

Surface doses will be different from cFF beams due to the reduction in electron contamination and increase in the low energy photon component. Hence they should be modelled carefully and any tabulated skin doses (for open beam or through treatment couches) re-measured for FFF beams. Due to the reduction in effective beam energy the Varian linac is observed to have higher surface doses for FFF than for cFF beams at all field sizes. When the beam energy is matched it is generally seen that FFF surface doses are higher for smaller fields and lower for larger fields with a crossover point somewhere in the middle (Vassiliev *et al* 2006a, Cashmore 2008, Kragl *et al* 2009, Wang *et al* 2012). In all FFF beams the general reduction in scattered radiation means that the variation in field size dependent factors such as surface dose will be less.

Dose outside of the field edge has been studied by several authors (Kry *et al* 2010, Cashmore *et al* 2011, Kragl *et al* 2011b, Almberg *et al* 2012) and will be lower for FFF beams due to reduced head scatter and leakage radiation. Treatment planning systems are generally poor at predicting peripheral doses with any degree of accuracy; further study of out-of-field and

surface doses and their modelling is needed. Peripheral doses are discussed in more detail in later sections.

There is very little information concerning the use of wedges with FFF, but this is certainly possible in principle using either physical or virtual methods, and has been demonstrated in conformal planning studies (Cashmore 2007, Stevens *et al* 2011, Kretschmer *et al* 2013). However, at this time there are no commercial systems offering this functionality.

In general the reduction in scattered radiation from the treatment head, and the reduction in variation of beam spectrum across the field means that variations in the beam characteristics with field size are reduced for FFF beams. This means that the FFF beam model has less variation, and it is therefore possible that FFF beam models may be more accurate than their cFF counterparts (Kragl 2011a, Cashmore *et al* 2012).

5. Reference dosimetry

5.1. Introduction

Traceable dosimetry in the clinic depends on the transfer of a dosimetry standard, such as an ionisation chamber, from a beam in the primary calibration laboratory to the beam in the clinic. Since the chamber calibration coefficient depends on beam quality, it is necessary to quantify any difference in calibration between the two beams. The IPEM 1990 Code of Practice (CoP, Lillicrap *et al* 1990) implicitly assumes that the beam quality index, $TPR_{20/10}$, completely characterises beam quality for the purpose of calibrating the recommended type of secondary standard (SS) chamber, i.e. the NE2611[7]. With the introduction of FFF beams into clinical use it has become important to understand limits on the validity of this assumption. When a chamber is used for measurements in two beams that have the same beam quality index, it is still possible for the beams to differ in beam quality. In this case a correction is required for any resulting difference in chamber sensitivity between the two beams. One can understand that such a difference in chamber sensitivity may arise if the two beams have different energy spectra, which is possible even when they happen to have the same beam quality index.

The relative importance of the various sources of uncertainty in reference dosimetry depends on the characteristics of the beam in which measurements are made. So it is to be expected that, because the characteristics of FFF beams differ from those of cFF beams, the uncertainty of reference dosimetry in FFF beams will also be different. For example:

- dosimetry in an unflattened beam will necessarily involve a correction for beam non-uniformity across the measuring device, whereas in a cFF beam this correction should be negligible;
- typical dose rates in FFF beams are much higher than in cFF beams so that the effects of ion recombination may be much more important.

As indicated below, the uniformity correction can be made very small by an appropriate choice of reference detector. However the correction for ion recombination may be so large that the associated uncertainty is quite significant.

This section provides recommended values for the beam quality correction between cFF and FFF beams produced by a conventional linac, for a chamber of type NE2611, and presents

[7] The type originally recommended, the NE2561, was designed by NPL and manufactured by Nuclear Enterprises. That design was superseded by the radiologically equivalent type NE2611, however manufacture and repair of the NE2611 chamber type was taken over by NPL when Nuclear Enterprises stopped production. New secondary standard chambers are designated as being of type NPL 2611.

a worked example of the cross-calibration of a field chamber and its use to measure absorbed dose, including an analysis of uncertainty.

5.2. Absorbed dose measurement in FFF beams

Traceability is achieved through cross-calibration of a suitable field chamber (FC) against the secondary standard (SS), by substitution, in the user's static FFF beam, under reference conditions[8]. The IPEM 1990 CoP gives the simplest possible expression for a measurement of absorbed dose D under reference conditions:

$$D = N_{\mathrm{D}} R \tag{1}$$

where D is absorbed dose to water at the position of the centre of the chamber when the chamber and sheath are replaced by water, R is the fully corrected chamber reading and N_{D} is the chamber calibration coefficient for the radiation quality to convert the corrected reading to absorbed dose to water. More recent codes of practice, including TRS-398 (Andreo *et al* 2000) and TG-51(Almond *et al* 1999), make clear the dependence of terms in this expression on beam quality Q, on the absorbing medium and on the chamber type. In the same way, equation (1) may be written using a more explicit notation:

$$D_{\mathrm{w}}(Q) = N_{\mathrm{D,w}}^{\mathrm{cham}}(Q) R^{\mathrm{cham}}(Q). \tag{2}$$

The FC calibration coefficient is derived by combining measurements made using a calibrated SS with measurements made using the FC. Equation (2), applied to the two measurements, leads to the identity

$$N_{\mathrm{D,w}}^{\mathrm{FC}}(Q) R^{\mathrm{FC}}(Q) = N_{\mathrm{D,w}}^{\mathrm{SS}}(Q) R^{\mathrm{SS}}(Q). \tag{3}$$

This may be rearranged to give the usual expression for the FC calibration coefficient:

$$N_{\mathrm{D,w}}^{\mathrm{FC}}(Q) = N_{\mathrm{D,w}}^{\mathrm{SS}}(Q) \frac{R^{\mathrm{SS}}(Q)}{R^{\mathrm{FC}}(Q)}. \tag{4}$$

Equation (4) makes explicit the assumption in the 1990 CoP that the SS calibration coefficient is already available for the user's beam quality, Q. This is not the case for FFF beams because, currently, the NPL calibration is only provided for cFF beams and a correction $k_{\mathrm{FFF}}^{\mathrm{SS}}$ may be required for the change in sensitivity of the SS chamber between cFF and FFF beams:

$$k_{\mathrm{FFF}}^{\mathrm{SS}} \equiv \frac{N_{\mathrm{D,w}}^{\mathrm{SS}}(Q_{\mathrm{FFF}})}{N_{\mathrm{D,w}}^{\mathrm{SS}}(Q_{\mathrm{cFF}})}. \tag{5}$$

In this case, the equation for the measurement of absorbed dose to water in an FFF beam using a SS becomes

$$D_{\mathrm{w}}(Q_{\mathrm{FFF}}) = N_{\mathrm{D,w}}^{\mathrm{SS}}(Q_{\mathrm{cFF}}) k_{\mathrm{FFF}}^{\mathrm{SS}} R^{\mathrm{SS}}(Q_{\mathrm{FFF}}) \tag{6}$$

and equation (4) becomes

[8] The reference conditions in FFF beams are unchanged from the IPEM 1990 CoP: the field size is $10 \times 10\,\mathrm{cm}^2$ at the measurement depth, which is $5\,\mathrm{g \cdot cm^{-2}}$ in water. Calibration by substitution is recommended so that both chambers may be positioned on the beam central axis. The recommendations given here are intended for beams having energy 10 MV or below, but this limit is expressed in terms of the quality index as the condition $\mathrm{TPR}_{20/10} \leqslant 0.750$.

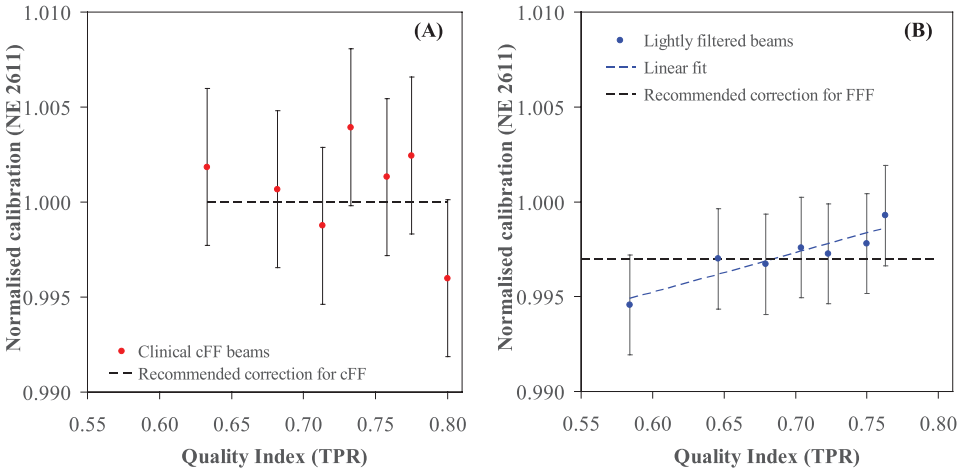

Figure 9. Calibration coefficients for an NE2611 secondary standard chamber, measured directly using the NPL primary standard calorimeter and normalised by the calibration coefficient as currently disseminated by NPL: (A) measured in clinical cFF beams; (B) measured in non-clinical FFF beams.

$$N_{\mathrm{D,w}}^{\mathrm{FC}}(Q_{\mathrm{FFF}}) = N_{\mathrm{D,w}}^{\mathrm{SS}}(Q_{\mathrm{cFF}})k_{\mathrm{FFF}}^{\mathrm{SS}}\frac{R^{\mathrm{SS}}(Q_{\mathrm{FFF}})}{R^{\mathrm{FC}}(Q_{\mathrm{FFF}})}. \tag{7}$$

With this calibration coefficient, the equation for the measurement of absorbed dose to water in an FFF beam using a FC is simply

$$D_{\mathrm{w}}(Q_{\mathrm{FFF}}) = N_{\mathrm{D,w}}^{\mathrm{FC}}(Q_{\mathrm{FFF}})R^{\mathrm{FC}}(Q_{\mathrm{FFF}}). \tag{8}$$

There is no need for a $k_{\mathrm{FFF}}^{\mathrm{FC}}$ correction because the FC has been calibrated in the beam of interest.

5.3. Chamber types

The NE2611 remains the recommended SS chamber type. For measurements made using an ionisation chamber of this type in FFF beams produced by a conventional linac, we recommend that the quality-dependent correction for FFF beams be assigned a value $k_{\mathrm{FFF}}^{\mathrm{SS}} = 0.997 \pm 0.003$ (with a confidence probability of 95%), as discussed in the next subsection.

The FC must be suitable for reference dosimetry in MV photon beams: its calibration must be stable over a period of at least 1 year and the response must be rotationally symmetric. As far as the FC is concerned, there are no additional requirements associated with reference dosimetry in FFF beams beyond those for cFF beams.

5.4. Beam quality and quality index

Tissue phantom ratio $\mathrm{TPR}_{20/10}$ is recommended for use as the quality index (QI) in both FFF beams and cFF beams. Two beams may have the same QI but nevertheless differ in their beam quality: any difference in beam quality is likely to affect chamber calibration coefficients. This is illustrated in figure 9, which shows the calibration coefficient of an NE2611 chamber, measured directly with the NPL primary standard calorimeter and normalised, as a function of

QI, by the calibration coefficient as disseminated by NPL. If it differs from unity, this ratio of calibration coefficients would be the correction required because the measured beam has a different quality from the calibration beam. The calibration in clinical cFF beams was measured in the Elekta linac at NPL (Pearce *et al* 2011) and the calibration in lightly filtered beams was measured in the research linac formerly used at NPL. The uncertainties in the figure indicate the consistency of the measured ratio with the recommended correction, with a confidence probability of 95%. For the cFF data, the measured ratio is indeed consistent with unity. The 'lightly filtered' beams have an inherent filtration which is even less than is found in currently available clinical linacs delivering FFF beams[9]. Accordingly, the quality-dependent correction for clinical FFF beams is expected to lie between the plotted points and unity. The value $k_{\mathrm{FFF}}^{\mathrm{SS}} = 0.997 \pm 0.003$ is recommended for FFF beams provided the QI less than 0.750. This includes all FFF beams currently in routine clinical use. This is a preliminary recommendation, pending the availability of measured data in a representative range of FFF beams in clinical use. Note that previously, NPL certificates have included either no mention of FFF at all or an explicit recommendation to take $k_{\mathrm{FFF}}^{\mathrm{FC}}$ as unity.

5.5. Reference conditions

Reference dosimetry should be carried out in a full scatter water phantom. Solid materials may be used in place of water for routine measurements, but their water-equivalence must be verified. The recommended measurement depth is $5\,\mathrm{g}\cdot\mathrm{cm}^{-2}$, at which depth the reference field size is $10 \times 10\,\mathrm{cm}^2$. The reference point for the FC is at the geometric centre of the chamber cavity, the same as for the SS. The QI is the value of $\mathrm{TPR}_{20/10}$, which must be determined in a $10 \times 10\,\mathrm{cm}^2$ field[10], and the recommendations given here only apply if the QI is less than 0.750.

5.6. Other corrections

5.6.1. Beam non-uniformity. The SS chamber is calibrated to give absorbed dose to water at the position of the geometric centre of the chamber, in homogeneous water, in a cFF beam. The transfer of this calibration to an FFF beam is defined here so that the measurement in the FFF beam also gives absorbed dose, and at the same point. This requires that any effect on the SS chamber response due to the non-flat profile of an FFF beam must be corrected. In fact the curved profile reduces chamber response for two distinct reasons: volume averaging, and perturbation of the secondary electron fluence by the chamber air cavity. It turns out that these two effects may be comparable in magnitude (Azangwe *et al* 2014). For a thimble chamber, the effect of volume averaging is approximately proportional to the square of the length of the chamber sensitive volume. For the NE2611 chamber in 6 MV FFF and 10 MV FFF beams, the effect of volume averaging is of the order 0.1% and may be considered negligible. In the absence of further information, the perturbation effect for the NE2611 chamber in FFF beams may be approximated by unity, and assigned a standard uncertainty of 0.2%. Taking these effects together, the recommended value for the beam uniformity correction for the SS used in FFF beams is unity, with a standard uncertainty of 0.2%. It is recommended that no correction for field uniformity be applied to FC readings, whether during cross-calibration or

[9] This is because the NPL research linac x-ray target was thinner than is typical of clinical linacs. Although the beam was flattened, the maximum field size was only 10 cm and the flattening filter was very thin.

[10] If for some reason it is not possible to set a $10 \times 10\,\mathrm{cm}^2$ field size in the FFF machine then a field size correction factor must be applied to convert the $\mathrm{TPR}_{20/10}$ value, measured in the machine specific reference (msr) field, into the QI. This case was considered explicitly for TomoTherapy machines by Thomas *et al* (2014).

during subsequent measurements under reference conditions: the FC calibration factor is then valid only under reference conditions or when the effect of beam non-uniformity is the same as under reference conditions. In this approach, the correction for field uniformity, which for an NE2571 chamber may be of the order 1%, is effectively incorporated into the FC calibration coefficient.

5.6.2. Saturation. Ion recombination is significant in the high dose rates typical of clinical FFF beams. Under these conditions, volume recombination completely dominates initial recombination and the total ion recombination may be assumed to be proportional to nominal dose rate. Charge multiplication is negligible for the NE2611 operated at 200 V and for Farmer-type chambers operated at up to 400 V. Charge multiplication may not be negligible if the FC is small and operated at a high polarizing voltage. Significant charge multiplication would invalidate the two voltage method used to determine the correction for ion recombination. If there is any doubt, a full saturation curve (Jaffe plot) should be used to determine the range of polarizing voltages from which a linear extrapolation may be made. The resulting factor would only correct for ion recombination. Since charge multiplication is independent of dose rate, and provided the FC is always to be operated at the same polarizing voltage, the correction for charge multiplication would effectively be incorporated into the FC calibration coefficient.

5.6.3. Polarity effect. The polarity effect is more significant for small volume chambers and may be field size-dependent, but is very small for the NE2611 and no correction for polarity is applied during calibration at NPL. Therefore, provided the SS chamber is always operated with the collecting electrode positive with respect to the chamber wall, any polarity effect is incorporated correctly into the SS calibration coefficient. The same approach may be taken with the FC, i.e. never to apply a correction for the polarity effect so that the result of the cross-calibration is to incorporate a correction for the polarity effect under reference conditions.

5.6.4. Leakage. The natural leakage that remains, after correcting electrometer readings for background, must be checked and should be negligible for both SS and FC readings. Any radiation-induced leakage should be taken as an indication that the chamber is faulty.

5.6.5. Other corrections. Assuming that the SS electrometer and chamber have been calibrated independently, the SS electrometer calibration must be applied, together with whatever linearity and range corrections are appropriate for the readings taken. The cross-calibration procedure described here leads to a calibration coefficient for the FC and electrometer regarded as a single instrument. All ionisation readings must be corrected to standard air density using the ideal gas law. Relative humidity should be checked and, provided it is found to be within the range 20%–70%, no further correction is required for either SS or FC readings.

5.7. Field chamber cross-calibration and absolute dose measurement: a worked example with remarks on uncertainty

Calibration by substitution is recommended, so that all measurements are made with the chamber reference point on the beam central axis. Side-by-side calibration is not recommended because of the increased uncertainty from the effects of positioning a chamber within a dose

Table 2. Worked example: FC cross-calibration in an FFF beam.

Step	Measurement condition/ factor	Value	Remarks	Uncertainty	Source of uncertainty
1. Determine QI from FC measurements ('nC')	$d = 20$ cm	13.201	Readings include k_{TP} but not k_{ion}	0.05%	Reading repeatability
	$d = 10$ cm	18.733		0.5 mm 0.5% mm^{-1}	Depth sens. coeff.
	QI ratio	0.705		0.003	Combined in quadrature
2. FC measurements in reference conditions ('nC')	$V = -360$ V, $d = 5$ cm	21.951	Normal voltage	0.05%	Reading repeatability
	$V = -90$ V, $d = 5$ cm	20.938	Reduced voltage	0.05%	
3. Calculate FC k_{ion} corrections and determine $TPR_{20/10}$ and corrected reading ('nC')	k_{ion}, $d = 5$ cm	1.0162	Two voltage formula	0.02%	Error propagation in formula
	k_{ion}, $d = 10$ cm	1.0138	Scaled by dose rate	0.02%	
	k_{ion}, $d = 20$ cm	1.0097		0.02%	
	$TPR_{20/10}$	0.702	Fully corrected	0.003	Combined in quadrature
	R^{FC}	22.307	Fully corrected	0.3%	Includes depth
4. Determine SS dose calibration (cGy nC^{-1})	$N_{D,w}^{SS}$	10.149	Interpolated for corrected $TPR_{20/10}$	0.7%	Standard uncertainty ($k = 1$)
5. SS measurements in reference conditions, includes k_{TP}, k_{elec}, k_{lin} (nC)	$V = -200$ V, $d = 5$ cm	9.6282	Normal voltage	0.05%	Reading repeatability
	$V = -50$ V, $d = 5$ cm	8.8991	Reduced voltage	0.05%	
6. Calculate SS k_{ion}	$d = 5$ cm	1.0273	Two voltage formula	0.02%	Error propagation in formula
7. Uniformity correction for SS and corrected reading (nC)	k_{unif}	1.000	Within uncertainty	0.2%	Estimated from beam profile
	R^{SS}	9.8910	Fully corrected	0.2%	Combined in quadrature
8. FC dose calibration (cGy 'nC'$^{-1}$)	k_{fff}	0.997	Recommend	0.15%	Standard uncertainty ($k = 1$)
	Ratio, R^{SS}/R^{FC}	0.4434		0.36%	
	$N_{D,w}^{FC}$	4.487		0.8%	

gradient when off-axis in an FFF beam (Eaton *et al* 2015). All readings, with the FC and with the SS, are assumed to be taken for repeated delivery of a fixed number of MU, to have been corrected for temperature and pressure, and to include electrometer calibration factors, linearity and range corrections. Readings should be repeated enough times to assure stability and the mean taken of the stable readings. The steps required to carry out the cross-calibration of a field chamber in an FFF beam are explained below and illustrated by a worked example in table 2.

Step 1. Use the FC to determine the user's FFF beam quality index.

Take isocentric FC readings R_{20}^{FC} and R_{10}^{FC}, at depths 20 cm and 10 cm respectively, and obtain the ratio $\frac{R_{20}^{\mathrm{FC}}}{R_{10}^{\mathrm{FC}}}$. After correction for ion recombination at each depth this ratio becomes the quality index $\mathrm{TPR}_{20/10}$

The temperature and pressure corrections tend to cancel in the ratio of corrected chamber readings, leaving their repeatability, precision and the measurement depths as the main sources of uncertainty in the ratio.

Step 2. Take readings with the FC under reference conditions and measure saturation.

Take readings R_A^{FC} and R_B^{FC} at depth 5 cm at the usual operating voltage V_A^{FC} and also at a reduced voltage V_B^{FC}[11]. If the electrometer allows this, the reduced voltage should not be more than 1/3 of the usual operating voltage.

Step 3. Derive and apply the correction for ion recombination to the FC readings.

The correction for saturation for the FC reading at 5 cm depth is $k_{\mathrm{s},5} = 1 + \frac{\frac{R_A^{\mathrm{FC}}}{R_B^{\mathrm{FC}}} - 1}{\frac{V_A}{V_B} - 1}$. The correction at other depths may be measured, or derived from the 5 cm correction. Using the assumed proportionality of ion recombination to dose rate, the correction at depth d cm is given by $k_{\mathrm{s},d}^{\mathrm{FC}} = 1 + (k_{\mathrm{s},5}^{\mathrm{FC}} - 1)\frac{R_d^{\mathrm{FC}}}{R_A^{\mathrm{FC}}}$. Applying this correction for $d = 10$ cm and $d = 20$ cm gives the QI for the FFF beam as $\frac{R_{20}^{\mathrm{FC}} k_{\mathrm{s},20}}{R_{10}^{\mathrm{FC}} k_{\mathrm{s},10}}$.

Step 4. Interpolate the SS calibration coefficient to the measured QI for the user's FFF beam.

Use the cubic fit provided in the NPL calibration certificate to obtain $N_{\mathrm{D,w}}^{\mathrm{SS}}(Q_{\mathrm{cFF}})$ at the value determined in the previous step.

Step 5. Take readings with the SS under reference conditions and measure saturation.

Take readings R_A^{SS} and R_B^{SS} at depth 5 cm at the usual operating voltage V_A^{SS} and also at a reduced voltage V_B^{SS}.

Step 6. Derive and apply the correction for ion recombination to the readings under reference conditions.

The correction for saturation for the SS reading $k_{\mathrm{s}} = 1 + \frac{\frac{R_A^{\mathrm{SS}}}{R_B^{\mathrm{SS}}} - 1}{\frac{V_A^{\mathrm{SS}}}{V_B^{\mathrm{SS}}} - 1}$.

The corrected SS reading is $R_A^{\mathrm{SS}} k_{\mathrm{s}} \equiv R^{\mathrm{SS}}$. The corrected FC reading is $R_A^{\mathrm{FC}} k_{\mathrm{s},5} \equiv R^{\mathrm{FC}}$.

Step 7. Apply corrections for beam non-uniformity.

The SS is calibrated to give absorbed dose at the geometric centre of the chamber in a flat beam. Therefore a correction for beam non-uniformity must be applied when the beam profile is not flat. Pending the availability of better information about the effects, the factor $k_{\mathrm{unif}}^{\mathrm{SS}} = 1.000 \pm 0.004$ (confidence probability 95%) is assumed to correct for volume averaging and fluence perturbation effects on the NE2611 chamber.

The FC readings are not corrected for beam non-uniformity effects: the resulting calibration coefficient takes account of the effects of beam non-uniformity.

Step 8. Obtain the FC calibration coefficient.

Combine the information from steps 4 and 6 with the FFF correction to the SS sensitivity

[11] It is assumed in this worked example that the two voltage method using V_A^{FC} and V_B^{FC} is valid—see discussion in section 5.6.2.

Table 3. Worked example: FC measurement of absorbed dose to water in an FFF beam.

Step	Condition	Result	Remarks	Uncertainty	Source of uncertainty
9. FC reading 'nC'	$d = 5$ cm	21.951	Includes k_{TP}	0.3%	Depth of measurement
	k_{ion}	1.0162		0.02%	Error propagation in formula
	$d = 5$ cm	22.305 'nC'	Fully corrected	0.3%	Combined in quadrature
FC dose calibration (cGy 'nC'$^{-1}$)	$d = 5$ cm	4.487	Determined earlier	0.8%	
Absorbed dose (cGy)	$d = 5$ cm	100.08	FC measurement	0.9%	

$$N_{D,w}^{FC}(Q_{FFF}) = N_{D,w}^{SS}(Q_{cFF})k_{FFF}^{SS}\frac{R^{SS}}{R^{FC}} \tag{9}$$

where $k_{FFF}^{SS} = 0.997 \pm 0.003$ (confidence probability 95%).

Assuming that the same thermometer and barometer are used to obtain air density corrections for the FC and SS readings, most sources of uncertainty in temperature and pressure tend to cancel in the ratio of corrected chamber readings.

If the cross-calibration is performed in a solid phantom[12], the effects of any water-inequivalence of the medium surrounding the chamber must be assessed. For example, the mass density of PMMA is almost 20% greater than that of water and would be expected to lead, via depth-dose gradient effects, to a chamber over-response. However, provided the chamber diameters are similar, the over-responses will also be similar. For example if an NE2571 chamber is calibrated against an NE2611 chamber using a PMMA phantom, the resulting error in calibration is estimated to be smaller than 0.1%. The additional steps required to measure absorbed dose to water in an FFF beam using a newly calibrated field chamber are explained below and illustrated by continuing the worked example in table 3.

Step 9. Use the FC to measure absorbed dose to water.

Absorbed dose to water under reference conditions is given by

$$D_w(Q_{FFF}) = N_{D,w}^{FC}(Q_{FFF})R^{FC}(Q_{FFF}) \tag{10}$$

where the FC reading is corrected for air density and ion recombination but not for beam uniformity. The uncertainty of this measurement receives contributions from all sources of uncertainty in the temperature and pressure measurements used to correct the FC reading to standard air density.

Step 10. An analysis of uncertainty.

Tables 2 and 3 assemble the steps in a cross-calibration and subsequent measurement of absorbed dose, including estimates of uncertainty. Of course, the user must replace many of the uncertainties listed in the table by estimates that apply to their own measurements.

[12] The PMMA phantom commonly used for cross-calibration in cFF beams may also be used in FFF beams but, as emphasised above, not in its side-by-side configuration. All measurements must be made with one chamber position set up on the beam central axis while a blanking rod is used in the other chamber position.

5.8. The use of alanine/EPR as an audit dosimeter, and for traceability

The NPL offers a mailed alanine dosimetry service that can be used to measure absorbed dose in an FFF beam. The preferred route to obtain a calibration factor for the use of a FC in an FFF beam is via the calibration of a SS chamber in a cFF beam as just described in this section. However, alanine dosimetry can also be used to derive a FC calibration, or to make a measurement of absorbed dose under non-reference conditions. For the calibration, sequential readings should be taken with the alanine and with the FC for repeated irradiations under reference conditions. The irradiated alanine dosimeters are mailed back to NPL, where the alanine dosimeters are read out to determine the dose delivered. This absorbed dose is used in equation (2) to obtain the calibration coefficient for the FC

$$N_{\mathrm{D,w}}^{\mathrm{FC}}(Q_{\mathrm{FFF}}) = \frac{D_{\mathrm{w}}^{\mathrm{alanine}}(Q_{\mathrm{FFF}})}{R^{\mathrm{FC}}(Q_{\mathrm{FFF}})}. \tag{11}$$

6. Verification

Most commercially available equipment for dosimetry verification is also suitable for use in FFF beams. This should be checked in the manufacturers' specifications.

6.1. Ionisation chambers

Chambers of 0.2 cm^3 or smaller will be more suitable for use in the high dose gradients of FFF beams. Larger chambers should be avoided. Refer to sections 4.1.1 and 5—Reference Dosimetry for a discussion of chamber choice. Chambers should be cross-calibrated at depths similar to the depth of verification measurements to avoid introducing differences from ion recombination variation with depth.

6.2. Daily check devices

Daily check devices may saturate at high dose rates so must be checked for range of suitable dose rates before use. If the device saturates at the clinical dose rate it is still possible to use it routinely at a lower, more suitable dose rate. A routine QC check should then be added to the QA programme which ensures that the dose delivered at the clinical dose rate remains the same as that delivered with the lower dose rate used with the daily check device. Alternatively, additional build-up material could be used to reduce the dose rate at the device to below the saturation level.

6.3. Detector arrays

Detector arrays currently on the market are suitable for use in FFF beams. With diode arrays, the dose rate dependence of modern diodes is minimal (see, for example, manufacturers' specification data) so array calibration in an FFF beam is not a problem. Arrays calibrated by the manufacturer have also proved suitable for use in FFF beams. As with daily check devices, saturation at high dose rates is possible with arrays so the range of suitable dose rates should be checked before use, either physically or by reference to the manufacturer's literature.

6.4. Electronic portal imaging devices

Most electronic portal imaging devices (EPIDs) saturate at high dose rates and so are unsuitable for direct FFF beam verification. However, the dose rates for exit dosimetry will be low enough to avoid saturation with these EPIDs. Elekta and Varian have recently developed solutions to this saturation problem, which are now commercially available. Refer to manufacturers' information to determine the suitability of your EPIDs for FFF beam verification.

6.5. Film

Film is expected to be usable in FFF beams both as a relative and absolute dosimeter. Energy independent radiochromic films should be suitable for use in FFF beams.

6.6. Diodes

FFF beams are mostly used for IMRT and VMAT with inherent gradients where *in vivo* dosimetry with diodes is contraindicated.

For unmodulated beams, due to the steep dose gradients in FFF beams, tiny positioning errors cause large dose errors, so the accuracy of diode measurements is compromised. It is recommended therefore to perform *in vivo* dosimetry in unmodulated beams only on the central axis of the beam where the beam profile is flattest (and thus more forgiving of positional errors) and the calibration of the diode is performed.

6.7. TLDs and MOSFETs

TLDs and MOSFETs are both expected to work in FFF beams as long as they have been calibrated appropriately. Neither of these detectors has been formally studied in FFF beams yet. TLD batch sensitivity spread is likely to be higher in FFF beams (~10% difference (Cashmore 2015)). As with diodes, *in vivo* dosimetry with these detectors should be performed on the central axis of the beam.

6.8. Monitor unit check software

Monitor Unit check programs currently on the market require separate models for FFF beams to be created (refer to manufacturers' information). Some MU check software algorithms model FFF beams as if they were flattened beams. Thus no account is taken of the spectrum being constant across the beam and scattered radiation is modelled as if there were a flattening filter. Checks should be made to ensure the accuracy of the calculations is acceptable, whichever algorithm is implemented in the software.

7. Quality control

In general the QC program for FFF beams is expected to be similar to that of conventional beams, the main differences being caused by the change in profile shape. Tolerances and frequencies are not expected to change significantly from conventional QC; however this needs to be established from experience with the relevant data now beginning to be published (Clivio *et al* 2014).

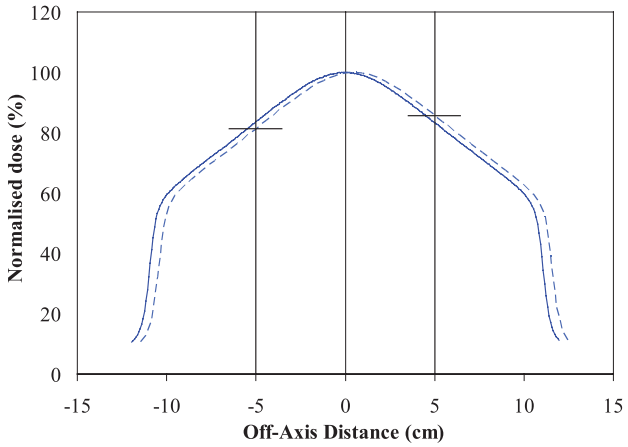

Figure 10. Showing how beam steering variation causes a shift in the FFF beam rather than the beam tipping. The lines at ±50 mm show how the symmetry parameter is still relevant—the shifted beam profile (dashed) gives decreased signal at −50 mm and increased signal at +50 mm, resulting in an increase in the symmetry parameter.

7.1. Equipment

Most commercial dosimetry equipment on the market is already suitable for FFF. But before use, the dose rate specification should be checked with the manufacturer, especially for older equipment. Certain types of equipment are likely to be more susceptible to high dose rates; in particular liquid filled ion chambers due to the high signal generated. These should be carefully checked before use. QC using EPIDs may also encounter problems due to high dose rate saturation.

Equipment may need to be used under different operating conditions. Examples include:

- changing gain settings to prevent saturation;
- calibrating the device in the FFF beam;
- PTW provide an additional device to 'flatten' the beam for use with their daily output device.

7.2. Beam profile checks—flatness and symmetry

Without the FF, beam steering variations lead to positional shifts in the beam profile (see figure 10) rather than asymmetries (which are a result of mismatch between unflattened beam and FF shape). A number of new parameters have been proposed in the literature to characterise FFF beams along with suggested tolerances (Fogliata *et al* 2012, Clivio *et al* 2014). However, the traditional parameters of flatness and symmetry still have validity but have a slightly different interpretation.

- Symmetry—now relates to how far a beam is offset rather the degree to which a profile is 'tipped'—this is illustrated in figure 10. Care should be taken with equipment set-up since a set-up error would lead to an apparent symmetry change. Symmetry values should remain within 2%. Measuring central axis deviation (peak position) has also been proposed—which is easy to measure with current devices.

- Flatness now has little relevance to the profile shape but still works as a consistency check related to beam energy. However, if using a low resolution device the shape of the profile can lead to a 'quantisation' of results, depending on which detectors are included in the analysis. The flatness value will be even more highly dependent on field size and normalisation and algorithm implementation (software algorithms may be limited as to how they normalise the profile and therefore how flattened area is defined).

7.3. Energy checks

Conventional methods for energy checks such as $TPR_{20/10}$ or other simple ratio methods are just as applicable to FFF beams and can be used in exactly the same way (see section 5.4).

7.4. Field size checks

Refer to section 4.2 for advice on field size measurement and definition for FFF beams.

8. Clinical usage

8.1. Introduction

Historically, flat, uniform fields have been used almost exclusively for treatment planning, but modern radiotherapy now relies more on the use of small field hypofractionated and/or fluence modulated techniques. FFF is therefore ideal for stereotactic applications (SRS/SBRT) and IMRT/VMAT, although conventional planning has been shown to be possible for FFF (Cashmore 2007, Stevens *et al* 2011, Kretschmer *et al* 2013).

Care must be taken when evaluating comparative plan studies between cFF and FFF techniques as plan differences attributed to filter removal may simply be due to a change in beam energy e.g. 4 MV versus 6 MV for Varian.

8.2. SRS/SBRT

The flattening filter has very little effect on beam profiles for small fields (i.e. $<5\,cm^2$), and for these beams the filter is essentially an unnecessary component. As the field shape is almost unaffected by filter removal, SRS/SBRT plans planned using conformal, IMRT or VMAT techniques can be expected to remain relatively unchanged.

VMAT should ideally be used for SBRT in order to deliver the radiation in the shortest time period for these high dose-per-fraction treatments. Treatment times can be reduced by up to 50% for 6 MV FFF delivery with very little change in dose to target or organ at risk (OAR) (Stieler *et al* 2013, Abacioglu *et al* 2014, Dzierma *et al* 2014, Wen *et al* 2015). The relatively low modulation and high doses required for SBRT treatments means the advantage of the increased dose rate of FFF is not lost as it can be for low dose per fraction VMAT deliveries when the dose rate is reduced due to limits on leaf and gantry speed. Most studies have therefore focussed on aspects of delivery time and motion management whilst demonstrating equal plan quality.

Higher dose rates can aid motion management issues since the tumour/patient has less time to move during treatment, but could also present additional problems due to the interplay effect. Plans delivered at higher dose rates can be expected to cover fewer breathing cycles, reducing the averaging effect of longer deliveries.

Whilst the dosimetric effects of interplay are generally small at low dose rates (Ong *et al* 2013), they can be significant at those seen for FFF. Ong *et al* (2013) reported that single arc and single fraction SBRT lung treatments at 2400 MU min^{-1} were susceptible to interplay effects (dose differences of 5–10%), but the use of 2 arcs or $\geqslant$2 fractions reduced those effects to acceptable levels.

It has been concluded (Vassiliev *et al* 2009, Ong *et al* 2011, 2013, Mancosu *et al* 2012, Rao *et al* 2012, Reggiori *et al* 2012, Chung *et al* 2015) that equivalent plan quality can be achieved whilst reducing treatment delivery times despite the number of MU for FFF being higher than cFF. The significance of the reduction in treatment delivery time increases with the dose per fraction. One author has concluded that with the use of FFF beams it is now possible to deliver breath hold techniques in a standard cFF beam time-slot (Boda-Heggemann *et al* 2013). Treatment times may also be shortened by the reduced use of imaging. Prendergast *et al* (2013) compared FFF versus cFF delivery, for lung and liver SBRT, considering treatment time and the frequency and type of intra-fraction image guidance used. They reported that not only was the treatment time reduced for FFF but that the use of FFF was associated with a reduction in intra-fraction CT imaging which contributes to further improvement in treatment delivery efficiency.

8.3. IMRT/VMAT planning

With larger field sizes the effect of the FFF beam profile becomes more pronounced, so as the volume of the tumour (treatment length) and complexity of the plan increase so do the number of MU required to deliver these plans; this is a natural consequence of the FFF beam profile since it will require longer beam on times to deliver doses off-axis. The complexity of the plan is also seen to increase the number of segments required to treat the FFF plans compared to the conventional IMRT plans if plan quality is to be maintained.

Many studies have been undertaken to demonstrate plan equivalence between standard and FFF beams across a number of treatment sites (Nicolini *et al* 2012, Lechner *et al* 2013, Gasic *et al* 2014, Zhuang *et al* 2015). MU were increased for all FFF plans as expected, with the number of MU increasing with PTV volume. Beam on times were reduced for high dose per fraction SRS and SBRT techniques but unchanged for others. Very small differences in plan quality have been observed between techniques.

The combination of FFF and deep-inspiration breath-hold beams to treat left-sided breast patients was studied by Koivumäki *et al* (2015). VMAT, dynamic IMRT and field-in-field planning techniques showed little difference in target coverage or OAR sparing, but a reduction in beam on time of 18–39% for filter free was seen.

The way in which plans are generated can have a significant effect on the outcomes of treatment planning studies, and care should be taken to understand the optimisation procedure used and the degree to which the optimisation of the plans has been pushed to increase target coverage or reduce OAR doses. Most studies utilise clinical treatment plans generated for cFF beams and re-optimise for FFF using the same constraints. It is not yet confirmed using the same parameters is justified, or whether different parameters should be used to drive FFF optimisation.

8.4. Peripheral doses

FFF techniques can be successfully used to reduce treatment times for high dose-per-fraction treatments. For standard fractionation schedules however, the delivery of modulated treatments are generally limited by MLC and gantry speed, so the highest dose rates are rarely

achieved. Delivery times therefore remain similar to conventional techniques and the question is raised of whether FFF has any benefit for these patients.

Treatment times are not the only consideration for patient treatment, however, and the reduced scatter and leakage radiation associated with filter-free delivery may still favour FFF delivery, particularly for paediatric or pregnant patients.

The amount of leakage radiation from the treatment head is proportional the number of MU used, and will be raised for IMRT techniques compared to 3DCRT as they are inherently wasteful of monitor units (MU). Several authors have compared the delivery of 3D conformal plans against IMRT and reported that IMRT delivery may double the incidence of solid cancers in long-term survivors (Verellen and Vanhavere 1999, Hall and Wuu 2003, Kry *et al* 2005, Hall 2006, Ruben *et al* 2008).

Several authors have investigated peripheral doses in the FFF setting (Vassiliev *et al* 2006b, Kry *et al* 2010, Cashmore *et al* 2011, Kragl *et al* 2011b, Spruijt *et al* 2013, Murray *et al* 2015). These studies show FFF resulted in lower risks at all sites compared to equivalent flattened techniques, despite an increase in the number of MU required for FFF plans, with increasing impact at greater distances from the field. However, in absolute terms the risks remain low. In a Monte Carlo study by Kry *et al* (2010) however, unflattened beams were seen to increase the dose in the region from 3–15 cm from the field edge for IMRT delivery on a Varian accelerator. These differences may be due to variations in head design between Elekta and Varian machines, or due the Varian accelerator having a slightly lower effective energy (4 MV). Further investigations to quantify any potential benefits are warranted.

9. Conclusions and summary

FFF radiotherapy should not be thought of as a new treatment modality but as a minor variation from conventional photon radiotherapy. Most of the approaches used in conventional photon radiotherapy can be readily applied to FFF, but with care taken in the specific areas highlighted in this report where the higher dose rate and shape of FFF beams requires small modifications to existing techniques. The radiotherapy community needs to quickly learn about these variations since the usage of FFF beams will continue to increase, with some linacs used exclusively in FFF mode. The key points of this report are summarised as follows.

(1) The flattening filter is unnecessary for small field and IMRT techniques. Removal increases dose rate and dose per pulse, and reduces scatter, leakage and off-axis variations in beam energy. Varian and Elekta implementations of FFF are fundamentally different and should be well understood: Elekta linacs energy-match the beam, Varian linacs have a lower effective energy.
(2) Shielding of existing linear accelerator bunkers is likely to be adequate for FFF beams, but a full assessment of expected workload and type of treatment should be carried out within a risk assessment prior to commissioning. Time averaged dose rates should be used rather than instantaneous dose rates in radiation protection calculations.
(3) Beam characteristics, whilst different from flattened beams, can still be described with the same nomenclature, and data can be obtained in the same manner for commissioning provided small detectors are used and the increased ion recombination is accounted for.
(4) FFF beams must be correctly renormalized to enable field size and penumbra to be determined; a new approach is recommended of dividing FFF profiles by profiles of a larger size to create a virtually flattened beam profile to which conventional field size and penumbra definitions can be applied.

(5) Reference Dosimetry for FFF beams follows a similar format as for cFF beams but with a preliminary quality-dependent correction of 0.997 recommended for clinical FFF beams. Careful attention must be paid to ion recombination corrections, which are higher than for cFF beams. Inter-comparison measurements should be sequential rather than side by side.
(6) Current quality control regimes and verification can easily be extended to include FFF modes, using existing equipment under careful 'fit for use' testing in FFF beams. Saturation of measuring devices could present a problem, but this can be circumvented by judicious choice of measurement conditions.
(7) Higher dose rates and the shape of the FFF beam lend themselves well to clinical implementation in SRS/SBRT treatments, where high doses/MU can be delivered in short times. Although minor variations are seen, treatment plans for cFF and FFF are of comparable quality. Despite increased MU for FFF beams, peripheral doses are generally lower. FFF may be the mode of choice if peripheral dose sparing is necessary, e.g. in paediatric or pregnant patients.
(8) Care should be taken when reviewing planning studies to ensure the evaluation and plan optimisation assessments are comparable. FFF beams are used across a large number of sites and there is a large amount of published literature available. Special consideration should be taken when reading and comparing literature as to the manufacturer used as this can result in considerably different beam energies.

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