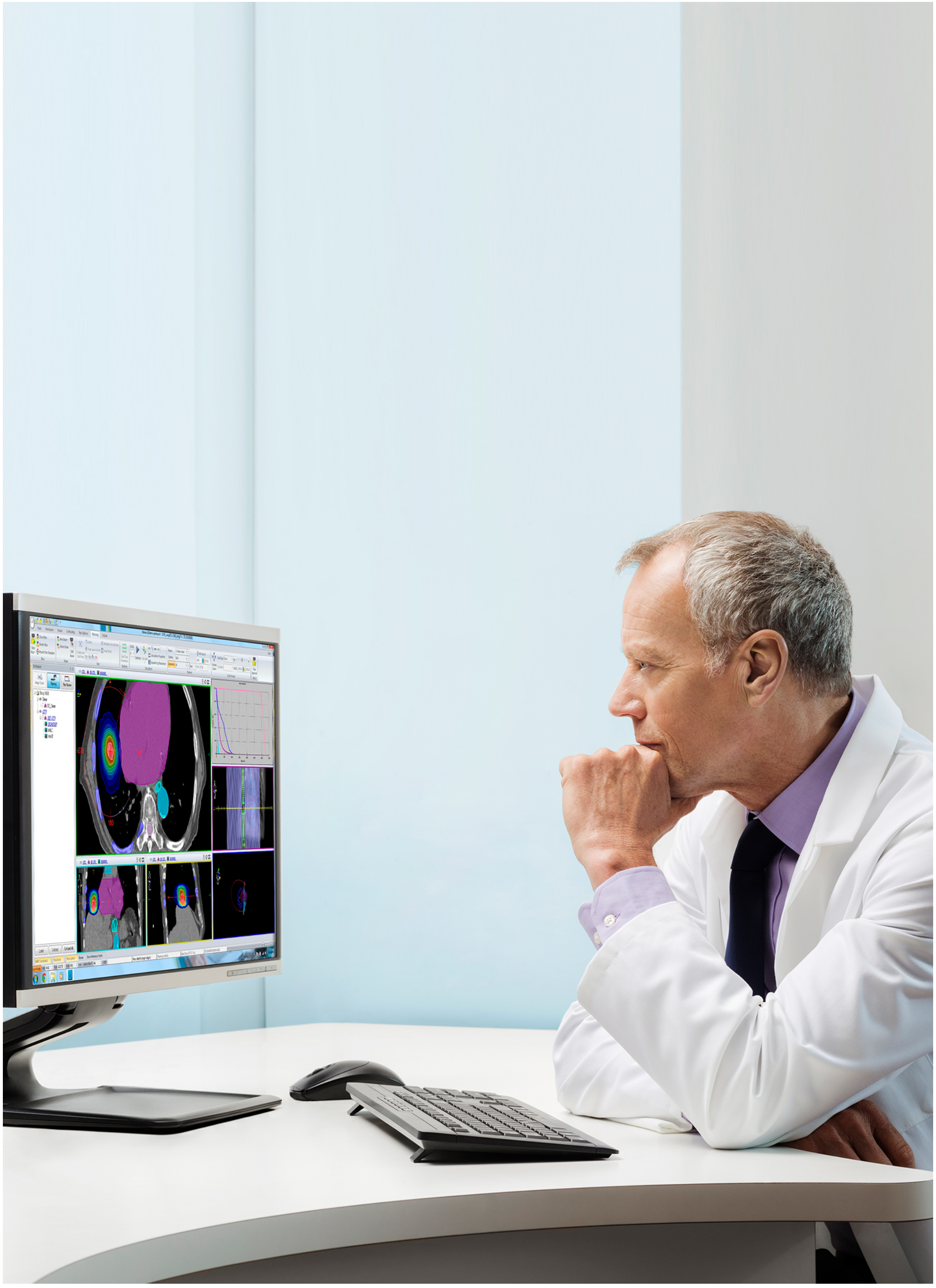
Technical Publication

Monaco

Technical Reference for: The IQM and Monaco Beam Models



Document ID: E062830/01
Publication date: 2022-11-25
Language: English

Referenced documents

Elekta does not supply all the documents that we refer to in this document with the equipment. Elekta reserves the right to make the decision on which of the documents we supply with the equipment.

CE Marking (Compliance with EU MDR)

Elekta maintains a Quality System that complies with all essential requirements of the European Medical Device Regulation EU MDR (2017/745) and all other standards applicable there under. A Technical File is maintained for review via Elekta directly.

Monaco	
CE 2797	

Contact information

MANUFACTURER **Elekta Solutions AB** Kungstensgatan 18, Box 7593 SE-103 93 Stockholm Sweden Phone: +46 8 587 254 00 Fax: +46 8 587 255 00 eu_ar@elekta.com	**Australia Sponsor Information** Elekta Pty Ltd is the sponsor of this device. All inquiries in relation to this product to be made to the following: Sales.Ausnz@elekta.com Elekta Pty Ltd Suite 10.02, Level 10 16 Arthur Street North Sydney NSW 2060 +61 (0) 2 8907 1800 A.B.N. 49 109 006 966
CH REP Elekta GmbH Obermühle 8 6340 Baar Switzerland	

TECHNICAL SUPPORT
Contact your local Elekta representative for technical support or visit www.elekta.com

Table of Contents

1 Introduction

1.1 Introduction

The Integral Quality Monitor (IQM) is a large-area ionization chamber used for quality assurance verification measurements of treatment delivery accuracy (beam energy, position, and shape) from medical linear accelerators. The readout of the IQM is used to compare the delivered treatment to the expected treatment and to compile data over time as part of a comprehensive clinical quality assurance program. The IQM is a product of iRT Systems GmbH (Koblenz, Germany).

1139_MNCO

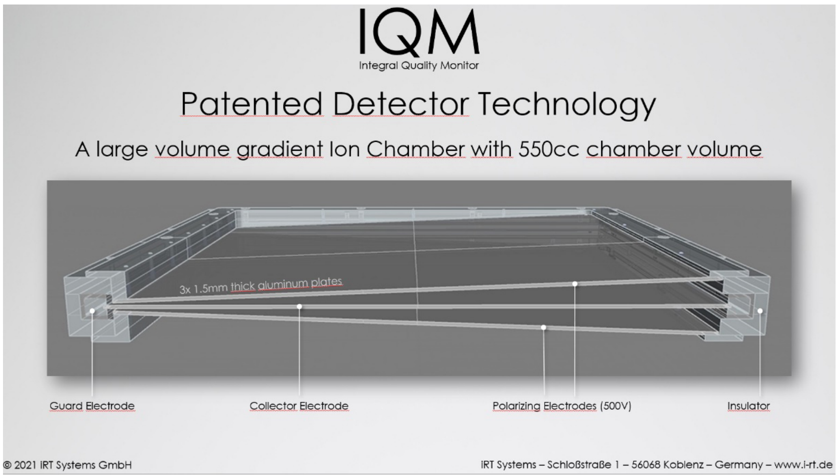
This document discusses the representation of the IQM in Monaco beam models for the photon Monte Carlo and Collapsed Cone Convolution dose algorithms. It compares dosimetric data measured with an IQM to similar data measured without an IQM and discusses how Monaco models can be adjusted to account for the changes in the dose distribution. The document concludes with a set of recommendations.

Monaco beam models are created by a dedicated team at Elekta. This beam modeling service saves the customer many hours of work and ensures that all models are created and verified by persons with appropriate experience and expertise. This document discusses some of the beam model parameters for informational purposes, but it is important to understand that Monaco users should never alter any of these parameters on their own [1].

2 How the IQM affects the linac photon beam

2.1 Physical Effects of the IQM

The heart of the IQM is a single large ionization chamber with three aluminum plates that serve as electrodes. The plates are all 1.5 mm thick and nearly perpendicular to the beam axis, with the upper and lower plates tilted slightly to create two wedge-shaped air cavities.



1140_MNCO

Radiologically, the device is thus nearly equivalent to a 4.5 mm thick slab of aluminum, or to a blocking tray of slightly above-average thickness.

In the discussion that follows, the variable Z will represent distance along the beam axis from the nominal source point of the accelerator. The IQM is 3.5 cm thick [2]. When mounted on an Elekta Versa HD linac, its distal surface is at about Z = 59 cm. The vertical location of the aluminum plates is therefore between 55.5 and 59 cm.

A radiotherapy photon beam has three dosimetrically significant components, each of which the IQM affects in a different way:

1 ***Primary photons*** are defined as photons that are not scattered after their initial creation in the photon target near $Z = 0$. Most of the dose delivered to the patient comes from such unscattered photons. The IQM *attenuates* these photons, mostly by means of Compton interactions that give rise to lower-energy scattered photons. The attenuation coefficient of aluminum decreases with energy in the radiotherapeutic range, so photons of lower energy are attenuated more than photons of higher energy. The action of the IQM on the primary photons thus has two observable dosimetric effects:

   a. a decrease in overall beam output;

   b. a "hardening" of the energy spectrum, meaning a shift to higher energies of the probability density function representing the spectrum.

2 ***Scattered photons*** are photons that result from interactions in the accelerator head below the photon target and above the patient surface. Most of these photons are scattered by Compton interactions, but this beam component also includes annihilation photons resulting from pair production and bremsstrahlung photons created below the target by secondary electrons. The most important sources of scattered photons are the flattening filter (if present) and the primary collimator, both of which are located in the range $Z = 1.6$ to 16 cm, below the photon target and above the MLC. Scattered photons incident on the IQM are attenuated and scattered just like primary photons, but such doubly or multiply scattered photons are a correction to a correction and not dosimetrically significant. What matters is that the IQM scattering of the primary photons (see above) turns the IQM into an *additional* source of scattered photons. This new source of scattered photons at around Z=57 to 58 cm will result in a broadening of the penumbra and an increase in dose outside the field, especially for large fields.

3 ***Contaminant electrons*** are generated in the treatment head when photons undergo Compton scattering or pair production. For electrons, as for photons, the IQM is both a modifier of incident particles and a source of new particles. Incident electrons with kinetic energies up to about 2 MeV will be completely blocked by the 4.5 mm of aluminum in the device. The rare incident electron of higher energy will typically lose more than 2 MeV of its kinetic energy in transit and will undergo significant angular scattering. The IQM thus obscures upstream electron sources and becomes itself the main source of contaminant electrons reaching the patient. In contrast to a beam with no distal QA device, an IQM beam generates electrons much closer to the patient surface, and with no intervening collimation. The result is a significant increase in dose at shallow depths both inside and outside the field. The effect is much stronger for large fields than for small fields, because the number of new contaminant electrons is proportional to the area of the IQM surface that is irradiated by the primary photons.

To review, then, we can expect to see four distinct dosimetric effects due to the presence of the IQM at the distal end of a photon beam:

A. Reduction of output, due mostly to attenuation of primary photons.

B. Spectral hardening, due to relatively greater attenuation of lower-energy primary photons.

C. Increased out-of-field dose, due to an additional source of scattered photons much closer to the patient surface.

D. Increased surface dose, due to a new source of uncollimated electrons relatively close to the patient surface.

The decrease in output (A) is by far the most important effect, both dosimetrically and clinically. The increase in surface dose (D) is the next largest effect, but may be clinically insignificant in cases where field sizes are small or surface dose is not a major concern. Effects B and C are less pronounced, but nonetheless clearly visible in measured data. All four effects will be illustrated below from the data of a Monaco customer site that uses an IQM with an Elekta Versa HD linac.

2.2 Reduction of output

The following table shows cGy/MU for a 10x10 cm field at 90 cm source-to-surface distance (SSD) and 10 cm depth for four Elekta photon beams with and without the IQM.

Table 2.1

Nominal Accelerating Potential and Fluence Mode	Without IQM (cGy/MU)	With IQM (cGy/MU)	Transmission Factor
6 MV, flattened	0.810	0.755	0.932
6 MV, flattening filter free (FFF)	0.800	0.743	0.929
10 MV, flattened	0.873	0.824	0.944
15 MV, flattened	0.914	0.868	0.950

The reduction in beam output for this linac (effect A) ranges from about 7% for the two 6 MV beams to about 5% for the 15 MV beam. Similar numbers have been reported in the literature:

- The IQM specifications web page [2] gives approximate transmission factors of 0.945 for 6 MV and 0.955 for 18 MV.
- Miori et al. [3] measured output reductions of 6.56% for a 6 MV beam and 5.27% for a 10 MV beam.
- Hoffman et al. [4] measured output reductions of 5.43% for a 6 MV beam, 4.60% for a 10 MV beam, and 4.21% for a 15 MV beam.
- Nguyen et al. [5] measured transmission factors of 0.926–0.933 for a 4 MV beam, 0.937–0.941 for a 6 MV beam, 0.937–0.939 for a 6 MV FFF beam, and 0.949–0.953 for a 10 MV beam.
- Nine participating cancer centers collaborated in 2017 on a multi-institutional study of IQM dosimetry [6]. The transmission factors stated in the paper abstract are 0.9412 for 6 MV, 0.9440 for 6 MV FFF, 0.9533 for 10 MV FFF, and 0.9578 for 18 MV.

The reduction of output is almost independent of field size, but not entirely. The following table shows output factors (relative to the 10x10 cm field) for the flattened 6 MV beam referenced in Table 2.1.

Table 2.2

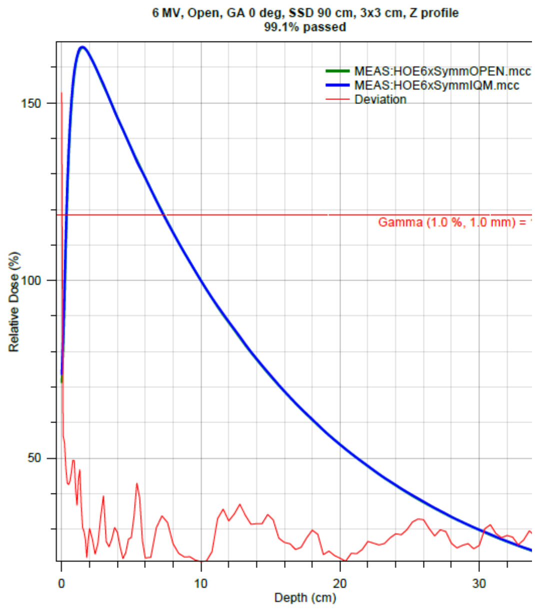
Field Size (cm)	Without IQM	With IQM	Ratio
1x1	0.698	0.698	1.000
2x2	0.806	0.806	1.000
3x3	0.847	0.848	1.001
4x4	0.879	0.878	0.999
5x5	0.906	0.905	0.999
7x7	0.950	0.949	0.999
10x10	1.000	1.000	1.000
15x15	1.057	1.058	1.001
20x20	1.096	1.099	1.003

Field Size (cm)	Without IQM	With IQM	Ratio
30x30	1.144	1.153	1.008
40x40	1.165	1.179	1.012

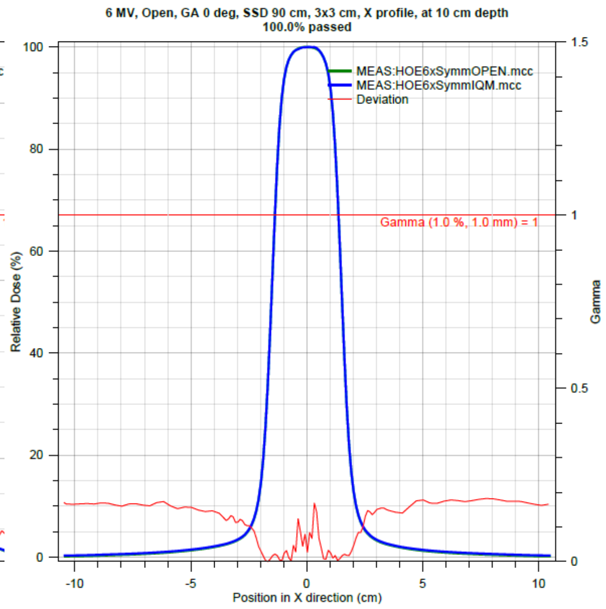
Up to a field size of 20x20 cm, the IQM alters the output factors by less than half a percent. For the largest fields, however, the output factor with the IQM is as much as 1.2% higher than without the device. The divergence arises because photons scattered by the IQM in a small field generally land outside the field boundaries, whereas photons scattered in a large field may land inside the boundaries, including on the central axis where the detector is located.

2.3 Effects that alter the shape of the dose distribution

The other three effects that were mentioned at the end of Section 2.1 will be illustrated using data for the 6 MV beam of the aforementioned Elekta customer site. Similar effects can be observed for other energies and other linacs. The first thing to note is that effects B, C, and D tend to be negligible for small fields. The following side-by-side images from the Monaco commissioning utility (MCU) show depth dose curves and lateral *X*-profiles at a depth of 10 cm for a 3x3 cm field.



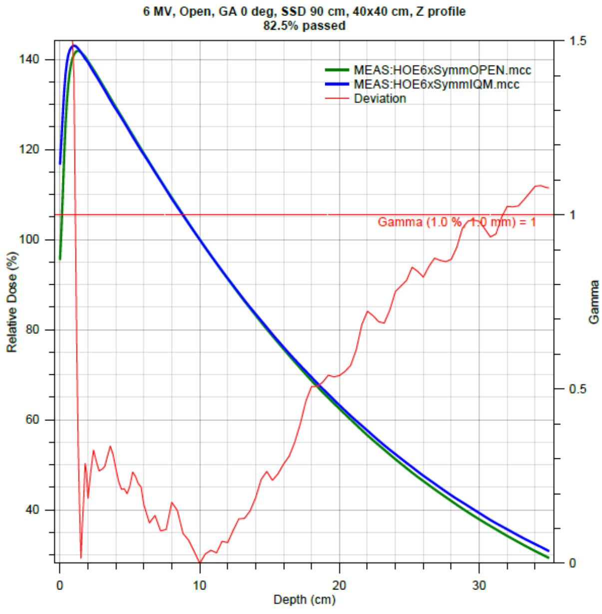
1141_MNCO



1142_MNCO

The blue lines represent the dose distribution with the IQM in place. A single renormalization factor (analogous to a tray factor) has been applied to the entire dose distribution, lateral profiles as well as depth dose. No other adjustments have been made. The green lines represent the dose distribution without the IQM in place. They are, in fact, not visible, because the shapes of the two dose distributions are practically identical. The same near-perfect agreement holds at all measured depths and for the *Y*-profiles as well as the *X*-profiles.

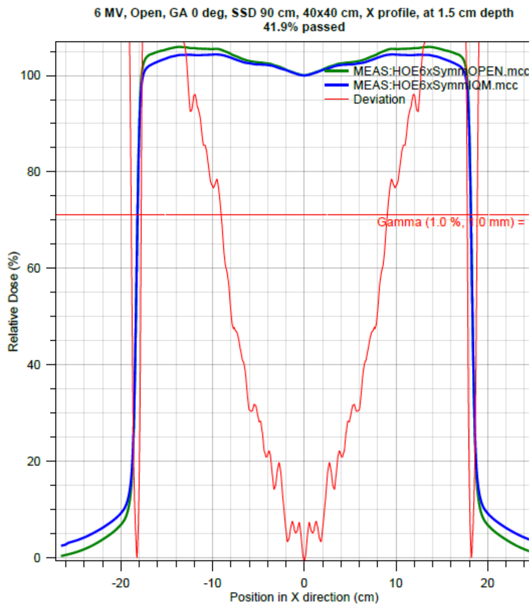
It is for large fields that the IQM begins to change the *shape* of the dose distribution. The following image shows depth dose curves for the same beam and a 40x40 cm field.



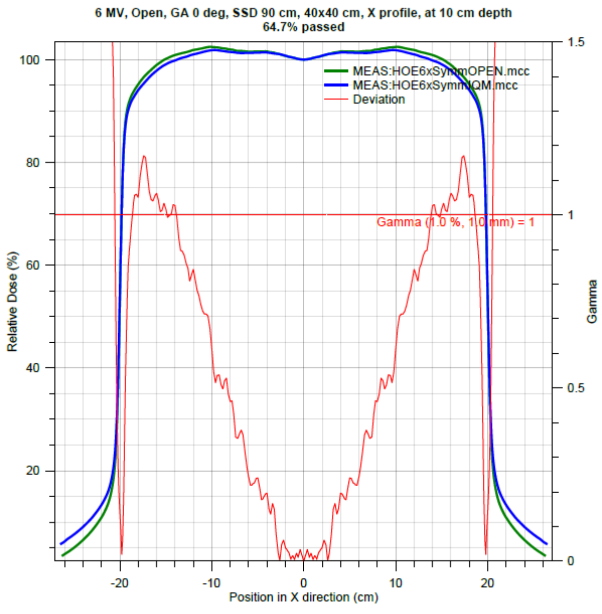
1143_MNCO

The increase in surface dose (effect D) is clearly visible at the shallow end of the plot, while the hardening of the spectrum (effect B) is clearly visible at the deep end.

The following side-by-side images show *X*-profiles of the same 40x40 cm field at depths of 1.5 cm and 10 cm.



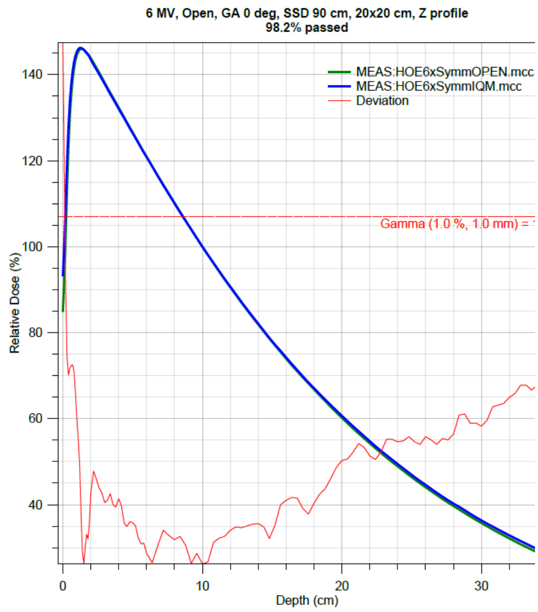
1144_MNCO

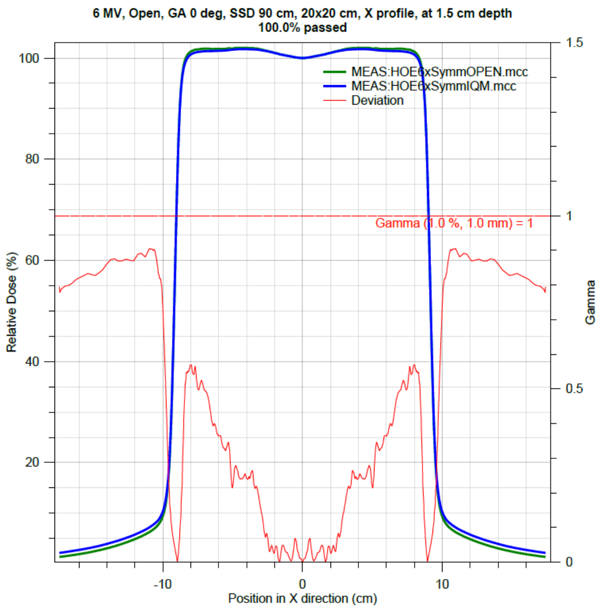


1145_MNCO

Note the increased dose outside the field at both depths. Electrons contribute something to the excess out-of-field dose at 1.5 cm, but the persistence of the excess dose to a depth of 10 cm shows that it is mostly due to scattered photons (effect C). We also observe here that the “shoulders” or “horns” of the flattened beam are lowered by the presence of the IQM. This alteration is further evidence of spectrum hardening (effect B).

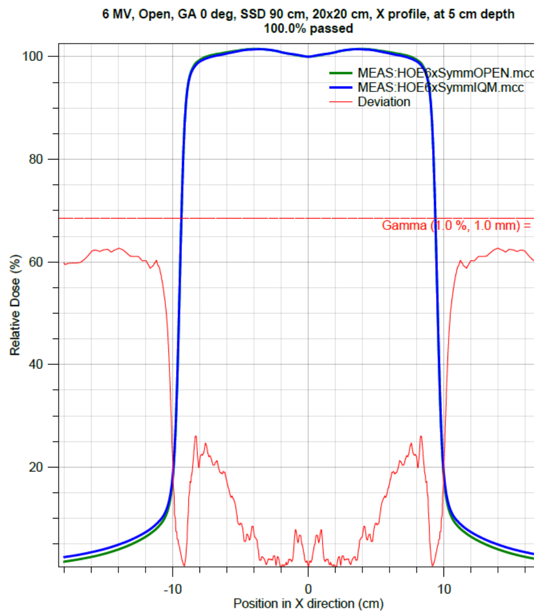
As a general guideline, we can take 20x20 cm to be the field size threshold at which dosimetric effects B through D cease to be negligible. For the same 6 MV beam as above, the following images show depth dose curves and *X*-profiles at three depths for a 20x20 cm field.

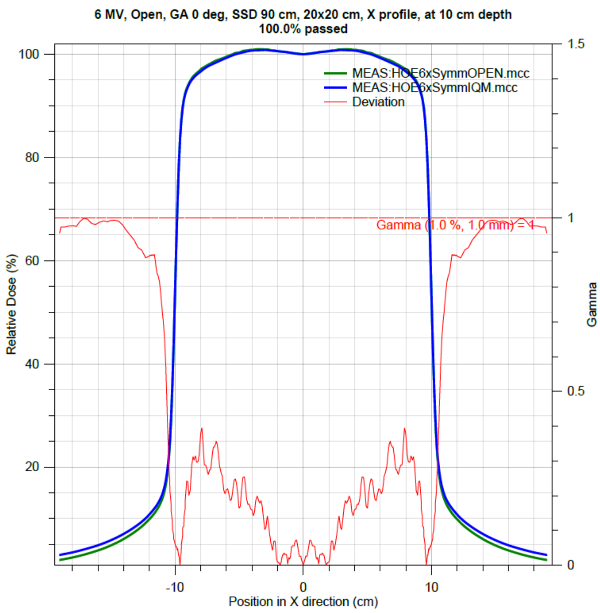




1146_MNCO

1147_MNCO





1148_MNCO

1149_MNCO

The spectrum hardening and surface dose increase are visible in the depth dose plot, but are not clinically significant. The increased out-of-field dose is visible in the lateral profiles at all depths, but the with-IQM dose still matches the no-IQM dose, just barely, to within 1%/1 mm gamma criteria.

3 Modeling the effects of the IQM

3.1 Reduction of Output

As discussed in the preceding section, the predominant effect of the IQM is a reduction of output due to attenuation of the primary photons. The reduction in output can be greater than five percent, and thus misadministrations can occur if the effect is not taken into account. It is critically important that clinics validate the IQM with their own linacs and put procedures in place to ensure that the IQM is attached to the linac if and only if it was incorporated into the patient treatment plan [7].

Fortunately, a reduction in output is easy to model. It can be done by introducing a multiplicative renormalization factor that applies to the entire dose distribution for all field sizes of the given beam energy. This once-for-all “IQM factor” is analogous to the blocking tray factor that has a long history in photon radiotherapy planning. The clinical consensus is that the IQM can be used safely in most cases with *only* this adjustment for the overall change in output:

- The IQM specifications web page [2] says that the “effect on beam quality” and the “effect on beam spectrum” are both “negligible”. For “integration of beam attenuation in TPS”, it suggests “tray factor or modified beam model”.
- Similarly, the aforementioned nine-center IQM study [6] concludes, “The magnitudes of changes which were found justify treating IQM either as tray factors within the treatment planning system (TPS) for a particular energy or alternatively as modified outputs for specific beam energy of linear accelerators, which eases the introduction of the IQM into clinical practice.”
- A 2016 conference presentation from the Christie Hospital NHS Trust [8] concludes, “Clinical models [i.e., models modified only by a renormalization factor] acceptable at 6 MV and 10 MV... Simple correction factor to change MU’s was all that was required.”

This simple renormalization approach works very well for small- and medium-size fields, but it neglects the field size dependence of the output factor for very large fields. It also neglects the effects that change the shape of the dose distribution, as will be discussed in the next section.

3.2 Effects that alter the shape of the dose distribution

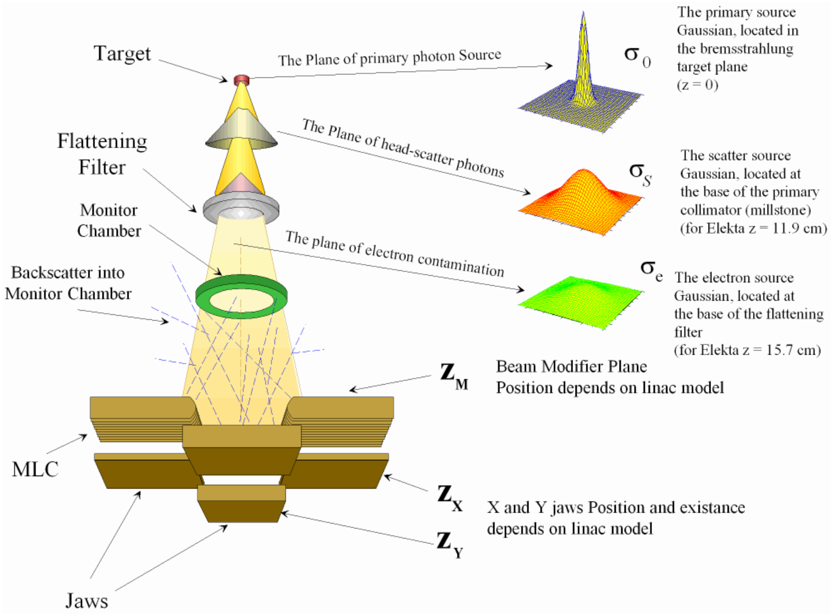
The dosimetric accuracy obtained by simply renormalizing the beam may not be satisfactory to some clinicians under some circumstances, particularly where large fields are involved or dose at shallow depths is clinically important. The remainder of this document will discuss what can be done to model the lesser dosimetric effects that were discussed above: spectral hardening, increased out-of-field dose, and increased surface dose.

The long-term solution is to update existing dose algorithms so that they model the IQM in a more precise way: as an attenuator of primary photons, and as a secondary source that emits scattered photons and contaminant electrons only where it is illuminated by the primary beam (i.e., only within the aperture defined by the jaws and MLC). The sections that follow discuss the photon dose algorithms as currently released in Monaco versions 5 and 6, describing how an IQM beam can be modeled more accurately without any changes to the existing algorithm source code.

3.3 Monaco photon Monte Carlo algorithm

3.3.1 The source model

This section discusses the photon Monte Carlo algorithm as currently implemented in Monaco, with the XVMC radiation transport code and the VSM (Virtual Source Model) 1.6 source model. The following illustration of the model is taken from the Monaco Dose Calculation Technical Reference Manual [9].



1150_MNCO

The model assumes the existence of three virtual sources: a primary photon source, a scattered (secondary) photon source, and an electron source. Some key parameters that determine the characteristics of the beam are as follows (see reference [9] for details):

A. General parameters :

1 NORM: A multiplicative factor governing the overall normalization of the beam.

B. Parameters that govern the primary photon source:

1 PRIMARY-PHOTONS (P_{pri}): The fraction of photons that are primary.

2 PRIMARY-SIGMA (σ_{pri}): One sigma (cm) of the two-dimensional Gaussian distribution representing the geometry of the primary source.

3 OAS-PROFILE: Radial fluence profile for primary photons.

4 ENERGY-MAX (E_{max}): Maximum photon energy (MeV), for primary and scattered photons.

5 B-VALUE: Parameter that governs the shape of the primary photon spectrum.

6 DELTA-B-OAS: Parameter that governs the off-axis variation of the primary spectrum.

C. Parameters that govern the scattered photon source:

1 SCATTER-DIST: *Z*-position (cm) of the scattered photon source.

2 SCATTER-SIGMA (σ_{scat}): Parameter that governs size of the scattered photon source.

3 SEC-B-VALUE: Parameter that governs the shape of the scattered photon spectrum.

4 SEC-DELTA-B: Parameter that governs the off-axis variation of the scattered spectrum.

D. Parameters that govern the contaminant electron source:

1 CHARGED-PARTICLES (P_{econ}): The fraction of all particles that are contaminant electrons.

2 CHARGED-DIST: *Z*-position (cm) of the contaminant electron source.

3 CHARGED-RADIUS (σ_{econ}): Parameter that governs size of the contaminant electron source.

4 CHARGED-E-MEAN ($\bar{E}_{econ}$): Mean energy (MeV) of the contaminant electrons.

5 CHARGED-E-MAX ($E_{charged,max}$): Maximum energy (MeV) of the contaminant electrons.

Note: *Three primary photon source parameters affect the secondary photon source also:*

- PRIMARY-PHOTONS (P_{pri}) is the fraction of photons that are primary. It follows that ($1 - P_{pri}$) is the fraction of photons that are scattered.
- ENERGY-MAX is the maximum energy for scattered photons as well as for primary photons.
- For primary photons, OAS-PROFILE determines the statistical weight directly. For secondary photons, the statistical weight is set to the value that OAS-PROFILE determines for a primary photon passing through the point where the secondary photon originates.

The preceding parameterization of the photon beam allows us to take the following steps to adjust for the presence of the IQM:

1 Decrease NORM to account for the reduction in output.

2 Lower the value of PRIMARY-PHOTONS, because the IQM introduces additional secondary photons, so that relatively fewer photons are primary.

3 Adjust the central-axis primary energy spectrum (ENERGY-MAX and B-VALUE) to account for spectrum hardening as manifested in the central-axis depth dose curve.

4 Adjust OAS-PROFILE, as necessary, to account for spectrum hardening as manifested in the “horns” of the photon beam profile at shallower depths.

5 Adjust DELTA-B-OAS, as necessary, to adjust the off-axis spectrum as manifested in the decay of the “horns” at deeper depths.

6 Increase SCATTER-DIST to reflect the fact that the scattered photons are on average being generated closer to the patient surface.

7 Adjust the scattered photon spectrum (SEC-B-VALUE and SEC-DELTA-B), if there is any observed benefit to doing so.

8 Increase the value of CHARGED-PARTICLES, because the IQM generates electrons much closer to the patient surface, with no downstream collimation to absorb them.

9 Increase CHARGED-DIST to reflect the fact that the contaminant electrons are on average being generated much closer to the patient surface.

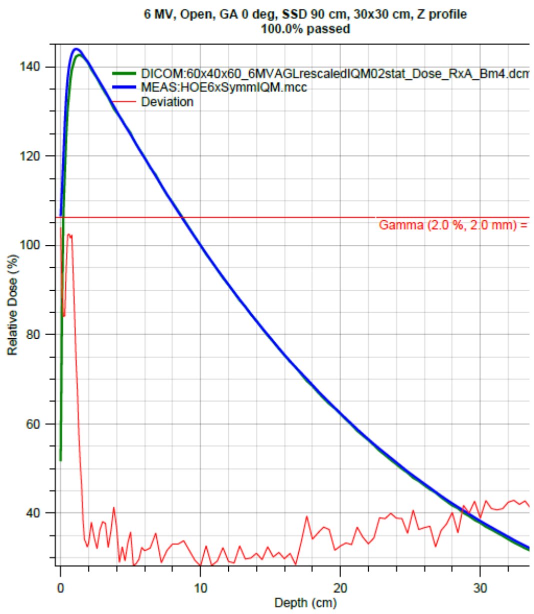
Steps 1 to 5 can faithfully represent the changes in primary photon fluence due to the presence of the IQM. That is, dosimetric effects A and B can be handled very well. The weak points in the above enumeration are steps 6 and 9, respectively for scattered photons and contaminant electrons. For either type of particle, there is, in reality, one major source of secondary particles at $Z < 20$ cm and another significant source of secondary particles at $Z > 50$ cm. It is challenging to create a single virtual source to represent two such different real-world sources. To model the increased out-of-field dose (effect C) and the increased surface dose (effect D) due to the IQM, the beam modeler must experiment interactively with two source weights (PRIMARY-PHOTONS and CHARGED-PARTICLES), two secondary source positions (SCATTER-DIST and CHARGED-DIST), and two secondary source sizes (SCATTER-SIGMA and CHARGED-RADIUS), in search of a clinically optimal result across all field sizes. The results below show that this challenge has been satisfactorily met.

3.3.2 Results

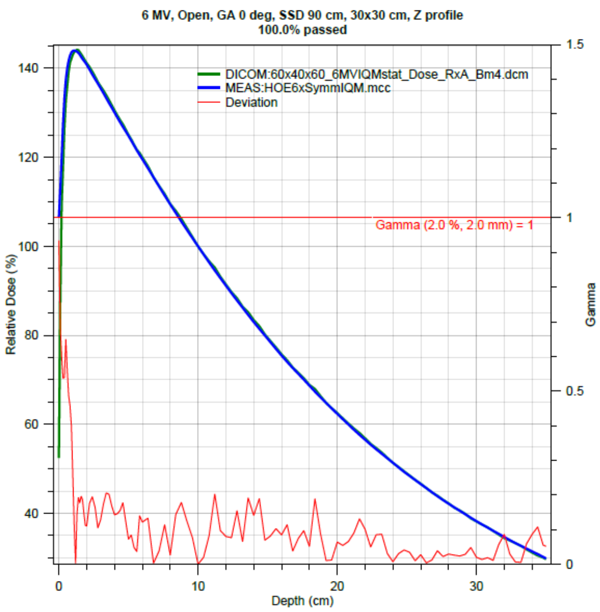
This section compares computed Monte Carlo dose to measured dose for the same 6 MV Elekta Versa HD beam as shown above. The Elekta beam modeling team has created two different models for this beam with the IQM in place: the first is simply renormalized; the second is fully remodeled. For field sizes up to and including 20x20 cm, both models easily satisfy 2%/2 mm gamma criteria. The remodeled beam gives marginally better results in some places (not everywhere), but the differences are not clinically significant.

When we turn our attention to larger fields, we find that the modeling strategies discussed in the previous section work quite well. For the renormalized beam, the output factor is 1.44% too high for the 30x30 cm field and 1.79% too high for the 40x40 cm field. For the remodeled beam, these values are reduced to 0.62% and 1.02% respectively. Both models capture the overall reduction in output (effect A), but the remodeled beam goes beyond the renormalized beam in capturing some of the field size dependence of the effect.

We need to look at dose profiles to see how the remodeling of the beam captures dosimetric effects B through D. Here are the depth dose curves for the 30x30 cm field, with the renormalized beam on the left and the remodeled beam on the right:



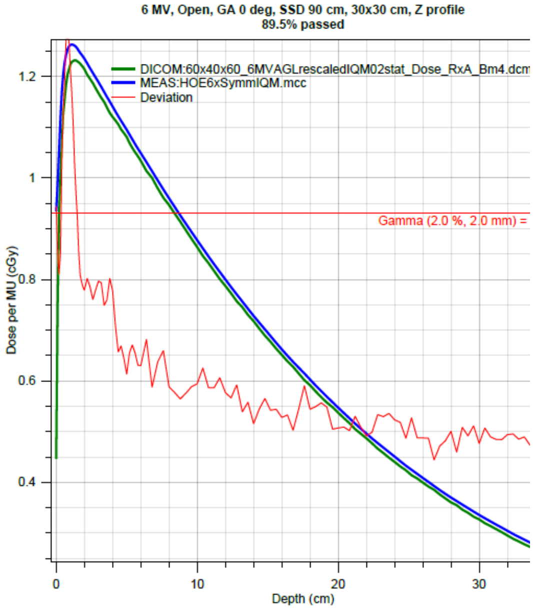
1151_MNCO



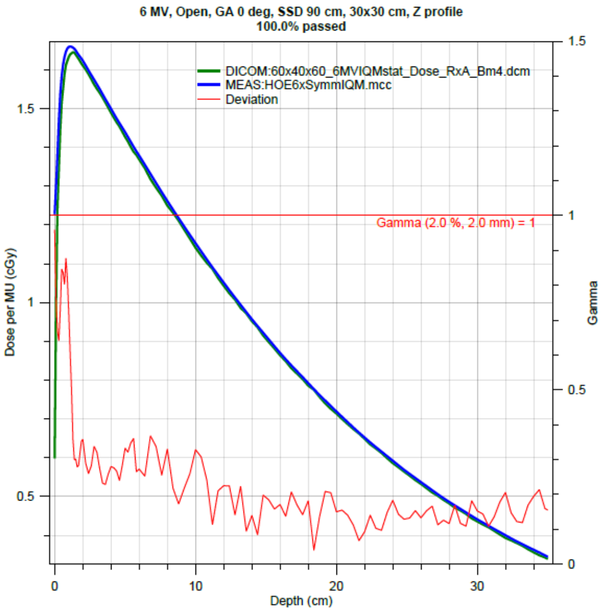
1152_MNCO

The two plots above are presented in relative dose mode, removing the output factor difference described in the previous paragraph. Unmodeled spectrum hardening is visible in the deep end of the depth dose plot on the left, but is effectively eliminated on the right. A significant surface dose deficit is visible on the left, but is greatly reduced on the right.

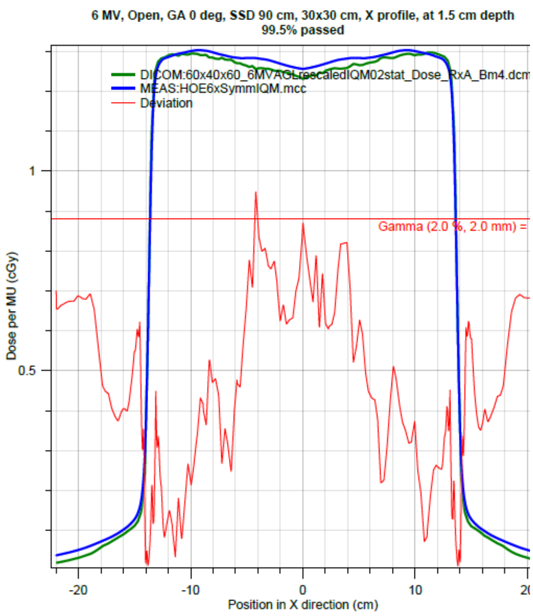
The following plots are presented in absolute dose mode, incorporating output factor differences as well as differences in the shape of the dose distribution. These results thus correspond to final dose distributions as calculated in Monaco. In all cases, the renormalized beam is on the left, and the remodeled beam is on the right. The first pair of images shows the same depth dose curves as above, with their proper normalizations. The second pair compares *X*-profiles at a depth of 1.5 cm, and the third pair compares *X*-profiles at a depth of 20 cm.



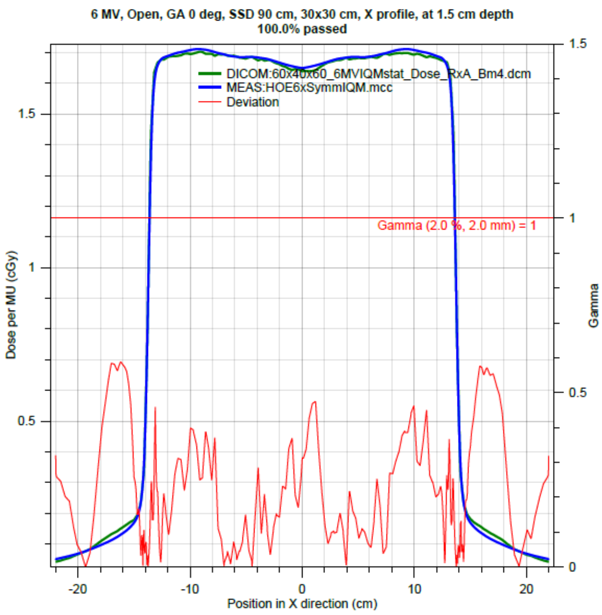
1153_MNCO



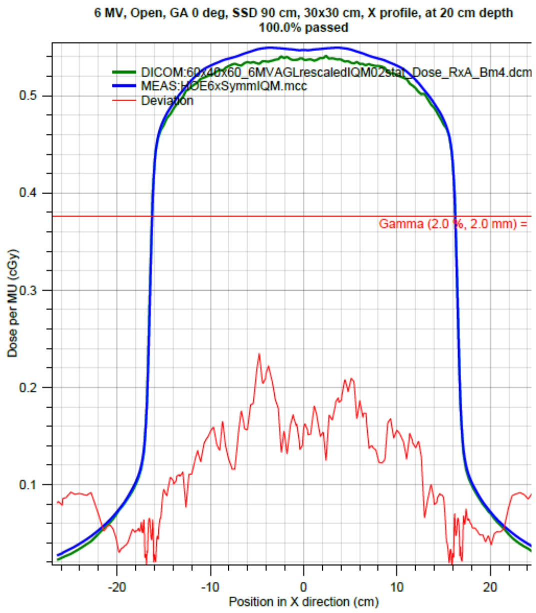
1154_MNCO



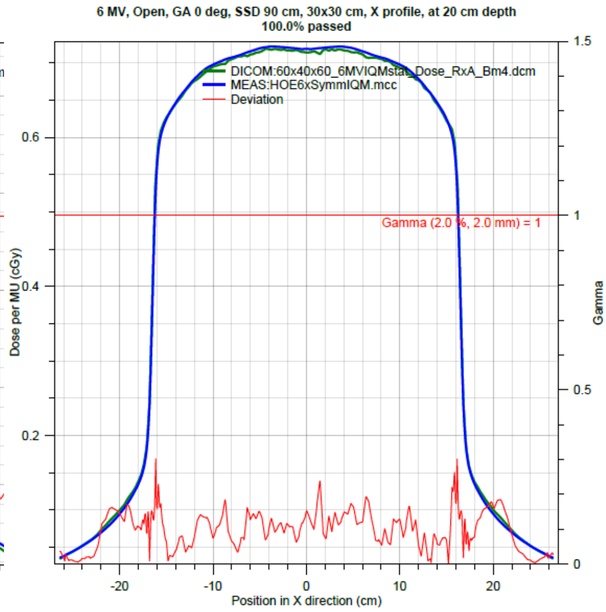
1155_MNCO



1156_MNCO



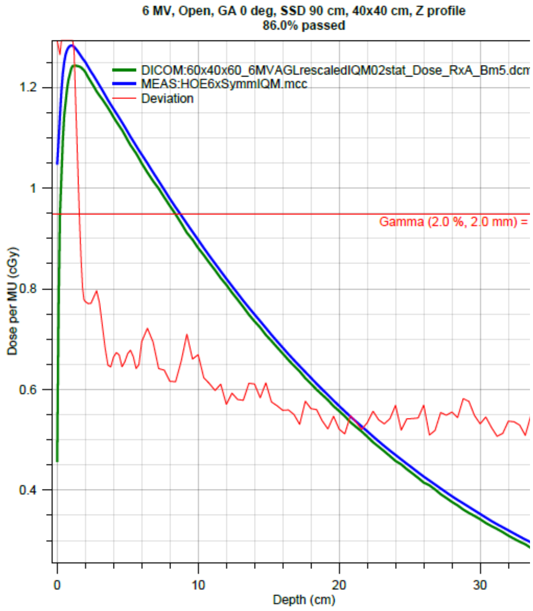
1157_MNCO



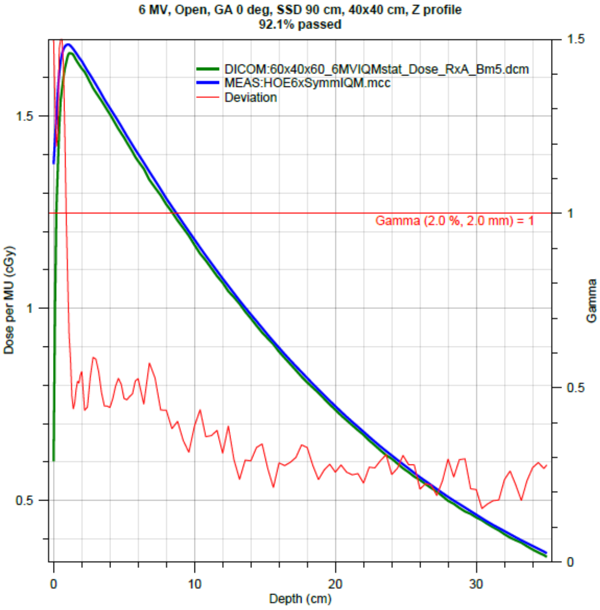
1158_MNCO

The profiles at 1.5 cm show that the remodeling has successfully captured the increased out-of-field dose (effect C) and the increased surface dose due to contaminant electrons (effect D). The profiles at 20 cm show that the remodeling has successfully captured the increased out-of-field dose (effect C) and the hardening of the spectrum that causes a higher central-axis dose at this depth (effect B).

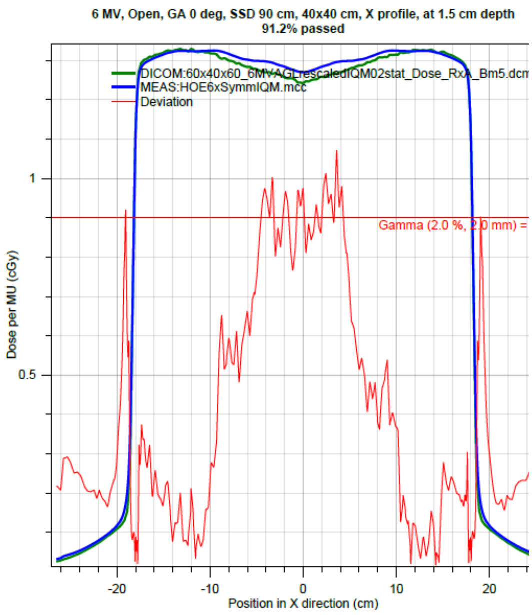
The following six plots are the same in organization and content as the previous six plots, but for a 40x40 cm field of the same beam.



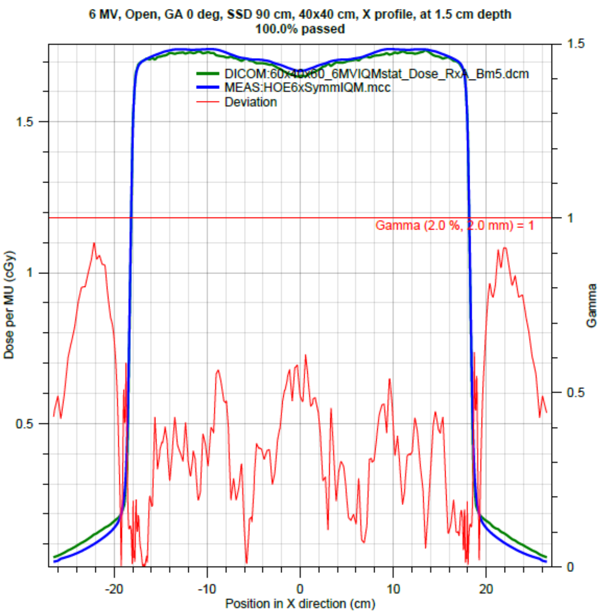
1159_MNCO



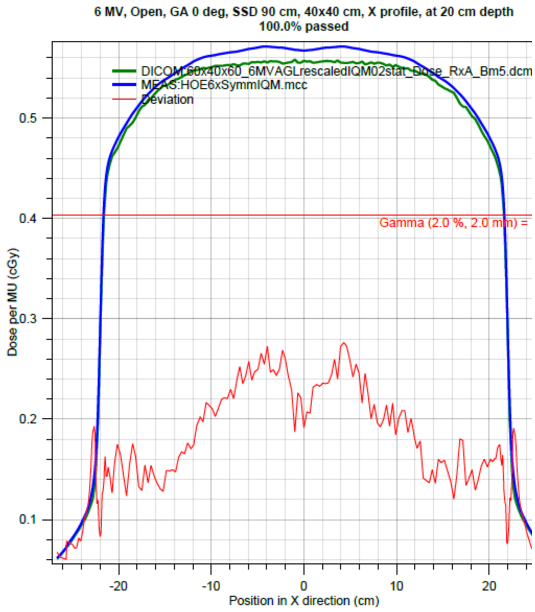
1160_MNCO



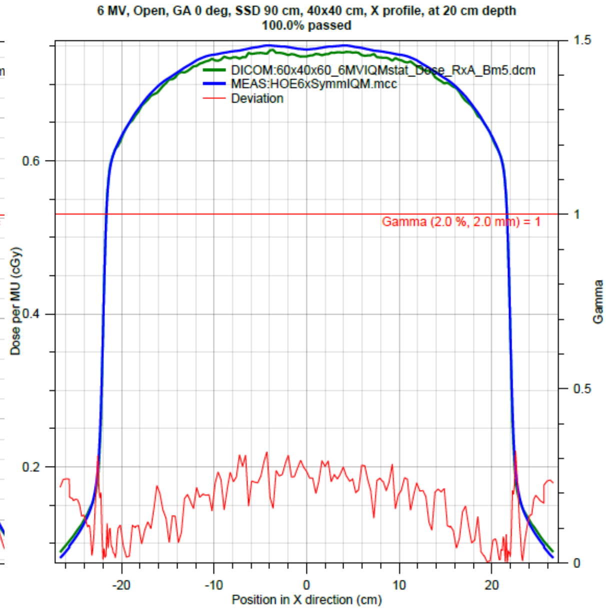
1161_MNCO



1162_MNCO



1163_MNCO



1164_MNCO

The 40x40 cm field is a worst-case scenario for dose computation with the IQM in place, and we see here that the remodeled beam gives clinically acceptable results.

3.4 Monaco Collapsed Cone Convolution algorithm

This section discusses the Collapsed Cone Convolution (CCC) algorithm as currently implemented in Monaco. The CCC source model is spread across multiple files in a folder named “dcm”. These files are protected by checksums and are not directly editable by Monaco users.

3.4.1 Normalization

Monaco currently uses the CCC algorithm only for conventional treatment planning, not for IMRT or VMAT. Theoretically, a conventional beam can be renormalized by introducing an empty photon block with a tray factor equal to the transmission of the IQM; however, Elekta recommends using a dedicated CCC beam model for treatment planning with the IQM. The Monaco user can create a dedicated IQM beam model in the Treatment Unit Storing tool by copying the original beam model and changing the normalization in the copy. Alternatively, the user can request a renormalized CCC beam model from Elekta customer service. In either case, the beam normalization is changed by adjusting the absolute calibration conditions, typically specified as cGy/MU at a depth of 10 cm in a 10x10 cm field.

The CCC and Monte Carlo algorithms differ in the way they handle the calibration conditions. The Monte Carlo algorithm addresses them indirectly. The specification is known to the beam modeler, but never explicitly communicated to the dose algorithm. Instead, the Monte Carlo beam modeler adjusts the NORM parameter up or down empirically until it gives a result that matches the target calibration dose.

In contrast, the CCC algorithm addresses the calibration conditions directly. The specification is entered as part of the beam model (in file PERTDAT1.DAT), and the algorithm forces the dose to have the specified value at the specified depth for the specified field size at the specified SSD. The way to capture the attenuation due to the IQM is then to increase the value labeled “Calculation calibration MU/Gy” in file PERTDAT1.DAT. For example, if the IQM reduces the output of the beam by 5%, then this value in the file might be increased from 100.0 to 105.263.

3.4.2 Primary Photons

The attenuation of the primary photons is captured by the change of normalization, as discussed in the previous section. The hardening of the primary photon spectrum can be modeled in the following ways:

1. The CCC beam model represents the central-axis primary energy spectrum as a table of bin energies and bin weights stored in file SPECTRUMCC1.DAT. This spectrum can be changed as necessary to skew the distribution toward higher energies.
2. The effect of spectrum hardening on the “horns” of the lateral photon beam profile can be incorporated into the fluence matrix, as stored in binary file FLUENMTXCC1.DAT.
3. As necessary, off-axis softening and kernel hardening of the revised spectrum can be modeled using the parameters MUP_A[123] and KAPPAP_C in file OFFAX1.DAT. For details, see section 5.3.6.3 of the Oncentra Physics and Algorithms manual [10].

3.4.3 Scattered Photons

In the CCC source model, the energy spectrum of the scattered photons is determined from the spectrum of the primary photons. The CCC model therefore presents the user with no free parameters for modeling the secondary photon spectrum. The model does, however, allow the user to change the intensity and size of the scattered photon source. Specifically:

1. The intensity of the scattered photon source can be altered by changing the parameter labeled “Total scatter from flattening filter” in file PHOTCALC1.DAT.
2. The size and shape of the scattered photon source can be altered by changing the parameters labeled “Flattening filter projected radius” and “Slope parameter for flattening filter” in file PHOTCALC1.DAT.

3.4.4 Contaminant Electrons

The CCC modeling of electron contamination is described in section 5.3.5.2 of reference [10]. Unlike the Monte Carlo model, the CCC model does not include an explicit source from which the electrons originate. Instead, the distribution of electron contamination dose in the patient is modeled directly using parameters named alpha, beta, and gamma. The beta parameter, which governs the falloff of the electron dose with increasing depth, should not be much altered by the presence of the IQM. On the other hand:

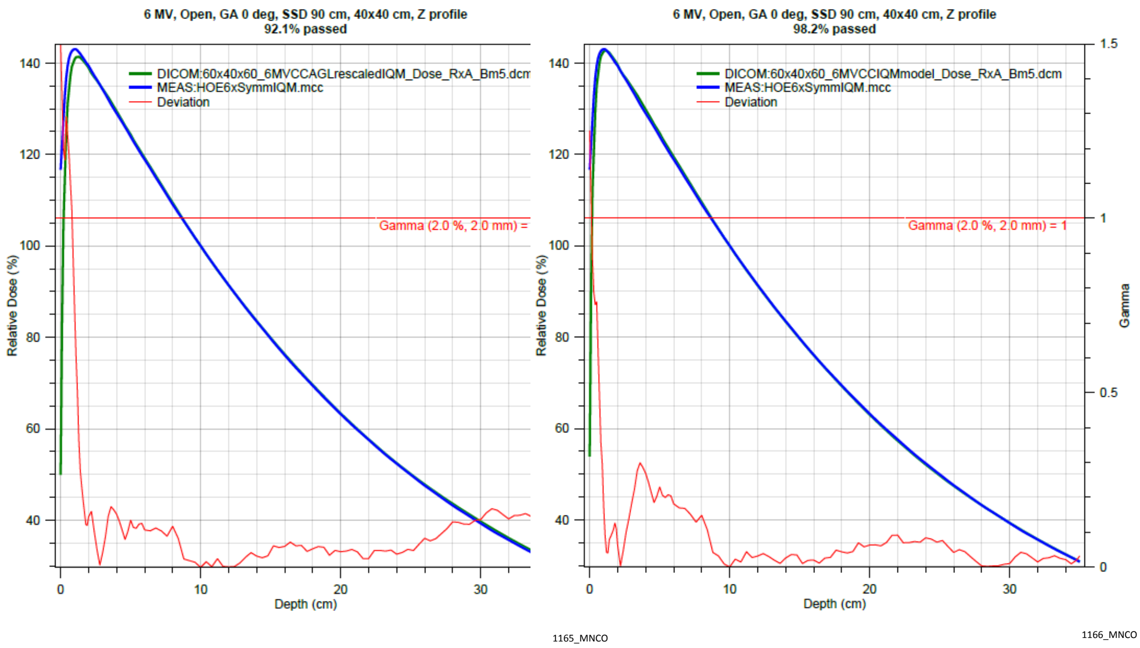
1. The alpha parameter represents the intensity or relative weight of the electron dose. The value labeled “Alpha” in file CHPCONT1.DAT can be increased to model the additional surface dose due to the presence of the IQM.
2. The gamma (γ) parameter governs the lateral spread of the electron dose, by means of a multiplicative factor $e^{-\gamma R^2}$, where $R=\sqrt{X^2+Y^2}$.The value labeled “Gamma” in file CHPCONT1.DAT can be decreased to model the increased lateral distance traveled by the electrons that are created in the IQM and not subsequently collimated.

3.4.5 Results

This section compares computed CCC dose to measured dose for the same 6 MV Elekta Versa HD beam as shown above. For the CCC algorithm, as for the Monte Carlo algorithm, the Elekta beam modeling team has created two different IQM-specific beam models: one simply renormalized, the other fully remodeled. Once again, both models easily satisfy 2%/2 mm gamma criteria for field sizes up to 20x20 cm, with the remodeled beam looking slightly better, but not significantly so.

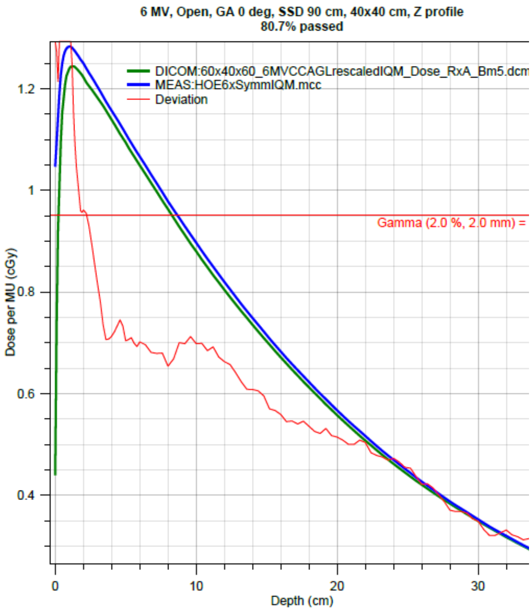
The CCC results for large-field output factors also resemble the Monte Carlo results. For the renormalized beam, the output factor is 1.67% too high for the 30x30 cm field and 1.92% too high for the 40x40 cm field. For the remodeled beam, the errors are reduced to 0.80% and 1.13% respectively. Again we see that remodeling partially captures the field size dependence of the overall reduction in output due to the presence of the IQM.

To avoid excessive repetition, the figures below display CCC profiles only for the worst-case 40x40 cm field. Here are the CCC depth dose curves in relative dose mode, with the renormalized beam on the left and the remodeled beam on the right:

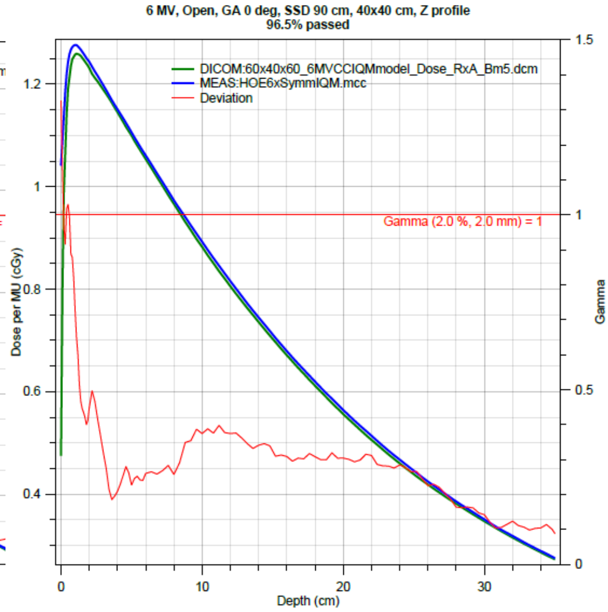


At the deepest depths, one can see hints that the hardening of the spectrum is better captured by the fully remodeled beam. At the shallowest depths, one can see clearly that the excess surface dose is more accurately modeled on the right.

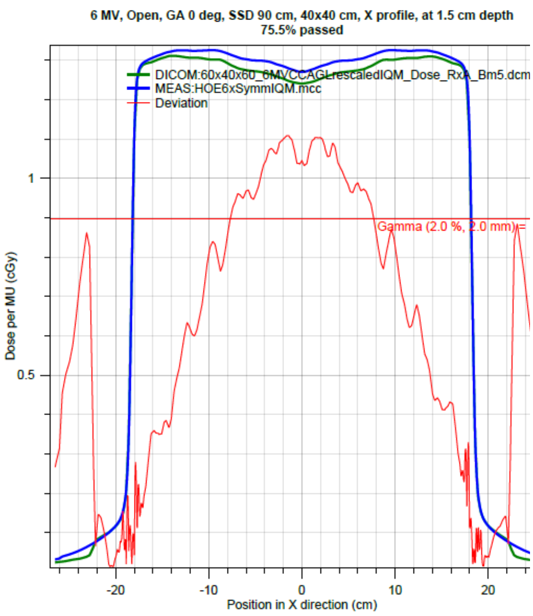
The following plots are presented in absolute dose mode, with the renormalized beam on the left and the remodeled beam on the right. The first pair of images shows the above depth dose curves with their proper normalizations. The second pair compares *X*-profiles at a depth of 1.5 cm, and the third pair compares *X*-profiles at a depth of 20 cm.



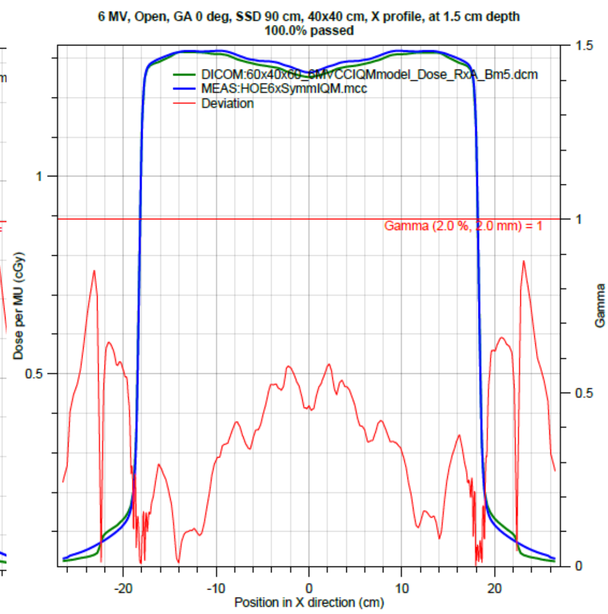
1167_MNCO



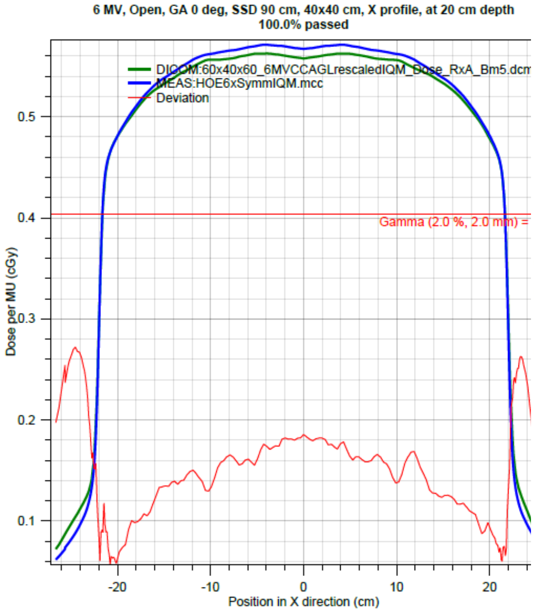
1168_MNCO



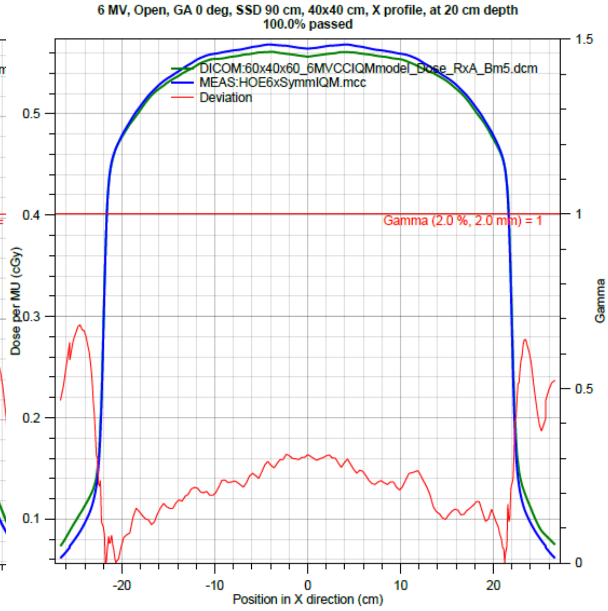
1169_MNCO



1170_MNCO



1171_MNCO



1172_MNCO

As expected, the remodeled CCC beam produces superior dosimetric results for very large fields.

4 Recommendations

4.1 Recommendations

The IQM attenuates a radiotherapy photon beam and requires a modified Monaco beam model for treatment planning. This document has described two different methods for creating such IQM-specific beam models. The first method is a simple renormalization; the second is a full remodeling.

The clinical consensus is that a simple renormalization is usually adequate. We saw above that renormalization alone works very well for square fields up to and including 20x20 cm. Radiation fields used in treatment planning are not square, however, and are often quite elongated. For example, the following table (courtesy of iRT Systems) shows the frequency distribution of field sizes for thousands of 6 MV VMAT plans created with Monaco at fourteen different clinics.

	Y-Dimension [cm]															
	0.5	0.8	1	2	3	4	5	6	8	10	12	15	20	25	30	35
0.4	4	12	20	21	29	26	11	14	5	6	4	2	0	0	0	0
0.5	10	17	30	73	99	90	89	103	73	33	49	58	28	13	4	1
0.8	20	28	84	342	412	546	537	853	753	449	475	757	963	792	70	8
1	59	75	335	1705	3371	4824	6143	12494	14530	9437	9212	20232	29235	21710	4198	424
2	65	108	473	3794	7614	11555	16325	36976	50309	37582	39209	75063	94214	42879	11915	2430
3	97	54	186	959	3672	7472	12669	31850	48073	37362	40169	71928	80530	29821	7531	2167
4	19	10	36	142	861	2666	6672	19688	31569	24162	25320	45280	48591	17520	4260	1569
5	9	7	5	38	227	650	2742	8961	16643	12814	13664	23487	23801	7931	2150	862
6	4	1	0	5	47	367	1133	3962	7319	6960	7739	13692	13039	3639	1000	554
8	1	0	2	3	8	44	254	714	1495	1807	2129	3589	3239	844	211	187
10	0	0	0	0	0	13	21	52	251	248	182	314	429	70	21	29
12	0	0	0	0	0	0	0	1	36	29	22	7	37	1	0	4
14	0	0	0	0	0	0	0	0	10	1	6	0	0	0	0	0
16	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0
20	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0
25	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0
30	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0
35	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0
40	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0

1173_MNCO

The field sizes frequently exceed 20 cm in one dimension, but they are generally narrow in the perpendicular direction, and it is extremely rare for the area of the field to be greater than 400 cm^2. If we review the dosimetric effects discussed above, we find that the simple renormalization approach remains appropriate for such long, narrow fields. Effect A (reduction of output) is primarily a function of energy spectrum and secondarily a function of phantom scatter, for which field area is more relevant than field length. Effects C and D (increased out-of-field dose and surface dose) likewise depend on the total area of the IQM device that is irradiated by the primary photons. The only phenomenon that depends directly on the length of the field is the slight suppression of the beam "horns" due to spectral hardening (effect B). This effect is relatively small and affects only the outermost regions of rarely occurring fields.

Clinics that choose the simple renormalization approach will be able to commission their IQM more quickly. Only the following steps are required for each beam energy and fluence type:

- Take one or more point measurements of the transmission through the IQM.
- Request, or use the Treatment Unit Storing tool to create, a beam model in which only a single normalization or calibration value has been changed.
- Validate the model and begin treating with the IQM.

In contrast, the procedure for a full remodeling is more time-consuming and burdensome, requiring the following steps:

Recommendations

- Gather all of the water tank scans and point measurements specified for a new photon beam in the Monaco beam data collection list [11].
- Submit the complete set of measured data to Elekta via the Elekta Physics Platform (EPP).
- Wait for the Elekta beam modeling team to create and deliver an entirely new beam model, as for an entirely new beam.
- Validate the model and begin treating with the IQM.

The full remodeling procedure is recommended only for clinics that plan to use their IQM for fields with area larger than about 400 cm^2.

5 References

5.1 References

[1] *Monaco: Instructions for Use for: Monaco 6.1*. Elekta document ID: E018853. See warning 1.7 in section 1.15, "Treatment Unit Mapping".

[2] IQM System specifications web page: https://www.iqm-system.com/system/iqm-specifications/.

[3] Miori, G., et al. "Evaluation of an integral quality monitor device for monitoring real-time delivery". Poster number A.144 from the 9th national congress of the Italian Association of Medical Physics (AIFM) in Perugia, Italy in February 2016. Available online at https://i-rt.de/publications?_sft_publication_category=beam-attenuation .

[4] Hoffman, D., et al. "Characterization and evaluation of an integrated quality monitoring system for online quality assurance of external beam radiation therapy". *J. Appl. Clin. Med. Phys*. **18** (2017) 40–48.

[5] Nguyen, T.H.T., et al. "Effect of an integral quality monitor on 4-, 6-, 10-MV, and 6-MV flattening filter-free photon beams". *J. Appl. Clin. Med. Phys.* **22** (2021) 76–91.

[6] Casar, B., et al. "Influence of the Integral Quality Monitor transmission detector on high energy photon beams: A multi-centre study". *Z. Med. Phys.* **27** (2017) 232–242.

[7] *IQM User Reference Manual*. v1.35, SW-v1.8. See warning 7 in section 2.4, "Safety Notices".

[8] Perrin, B., et al. "Integrating the IQM into a commercial planning system". Christie Hospital NHS Trust, Withington, Manchester, UK. Conference presentation as shown during the ESTRO35 Conference in Turin, Italy in May 2016. Available online at https://www.iqm-system.com/publications/, under the heading "Beam Attenuation".

[9] *Monaco Dose Calculation Technical Reference Manual.* Elekta document ID: E014439.

[10] *Oncentra External Beam v4.5: Physics and Algorithms*. Version 170.730. Nucletron document ID 777.00357MAN-00.

[11] *Monaco: Beam Data Requirements, Photons and Electrons*. Elekta document ID: E056094.

We are healthcare technology innovators, specializing in radiotherapy treatments for cancer and brain disorders.

We help clinicians to improve patients' lives through our forward-thinking treatment solutions and oncology informatics, creating focus where it matters to achieve better outcomes.

Elekta

Elekta Offices

Elekta AB
Kungstensgatan 18
Box 7593, SE-103 93 Stockholm
Sweden
T +46 8 587 254 00
F +46 8 587 255 00

Europe, Middle East, Africa
T +46 8 587 254 00
F +46 8 587 255 00

North America
T +1 770 300 9725
F +1 770 448 6338

Latin America, South America
T +55 11 5054 4550
F +55 11 5054 4568

Asia Pacific
T +852 2891 2208
F +852 2575 7133

Japan
T +81 3 6722 3800
F +81 3 6436 4231

China
T +86 10 5669 2800
F +86 10 5669 2900

elekta.com

/elekta

@elekta

/company/elekta