Efficacy of the virtual cone method using fixed small multi-leaf collimator field for stereotactic radiosurgery

by

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This dissertation was prepared under the direction of the candidate’s dissertation co-advisors, Dr. Theodora Leventouri, Dr. Charles Shang, Department of Physics, and has been approved by the members of his supervisory committee. It was submitted to the faculty of the Charles E. Schmidt College of Science and was accepted in partial fulfillment of the requirements for the degree of Doctor of Philosophy.

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Abstract

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Dosimetric uncertainty in very small (< 2 x 2 cm²) photon fields is notably higher that has created research questions when using small-field virtual cone with variable multi-leaf collimator (MLC) fields. We evaluate the efficacy of the virtual cone with a fixed MLC field for stereotactic radiosurgery (SRS) of small targets such as trigeminal neuralgia.

We employed a virtual cone technique with a fixed field geometry, called fixed virtual cone (fVC), for small target radiosurgery using the EDGE (Varian Medical Systems, Palo Alto, CA) linac. The fVC is characterized by 0.5 cm x 0.5 cm high-definition MLC field of 10 MV flattening filter-free (FFF) beam defined at 100 cm SAD, while jaws are positioned at 1.5 cm x 1.5 cm. A spherical dose distribution equivalent to 5 mm cone was generated by using 10–14 non-coplanar partial arcs. The dosimetric accuracy of this technique was validated using the SRS MapCHECK (Sun Nuclear Corporation, FL) and the EBT3 (Ashland Inc., NJ) film based on absolute dose measurements. For the quality
assurance (QA), 10 treatment plans for trigeminal neuralgia consisting of various arc fields at different collimator angles were analyzed retrospectively using 6 MV and 10 MV FFF beams, including the field-by-field study (n = 130 fields). Dose outputs were compared between the SRS MapCHECK measurements and Eclipse treatment planning system (TPS) with Acuros XB algorithm (version 16.1). In addition, important clinical parameters of 15 cases treated for trigeminal neuralgia were evaluated for the clinical performance. Moreover, dosimetric (field output factors, dose/MU) uncertainties considering a minute (± 0.5–1.0 mm) leaf shift in the field defining fVC, were examined from the TPS, SRS diode (PTW 60018) measurements, and Monte Carlo (MC) simulations.

The fVC technique for small target radiosurgery was validated (≤ 3% difference) using the SRS MapCHECK and EBT3 film measurements. From the field-by-field study (10 treatment plans, 130 fields), the average differences in dose/MU between measurements (SRS MapCHECK) and TPS (Acuros XB) were found 4.0% and 2.5% for the 6 MV and 10 MV FFF beams respectively. However, the dose (at central axis) discrepancy jumped up to 10% for certain fields (arcs) depending on the gantry, collimator angles, and beam energy. In addition, the dosimetric parameters determined from diode measurements and TPS using the field defining fVC, including ± (0.5–1.0 mm) leaf shift, yielded 3–6% uncertainty. Furthermore, early results of the MC study (water phantom) for these fields have shown similar (trend) dosimetric variation as measured. However, the model needs further improvements for the final assessment.

The viability of the fVC method for small target radiosurgery and its dosimetric robustness considering potential minute leaf shift during the treatment are evaluated from measurements and MC calculations. This method is a straightforward that yields a higher
dosimetric certainty and is a reliable or superior alternative to a physical cone that can be routinely applied for the treatment of trigeminal neuralgia. This research can be extended further by a machine learning approach based on MLC log files as well as radiomics analysis from QA images.
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1. Introduction

1.1 Radiation Therapy

Radiation therapy is a medical application of the ionizing radiation to treat lesions such as cancers and tissue with abnormalities.\(^1\) The radiation causes damage not only to the cancerous tissue, but also to the healthy one. Therefore, the radiation beam needs to be delivered precisely targeted to maximize the dose to the lesion or tumor and minimize the dose to the surrounding healthy tissue. The radiation dose delivered to both, tumor and healthy tissue also needs to be measured accurately to ensure that the appropriate dose is delivered. If the tumor received less dose than intended, then tumor control would be compromised, and the treatment may not be effective in removing the cancerous cells. If nearby healthy tissues, including vital organs, receive doses higher than the tolerable limit, then the patient may have radiation-induced complications. Thus, the accuracy of dose delivery becomes a vital part of radiotherapy treatments.

In Radiation therapy, either photons or charged particles are commonly used\(^1\). In the photon therapy mode of a medical linear accelerator (LINAC), Bremsstrahlung x-rays (photons) are produced when accelerated electrons are incident on a target of a high-$z$ material, such as tungsten. The target is thick enough to absorb most of the incident electrons. As a result, the electron energy is converted into a spectrum of x-ray photons. The average photon energy of the beam is approximately one third of the maximum energy
of the incident electron beam. When the electron therapy mode is used, the electron beam exits the window of the accelerator tube in a narrow pencil shape (about 3 mm in diameter) without striking the target. Then, it strikes an electron scattering foil to spread the beam as well as get uniform electron fluence across the treatment field. The scattering foil consists of a thin metallic foil. The thickness of the foil is such that most of the electrons are scattered instead of converting their kinetic energy into bremsstrahlung photons. However, a small fraction of the total electron beam energy is still converted into bremsstrahlung and appeared as x-ray contamination of the electron beam. Efforts made to avoid contamination of the electron beam with photons. In some LINACs, the broadening of the electron beam accomplished by electromagnetic bending, instead of using the foil. Although this minimizes the x-ray contamination of the electron beam, some x-rays are still produced by electrons striking the collimator walls, or other high atomic number materials in the electron collimation system. Figure 1 shows schematic diagram of the two radiation therapy modes that we discussed.
1.1.1 Photon Interactions with Matter

Photons may undergo various possible interactions with the atoms (nuclei or orbital electrons) in the penetrating medium; the probability or cross-section of each interaction will depend on the energy \((\hbar \nu)\) of the photon and atomic number \((Z)\) of the medium.\(^3\) The interactions with nuclei may be direct photon-nucleus (photodisintegration) interactions or photon interactions with electrostatic field of the nucleus (pair production). The photon

Figure 1: Schematic diagrams of two radiation therapy modes: (A) Photon therapy mode, (B) Electron therapy mode.\(^3\)
interactions may be (1) with a loosely bound electron (Thompson scattering, Compton Effect, and triplet production) (2) with a tightly bound electron (Photoelectric effect, Coherent scattering) or atom as a whole (pair production). During the interaction, the photon may either be completely disappeared (Photoelectric effect, pair production, triplet production) and a portion of its energy is transferred into light charged particles (electrons, positrons) or it may be scattered coherently (Coherent scattering) or incoherently (Compton scattering) as shown in figure 2. The light charged particles produced in the absorbing medium will either deposit their energy to the medium through Coulomb interactions with orbital electrons of the absorbing medium (collision/ionization loss) or radiate their energy away through Coulomb interactions with the nuclei of the absorbing medium (radiative loss).

Figure 2: Modes of photon interactions with matter depending on energy and atomic number.
The charged particles such as electrons, protons, and α-particles are known as directly ionizing radiation because they have sufficient kinetic energy to produce ionization (the process in which a neutral atom acquires a positive or negative charge, resulting in an ion pair) by collision as they penetrate through matter. While the uncharged particles such as photons and neutrons refer to indirectly ionizing radiation as, they liberate charged particles when they interact with matter. In general, the photon interaction through a material or absorber of thickness $x$ is described as:

$$I(x) = I_0 e^{-\mu x}$$

Where $I_0$ and $I(x)$ are the intensities of incident and transmitted beams on the absorber respectively, and $\mu$ is the linear attenuation coefficient.

In photon therapy, megavoltage energy (MeV) x-rays/photons from a Linac interact with matter (phantom/human tissue) predominantly by Compton interactions i.e. a photon strikes an orbital electron with energy much greater than the binding energy of that electron, causing it to be ejected from the atom. The ejected electron is so highly energized that it will travel of the order of centimeters through the tissue before losing all of its energy. Along its path, the electron deposits its energy via ionization and excitation of other orbital electrons, which contributes to the absorbed dose (energy deposited per unit mass) to the human tissue or phantom. Although the incident particles are photons, it is actually electrons that contribute to the absorbed dose.
Compton electrons or other secondary electrons can kill tissue cells by interacting directly or indirectly with the DNA of the molecules. In direct interactions, the electrons produced by the radiation beam ionize the DNA molecules. In the indirect interactions, the electrons ionize molecules within the tissue (e.g., water) and produce free radicals. These free radicals are highly reactive and can subsequently interact with DNA molecules causing cell death.

1.1.2 Measuring Radiation Dose

The absorbed radiation dose which is the locally deposited energy per unit mass (J/kg or Gray, Gy), can be measured by a calibrated radiation detector in the water phantom. Such measurement is known as absolute dosimetry. As the water mimics the human tissue (relative electron density range 0.7–1.2), a gas-based ionization chamber is submerged in a water tank to perform the measurement. The radiation causes ionization in the gas cavity of the ionization chamber; the ionization charge can be collected and measured as a small current. This can be related to the absorbed dose at the detector. Absolute dosimetry is usually performed under a chosen set of parameters (e.g., the radiation field size is a 10 x 10 cm², the depth in water is 10 cm, the distance from the radiation source to the surface (SSD), or the detector level (SAD) is 100 cm).

Another method of measuring the dose under certain reference conditions is called relative dosimetry. The dose measured under certain conditions is relative to the dose measured under a reference set of conditions. One example would be measurement of dose by changing the field size but keeping all other conditions intact. The dose measured in one field size is normalized to the dose from the reference field size (usually 10 x 10 cm²).
The relative dose (output factors) is applied to the absolute dose calibration value to obtain the desired dose to a patient under the patient specific conditions.

**Figure 3:** Schematics of Source to Surface Distance (SSD) and Source to Axis Distance (SAD) set-ups in photon reference dosimetry (Courtesy: AAPM TG-51 protocol).

### 1.2 Cavity Theories

Dosimetry of megavoltage photon beams is done by using an ionization chamber based on the Bragg-Gray, Spencer-Attix cavity theories, or based on other cavity theories.\(^4\)\(^,\)\(^7\) We briefly describe the main cavity theories.
1.2.1 Bragg-Gray Cavity Theory

The Bragg-Gray (B-G) cavity theory states that the amount of ionization produced in a small gas-filled volume cavity surrounded by a larger, solid absorbing medium is directly proportional to the radiation energy absorbed by the solid medium.\(^1\) This cavity theory works if the following conditions (B-G conditions) are satisfied: (1) the range of the dose depositing electrons must be much greater than the cavity dimensions, (2) charged particle equilibrium (CPE) must exist, (3) the detector must not perturb the fluence of the dose depositing electrons. Figure 4 shows schematics of B-G cavity theory for the particles passing through gas cavity enclosed by wall material.

![Diagram of B-G cavity theory](https://www.flickr.com/photos/mitopencourseware/3706784753/in/photostream/)

**Figure 4:** A schematic of B-G cavity theory for the particles passing through gas cavity enclosed by wall material. (https://www.flickr.com/photos/mitopencourseware/3706784753/in/photostream/)
Under ideal measurement conditions, the first two conditions are satisfied. However, a gas cavity will always have a small effect on the incident beam fluence. This effect must be considered by way of a correction factor. Also, an ionization chamber is not pure gas, but rather consists of a central electrode and an outer wall etc. These facts are also considered by applying additional correction factors. In relative dosimetry all the B-G conditions and correction factors must be identical in both the reference field size and the test field size.

When identical charged particles of kinetic energy $T$ and fluence $\Phi$ pass through an interface between two different media $g$ and $w$, the absorbed doses for $g$ and $w$ are given by the equations:

Equation (2)

$$D_g = \Phi \left[\frac{dT}{\rho dx}\right]_{c,g}$$

Equation (3)

$$D_w = \Phi \left[\frac{dT}{\rho dx}\right]_{c,w}$$

Where $\left[\frac{dT}{\rho dx}\right]_{c,g}$ and $\left[\frac{dT}{\rho dx}\right]_{c,w}$ are the mass collision stopping powers of the media $g$ and $w$ respectively for electron beam energy $T$, assuming that the electron beam is monoenergetic.

If the fluence $\Phi$ is continuous across the interface, the ratio of the absorbed doses in the two media is:

Equation (4)
\[
\frac{D_w}{D_g} = \left(\frac{dT}{\rho d\chi}\right)_{c,w} = \frac{\bar{S}_w}{\bar{S}_g} = \bar{S}_w'
\]

Where \(\bar{S}_w, \bar{S}_g, \bar{S}_g'\) are the average mass collision stopping powers in the cavity medium \(g\), for a thin layer of wall material \(w\), and the ratio of average mass collision stopping powers of media \(w\) to \(g\) respectively.

If charge \(Q\) (of either sign) is produced in medium \(g\) and \(W\) is the average work required to produce charge \(e\), the dose \(D_g\) is expressed by:

Equation (5)

\[
D_g = \frac{Q}{m} \left(\frac{W}{e}\right)_g
\]

Thus, the B-G relation is expressed as:

Equation (6)

\[
D_w = \frac{Q}{m} \left(\frac{W}{e}\right)_g \cdot \bar{S}_g'
\]

Where \(\left(\frac{W}{e}\right)_g\) is the mean energy per unit charge (J/C). The importance of this equation is that \(Q\) is measurable quantity and from that the absorbed dose in the medium can be calculated.

1.2.2 Spencer-Attix Cavity Theory

To reduce the unrealistic assumptions of the B-G theory, some corrections are needed. Specifically, the stopping-power ratio is evaluated assuming continuous slowing-down
approximation (CSDA) of a monoenergetic beam of charged particles. In addition, $\delta$-rays (energetic electrons) are produced that contribute to the flux of the incident beam of electrons crossing the cavity resulting in non-monoenergetic beam with alternated fluence.

Spencer’s goal in modifying the B-G cavity theory was not only to incorporate the $\delta$-rays effect, but also to do it in such a way that the observed variation of ionization density with cavity size could be accounted for, at least for cavities small enough to satisfy B-G conditions. Spencer’s theory (Spencer and Attix, 1955; Spencer 1965, 1971) starts with the two B-G conditions, the existence of CPE, and the absence of bremsstrahlung generation. The derivation specially addresses the case of a distributed homogeneous source of monoenergetic electron beam of initial energy $T_0$ emitting $N$ particles per gram throughout a homogeneous medium $w$.

In this theory, the cavity, containing medium $g$ (usually air), is characterized with respect to size by a parameter $\Delta$, which is taken approximately equal to the range of the mean energy of the electron beam. The equilibrium spectrum, $\Phi_{e,\delta}^{e,\delta}$, of electrons, including $\delta$-rays generated in the surrounding medium is arbitrarily divided into two components in Spencer’s calculations: (1) the fast group electrons, with energies $T \geq \Delta$ that can transport energy, and (2) the slow group electrons, with $T < \Delta$ that do not have enough energy to cross the cavity, nor to transport energy. The absorbed dose at any point in medium $w$, where CPE exists, is given by

Equation (7)
\[ D_w^{\text{CPE}} = NT_0 = \int_{\Delta}^{T_0} \Phi_T^{e,\delta} \cdot S_w(T, \Delta) dT \]

Where \( S_w(T, \Delta) \) is the restricted stopping power for electrons of energy \( T \) in medium \( w \), which includes only energy losses to \( \delta \)-rays not exceeding \( \Delta \). The equilibrium spectrum, including \( \delta \)-rays, can be expressed as:

Equation (8)

\[ \Phi_T^{e,\delta} = \frac{NR(T_0, T)}{(dT/\rho dx)_w} \]

Where \( R(T_0, T) \) is the ratio of the differential electron fluence including \( \delta \)-rays to that of primary electrons alone.

Equation (9)

\[ D_w^{\text{CPE}} = NT_0 = N \int_{\Delta}^{T_0} \frac{R(T_0, T)}{(dT/\rho dx)_w} \cdot S_w(T, \Delta) dT \]

Equation (10)

\[ D_g = N \int_{\Delta}^{T_0} \frac{R(T_0, T)}{(dT/\rho dx)_w} \cdot S_g(T, \Delta) dT \]

Equation (11)

\[ \frac{D_g}{D_w} = \frac{\int_{\Delta}^{T_0} \frac{R(T_0, T)}{(dT/\rho dx)_w} \cdot S_g(T, \Delta) dT}{\int_{\Delta}^{T_0} \frac{R(T_0, T)}{(dT/\rho dx)_w} \cdot S_w(T, \Delta) dT} \]

The Spencer-Attix cavity theory gives better agreement with experimental observations.
than B-G theory, by taking account of $\delta$-rays production and relating the dose integral to the cavity size by introducing the electron beam range.

1.2.3 Burlin Cavity Theory

Burlin (1966, 1968) recognized the need for $\gamma$-rays cavity theory that would bridge the gap between small cavities for which the B-G or Spencer-Attix theory could be applied, and large cavities for which the wall influence is negligible. Burlin made the following assumptions:

1. The media $w$ and $g$ are homogenous.
2. A homogenous $\gamma$-rays field exists everywhere throughout $w$ and $g$. This means that no $\gamma$-rays attenuation correction is made in this theory for the presence of the cavity.
3. Charged particle equilibrium exists at all points in $w$ and $g$ that are wider than the maximum electron range.
4. The equilibrium spectra of secondary electrons generated in $w$ and $g$ are the same.
5. The fluence of the electrons entering from the wall is attenuated exponentially as it passes through the medium $g$, without changing its spectral distribution.
6. The fluence of electrons that originate in the cavity builds on to its equilibrium value exponentially as a function of distance into the cavity, according to the same attenuation coefficient that applies to the incoming electrons.

The Burlin cavity equation is:

Equation (12)
\[ \frac{D_g}{D_w} = d \cdot S_{w}^{g} + (1 - d) \left( \frac{\mu_{en}}{\rho} \right)_{w}^{g} \]

Where \( d \) is the parameter related to the cavity size that approaches unity for small cavities and zero for large one. \( D_g \) is the average absorbed dose in the cavity medium \( g \); \( D_w = (K_c)_w \) is the absorbed dose in medium \( w \) under CPE conditions; \( S_{w}^{g} \) is the mean ratio of mass collision stopping powers for \( g \) and \( w \), obtained either on the basis of B-G, or Spencer-Attix theory; and \( \left( \frac{\mu_{en}}{\rho} \right)_{w}^{g} \) is the mean ratio of the mass energy absorption coefficients for \( g \) and \( w \). Other cavity theories are not discussed here because they are beyond the scope in this study.

1.3 Small Field Dosimetry

1.3.1 Definition

In general, a radiation field of size \( \leq 3 \times 3 \) cm\(^2\) is considered as small field in megavoltage photon dosimetry,\(^8\) whereas the field size of \( \leq 1.5 \times 1.5 \) cm\(^2\) can be considered very small for 6 MV beams at 1% output factor uncertainty level.\(^9\) In other words, a radiation field can be considered as a small field if it satisfies one or more of the following conditions: (1) loss of lateral charged particle equilibrium (LCPE) in the beam axis, (2) collimating devices partially occlude the primary photon source in the beam axis, and (3) the detector size is large or similar to the beam dimensions. The definition of a small radiation field is also subject to the detector characteristics with respect to the lateral range of the charged particles produced in the photon beam.

For simplicity, a photon field should be considered as a small field if the distance
from the central axis to the field edge is smaller than the lateral range for charged particle equilibrium. In practice, this distance should be measured with respect to the outside dimension of the detector. Basic characteristics of small fields are loss of lateral charge particle equilibrium, partial source occlusion, detector volume averaging, high dose gradients, decreased dose output, field size smaller than detector size, and large dosimetric uncertainties. Beam schematics in the broad field (left) versus small field (right) is shown in figure 5.

**Figure 5:** Schematic representation the source occlusion effect for the broad photon beam (left) and small photon beam (right) dosimetry.\textsuperscript{10,11}

1.3.2 Dosimetric Challenges

In broad fields (≥\(4 \times 4 \text{ cm}^2\)), the field size exceeds the range of lateral scatter equilibrium (LSE).\textsuperscript{8} The LSE breakdowns when the distance between the point of measurement and
the edge of the field is shorter than the range of the laterally scattered electrons. The field size defined by the collimator setting corresponds well with the full-width half maximum (FWHM) of the lateral beam profile at the isocenter depth and measuring the FWHM is often used as a verification of the field size setting. In addition, dose distribution in broad fields is only dependent on the source-to-surface distance (SSD) and the energy. This makes using broad fields more convenient in a clinical setting. However, in small fields, the FWHM of the resulting field is not consistent with the geometrical definition of the field due to the partial source occlusion effect and the loss of LCPE described in previous section and resulting reduction of beam output. Because the central axis maximum dose value is reduced, the FWHM is determined by a lower position on the penumbral curve as shown in figure 6. The irradiation field size specified at 50% relative dose level becomes larger than the geometrical field size defined by the projected collimator settings, an effect which is called the apparent widening of the field. For a given source to detector distance (SDD), this effect depends on the source to collimator distance (SCD). It has been shown that the detector response and perturbation effects are determined by the FWHM at the measurement depth rather than on the collimator setting. Therefore, the FWHM of the lateral beam profile is the most representative and essential field size parameter for accurate small-field dosimetry. Hence, the field size for small-field dosimetry is defined in IAEA TRS-483\textsuperscript{10} as irradiation field size or FWHM of the field at the position of the detector. To facilitate establishing a relationship between FWHM and collimator setting, it is recommended that both FWHM and geometrical field size, together with the depth it belongs to, be recorded when reporting small-fields data.
Figure 6: Effect of overlapping penumbra due to source occlusion on the FWHM of the lateral beam profiles for (a) broad field, (b) small field, and (c) very small field at a depth with TCPE, illustrating the apparent field widening effect in smaller fields as compared to the collimator setting.\(^8\)

Thus, the calculation of total dose in the small-fields is affected by mainly three factors:\(^8\) (1) lateral charged particle equilibrium. It is lost, because the size of the field or the FWHM is small as compared to the range of the dose depositing electrons. (2) The finite size of the primary photon beam source or focal spot. Its size is usually defined as the FWHM of the bremsstrahlung photon fluence distribution exiting the target. This is relatively large compared to the field size; and as a result, small fields have a very large percentage of the field made up by overlapping penumbra which produces a lower beam output on the beam axis. (3) The finite size of the detector. A detector produces a signal that is proportional to
the mean absorbed dose over its sensitive volume and the signal is affected by the homogeneity of the absorbed dose over the detection volume (volume averaging). Besides volume averaging, the perturbation of the charged particle fluence due to the presence of a detector is a key issue and it must be noted that both effects are always entangled. In the presence of high dose gradients and in the loss of LCPE conditions, fluence perturbations become large and difficult to model since they can depend on minor variations of the detector design, even within engineering tolerances. Therefore, corrections for volume averaging will also have large uncertainty.

**Figure 7:** A schematic representation for the lateral charged particle equilibrium (LCPE) condition in broad (left) versus narrow (right) photon fields.

The dosimetric uncertainties will start to appear as soon as the external edge of the detector volume is at a distance from the field edge smaller than the range of lateral charged particle equilibrium ($r_{LCPE}$). To avoid such small-field conditions in central axis measurements, the
beam half width or radius must be at least as large as $r_{\text{LCPE}}$ plus half the size of the external dimension of the detector. The reference field size for any small field measurement must reach the conditional basis that it is not specified as a small field. In other words, to achieve a charged particle equilibrium (CPE) or avoid small field conditions, the full-width half maximum (FWHM) of the field must satisfy the condition:

Equation (13)

$$FWHM \geq 2r_{\text{LCPE}} + d$$

Where $d$ is the greatest distance between two points on the outer boundary of the chamber, and $r_{\text{LCPE}}$ is the lateral charged particle equilibrium range. This means the outer boundary of the detector to the field edge must be larger than the range of lateral charged particle equilibrium as shown in a figure 8 illustrating a small cylindrical ion chamber used in small field dosimetry.
Figure 8: An Illustration for a small field detector influenced by the relative active volume diameter (d) and the size of the radiation field in small field dosimetry. Light blue color (square) represents the radiation field and yellow color (circle) represents a field for the lateral charged particle equilibrium to exist.\textsuperscript{12}

The $r_{LCPE}$ is a practical parameter to determine quantitatively if a field is small and it can be expressed as a function of the photon beam quality specifier $TPR_{20,10}(10)$ according to\textsuperscript{10}

Equation (14)

$$r_{LCPE}(cm) = 8.369 \times TPR_{20,10}(10) - 4.382$$

Where $TPR_{20,10}(10)$ is the tissue phantom ratio in water at depth of 20 and 10 cm for a field size of 10 cm x 10 cm defined at the SDD of 100 cm.
When the beam quality specifier $\%dd(10,10)_X$ is used, and then $r_{LCPE}$ can be determined as

Equation (15)

$$r_{LCPE}(cm) = 77.97 \times 10^{-3} \times \%dd(10,10)_X - 4.112$$

Where $\%dd(10,10)_X$ is the photon component of the percent depth dose at 10 cm depth in water for a 10 cm x 10 cm field.

Equations (14) and (15) can be further simplified when beam transient charged particle equilibrium (TCPE) exists, i.e. where dose to water ($D_w$) and collision kerma ($K_{coll}$) are identical\textsuperscript{13}. It means the ratio of dose to water and collision kerma is a measure of the degree of CPE or TCPE. Given these approximations for clinical radiation therapy applications, the range of $r_{LCPE}$ values can then be represented in terms of $d_{max}$ of the beam for clinical physicist:\textsuperscript{14}

Equation (16)

$$r_{LCPE}(cm) \approx 0.67 \times d_{max} \pm 0.2$$

Where $d_{max}$ is the depth of maximum dose in cm for the 10 x 10 cm\textsuperscript{2} reference field.

The loss of LCPE and primary photon source occlusion effects are both responsible for a sharp drop in beam output with decreasing field size. This drop becomes more prominent when the photon beam energy increases or the density of the medium decreases because the electron range increases in both cases. The range of the lateral travel of electrons and its relation to LCPE becomes more pronounced in inhomogeneous media,
especially in a low-density medium such as lung.

Ionization chambers that have been the backbone of radiation therapy dosimetry due to their favorable characteristics (high sensitivity, long-term stability, reproducibility, and robustness). However, they are not always suitable, especially when high dose gradients, time-dose variance, and non-uniform beam distributions are encountered. Volume averaging and lack of electronic equilibrium, non-uniform beam intensity surrounding the detector, make the use of ionization chambers difficult in small field dosimetry, because the ionization chamber perturbs the particle fluence in the medium (figure 8). As a result, using the available perturbation factors of the existing dosimetry Codes of Practice (CoPs) or protocols such as IAEA TRS-398 and AAPM TG-51 are not sufficient. Despite of limitations such as low signal-to-noise ratio and higher polarity effects, small volume ionization chambers have been used recently for dosimetric measurements with substantial correction factors for very small fields as well.
Figure 9: An illustration of the volume averaging effect in one dimension. The solid curve represents a Gaussian curve approximating a small-field dose profile, dashed curve represents that measured by a detector of 5 mm if only volume averaging perturbs the fluence, and double arrow represents the dimension of the detector along the beam-scanning axis.\textsuperscript{10,16}

Solid-state detectors such as diodes, metal-oxide-semiconductor field-effect transistor (MOSFET) provide a better solution to the volume averaging effect. However, these dosimeters are not water-equivalent and exhibit a differential energy dependence with respect to water. Thus, their response with field size and possible effect of beam hardening
must be considered. In addition, the dose absorbed by the detectors sensitive volume depends on its density (usually higher density material as silicon in the sensitive volume perturbs the fluence in very small fields that cannot be ignored). Detectors of higher-density over-respond while lower density under-respond in very small fields. Correction factors are often required to convert the measured detectors’ readings to the absorbed dose.

A possible solution is to use intermediate field method (IFM)\textsuperscript{17}, sometimes referred as the daisy chain technique (cross-calibration of diode against the chamber in an intermediate field to connect measurements in large fields to small fields) to measure field output factor normalized to a 10 x 10 cm\textsuperscript{2} field\textsuperscript{10,18}. Although this technique accounts for the energy dependence of large fields, it does not consider the electron fluence perturbation due to the high density of silicon in small fields. It should be noted that even with the use of the intermediate field method, each of the detectors involved in very small fields must be corrected for the appropriate field output correction factors. Some studies have concluded that increasing the beam energy, increasing the air gap size, or decreasing the field size enhances the magnitude of the dose reduction\textsuperscript{16,19,20}. As no single detector ensures an accurate dose determination in the small fields, it is recommended to use at least two acceptable small field detectors with a careful output comparison. Recent studies included multiple small field detectors to evaluate dosimetric uncertainties as well as output correction factors for accurate and consistent clinical dosimetry in modern radiation therapy\textsuperscript{21–25}.

Advances in radiotherapy techniques have increased the use of small and non-standard fields in Stereotactic Radiosurgery/Body Radiotherapy (SRS/SBRT)\textsuperscript{26,27}, Intensity Modulated Radiation Therapy (IMRT)\textsuperscript{28} and Volumetric Arc Therapy.
In SRS/SBRT treatments, radiation dose is precisely delivered to the tumor in multiple non-coplanar and narrow beams using a combination of stereotactic apparatus. Such a highly conformal therapy makes use of imaging and localization devices for the accurate delivery of the treatment. Use of small radiation fields allows the dose to be delivered precisely in the target volume, sparing the healthy tissue. However, calculations of the dose delivered in small field’s dosimetry are more complicated than in standard treatments.

1.3.3 International Codes of Practice (CoPs)

Accurate and precise dosimetry in small fields posed significant challenge due to many factors including source occlusion, detector size limitation, and lack of lateral charge particle equilibrium.8 In 2008, the International Atomic Energy Agency (IAEA) in conjunction with the American Association of Physicists in Medicine (AAPM) released a letter proposing a new formalism for reference dosimetry of small and nonstandard fields.15 New and standard methodology was proposed to overcome such dosimetric challenges in small and non-standard fields that was not included in the existing Codes of Practice (CoPs) such as IAEA TRS-39830 and AAPM TG-51.31 Later, the updated version of TG-51 (addendum to TG51)32 for external beam radiation therapy and the International Commission on Radiation Units and Measurements (ICRU) Reports No. 9133 for small photon beam dosimetry were also available. The ICRU-91 mainly covered fundamentals of small-field dosimetry, treatment-planning algorithms, commissioning, quality assurance for the existing beam delivery systems, and role of image guidance during the delivery. It recommended a framework for prescribing, recording, and reporting stereotactic radiation therapy. According to the ICRU-91 report, use of radiation beam of energies above 10 MV
Alfonso et al 2008 proposed a new formalism for reference as well as relative dosimetry for machines that could not generate a conventional reference field, $f_{ref}$ of 10 cm x 10 cm at 100 cm SSD or SAD. It proposed a machine-specific reference field, $f_{msr}$ and provided alternative reference conditions for performing reference dosimetry in such field. The dimension of the machine-specific reference field should be equivalent to the conventional reference field and extend at least a distance of lateral charge particle equilibrium range ($r_{LCEP}$) away beyond the outer boundary of the reference class ionization chambers. Indeed, the full-width half maximum (FWHM) of the field must satisfy the equation (13). Recently, the IAEA-AAPM have jointly published a modified version of the existing Codes of Practice such as Technical Report Series No. 483 10,17 and TG-155 14 for the reference and relative dose determination to provide new guidelines for the reference and relative dosimetry in small and non-standard fields. These guidelines are the updated version of the existing CoPs such as Alfonso et al 2008 and ICRU-91 for the reference and relative dose determination specifically focused in small and non-standard photon fields’ dosimetry. The purpose of the recent CoPs is to describe the framework for applying detector-specific corrections and practical aspects of relative dosimetry of small fields. These reports provide guidance and recommendations on the various types of active and passive detectors as well as guidance on measuringdosimetric parameters such as PDD,
TPR/TMR, and OAR in small radiation fields.

According to the latest reports on reference dosimetry,\textsuperscript{10,14,15} the absorbed dose to water, $D_{w,Q_{msr}}^{f_{msr}}$, at the reference depth, in a beam of quality $Q_{msr}$ and machine-specific reference field $f_{msr}$, and in the absence of the ionization chamber is defined as

Equation (17)

$$D_{w,Q_{msr}}^{f_{msr}} = M_{Q_{msr}}^{f_{msr}} * N_{D_{w,Q_{o}}^0} * k_{Q,Q_{o}} * k_{f_{msr},f_{ref}}$$

Where $M_{Q_{msr}}^{f_{msr}}$ is the corrected dosimeter reading in the field $f_{msr}$, $N_{D_{w,Q_{o}}^0}$ is the calibration coefficient in terms of absorbed dose to water for an ionization chamber at a reference beam quality, $k_{Q,Q_{o}}$ is the beam quality correction factor, and $k_{f_{msr},f_{ref}}$ is a factor which corrects the differences between the conditions of field size, geometry, phantom material and beam quality of the conventional reference field, $f_{ref}$ and machine-specific reference field, $f_{msr}$. The factor $k_{Q_{msr},Q}^{f_{msr},f_{ref}}$ accounts for the difference of ion chamber responses in the’ $f_{ref}$ and $f_{msr}$ fields and is defined as

Equation (18)

$$k_{Q_{msr},Q}^{f_{msr},f_{ref}} = \frac{D_{w,Q_{msr}}^{f_{msr}}}{M_{Q_{msr}}^{f_{msr}}} \frac{M_{Q_{msr}}^{f_{msr}}}{D_{w,Q}^{f_{ref}}}$$

It can either be measured using a suitable detector and applying corrections obtained from a Monte Carlo simulation or calculated by the Monte Carlo simulation alone.
In the relative dosimetry, a field factor is introduced and the absorbed dose, \( D_{w, Q_{clin}}^{f_{clin}} \) at a reference point in a phantom for a clinical field, \( f_{clin} \) of quality \( Q_{clin} \) and in the absence of the chamber is defined as\(^{14,15,17} \)

Equation (19)

\[
D_{w, Q_{clin}}^{f_{clin}} = D_{w, Q_{msr}}^{f_{msr}} \times \Omega_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}}
\]

Where \( Q_{clin} \) is the beam quality of a clinical, non-reference field, \( f_{clin} \) and \( \Omega_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}} \) is a field output factor for the field \( f_{clin} \) with respect to \( f_{msr} \) field. In other words, the field output factor is given by \(^{14,15,17} \)

Equation (20)

\[
\Omega_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}} = \frac{D_{w, Q_{clin}}^{f_{clin}}}{D_{w, Q_{msr}}^{f_{msr}}} = \frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{msr}}^{f_{msr}}} \times k_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}}
\]

In fact, it is a ratio of the absorbed dose to water per monitor unit (MU) for the clinical field \( f_{clin} \) with beam quality \( Q_{clin} \) and the absorbed dose to water per MU in the machine-specific field \( f_{msr} \) with beam quality \( Q_{msr} \). It can either be obtained directly by the Monte Carlo simulation or be measured as a ratio of detector readings multiplied by the Monte Carlo calculated output correction factor, \( k_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}} \). In broad fields, the output factor (OF) is defined as ratio of detector readings in \( f_{clin} \) and \( f_{ref} \) or \( f_{msr} \), measured under reference conditions that is equivalent to the ratio of the corresponding absorbed doses due to the approximate constancy of stopping power and perturbation ratios with field size. However, the constancy argument is no longer valid for small fields due to detector-related
effects such as perturbation factors, energy dependence, volume averaging and the output factor is re-defined to field output factor (equation 20) as a true dose ratio along with a correction factor associated to the detector readings.\textsuperscript{12,17} Therefore, to distinguish between the conventional definition of output factor (ratio of detector readings) and the correct definition of output factor, TRS-483 CoP called it as field output factor. Moreover, the output (small field) correction factor that accounts for the differences between the detector response in the \( f_{\text{clin}} \) and \( f_{\text{msr}} \) fields is defined as

Equation (21)

\[
k_{f_{\text{clin}}, f_{\text{msr}}}^{Q_{\text{clin}}, Q_{\text{msr}}} = \frac{D_{f_{\text{clin}}}^{f_{\text{clin}}, Q_{\text{clin}}}}{D_{f_{\text{msr}}}^{f_{\text{msr}}, Q_{\text{msr}}}} \frac{M_{f_{\text{clin}}}^{Q_{\text{clin}}}}{M_{f_{\text{msr}}}^{Q_{\text{msr}}}}
\]

For the same nominal energy, the output correction factor mostly depends on the field size, expressed by the FWHM of the dose profile at the depth of measurement, and the type of detector. It is also called detector-specific correction factor that depends on many other factors such as the perturbation of particle fluence, volume-averaging effects on the field size, detector type, and the design, size, and non-water equivalency of most of the detectors. To derive output correction factors \( k_{f_{\text{clin}}, f_{\text{msr}}}^{Q_{\text{clin}}, Q_{\text{msr}}} \) for small fields from the data published in the literature, TRS-483 considered three types of datasets. One of these datasets considered the reference (ideal) detectors to be perturbation free except volume averaging. The major characteristics of these detectors are their near water equivalency, minimal fluence perturbation, and negligible energy dependence in the mega-voltage radiotherapy photon beams. Only few detectors such as plastic scintillators, microdiamond,
gafchromic films and some diodes fit into those categories that are suitable for small field dosimetry and have a near-unity output correction factor. Response of the detectors in small fields and determination of detector-specific output correction factors \( k_{Q_{\text{clin}}, Q_{\text{msr}}} \) have been extensively investigated by previous studies using one of the following three techniques: (1) empirical approach, where uncorrected signal ratios were determined and compared to the field output factors determined with reference conditions,\(^{19,22,25,34}\) (2) numerical approach, where \( k_{Q_{\text{clin}}, Q_{\text{msr}}} \) were determined by MC simulations,\(^{23,24,35-37}\) and (3) semi-empirical approach that combines both measurements and numerical/analytical calculations, where \( k_{Q_{\text{clin}}, Q_{\text{msr}}} \) were mostly determined through the comparison of measured uncorrected detector’s signal ratios with MC calculated field output factors.\(^{17,21,38}\)

According to the IAEA TRS-483, all field output factors are specified as a function of the field size expressed by the equivalent square small-field size \( S_{\text{clin}} \) which is determined by explicitly measuring the FWHM of the profiles both in-plane (X) and cross-plane (Y) directions at the depth of output measurements. For any rectangular small field, it is given by\(^{14}\)

Equation (22)

\[
S_{\text{clin}} = \sqrt{ \text{FWHM}(X) \times \text{FWHM}(Y)}
\]

It should be noted that this definition applies only in small fields whereas in broad fields, the equivalent square field is chosen such that it provides equal photon scatter contribution. The TRS-483 uncertainty estimates on the output correction factors apply 0.7 <
(FWHM(X)/FWHM(Y)) < 1.4. Outside this range, which is usually not violated except for the smallest equivalent-square small-field sizes (< 0.6 cm), a larger uncertainty on the correction factor should be considered.

Because a suitable detector for the entire range of field sizes from \( f_{msr} \) to \( f_{clin} \) may not be available. In this case, it is recommended to use an ion chamber down to an intermediate field, \( f_{int} \) (as small as possible but without small field conditions) and use of a suitable small field detector such as diode for smaller fields, thereby limiting the effect of diode’s energy dependence. This method is called an intermediate field method (IFM), and the final field output factor is defined as the product of those obtained from ion chamber (intermediate field) and small field detector (clinical field),

\[
\Omega_{Q_{clin}Q_{msr}}^{f_{clin} f_{msr}} = \left( \frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{int}}^{f_{int}}} \ast k_{Q_{clin}}^{f_{clin}, f_{int}} \right) SFD \ast \left( \frac{M_{Q_{int}}^{f_{int}}}{M_{Q_{msr}}^{f_{msr}}} \ast k_{Q_{int}}^{f_{int}, f_{msr}} \right) IC
\]

When small field conditions do not exist for the intermediate field, k-correction factor is approximately equal to unity for well-designed ionization chambers.

Major dosimetric parameters in relative/absolute dosimetry such as percent depth dose (PDD), tissue-maximum ratio (TMR), and off-axis ratio (OAR) are defined as

\[
PDD(z, f_{clin}, F) = \frac{D(z, f_{clin}, F)}{D(z_{max}, f_{clin}, F)}
\]

Equation (25)
\[ TMR(z, f_{\text{clin}}) = \frac{D(z, f_{\text{clin}})}{D(z_{\text{max}}, f_{\text{clin}})} \]

Equation (26)

\[ OAR(r, z, f_{\text{clin}}) = \frac{D(0, z, f_{\text{clin}})}{D(r, z, f_{\text{clin}})} \]

Where \( z \) is the depth, \( z_{\text{max}} \) is the depth of maximum dose, \( r \) is off-axis distance from the central axis of the beam, \( F \) is the source to surface distance, and \( f_{\text{clin}} \) is the clinical field. Basically, these dosimetric parameters are the ratios of two doses at different conditions and are measured routinely for every machine for a wide range of field sizes.\textsuperscript{25,39,40}

1.4 Virtual Cone Technique

For many years, specialized machines such as GammaKnife, CyberKnife, TomoTherapy, and other cone-based LINACs have been considered the standard machines to treat many functional abnormalities (brain, lung tumors) and small targets such as trigeminal and glossopharyngeal neuralgias. The trigeminal neuralgia is a condition characterized by pain coming from the trigeminal nerve, which starts near the top of the ear and splits into three; ophthalmic branch (toward the eye), maxillary branch (toward the cheek), and mandibular branch (toward the jaw) as shown in figure 10. We have two trigeminal nerves for each side of our face, but trigeminal neuralgia pain most commonly affects only one side. To treat such a small (\( \leq 0.05 \text{ cm}^3 \)) target and very close (\( \leq 0.5 \text{ cm} \)) to the critical organ (brainstem), the linac must deliver a highly focused and precise beam of radiation dose in a single fraction. Although the cone-based treatment remains the standard for such small targets (trigeminal neuralgia), it poses several challenges such as need of cone-specific dose calculation, rigorous geometrical quality assurance, and use of limited fields/arcs.\textsuperscript{41,42}
With the advancement of the refined multi-leaf collimator (MLC) technology such as high-definition MLC (leaf width, 2.5–3.0 mm) and robust treatment planning algorithms, the dose distributions or gradients from the LINAC can be collimated to match those obtained using physical cones. Recently, there is ongoing research on MLC-based SRS for treatment of relatively smaller lesions such as trigeminal neuralgia instead of using physical cones. It should be noted that the well-defined spherical or ellipsoidal dose distribution along with a high mechanical precision in dose delivery is required during the treatment. To achieve this, it is crucial to have a highly advanced linac with a sub-millimeter accuracy and correctly defined output, penumbra, and modeling for such a small field/beam during the treatment planning.

**Figure 10:** A representation of a trigeminal nerve which splits into three branches; ophthalmic, maxillary, and mandibular zones. The trigeminal neuralgia is the pain coming from the trigeminal nerve. (https://www.hopkinsmedicine.org/health/conditions-and-diseases/trigeminal-neuralgia)
First MLC-based radiosurgery for trigeminal neuralgia was reported and validated on the Varian Truebeam STx linac\textsuperscript{43} in which the 6 MV FFF beam was delivered by 18–21 static conformal fields. Next, a virtual cone (VC) technique of delivering small, spherical dose distribution using a standardized MLC field without physical cones was introduced using the Varian EDGE linac (Varian Medical Systems, CA).\textsuperscript{43,44} In this technique, the standardized MLC fields (non-square fields) together with multiple non-coplanar arcs were used to generate a very small, spherical dose distribution comparable to that produced by a 4–5 mm physical cone. They employed 10 MV FFF photon beams because of their high dose rates allowing for steep dose gradients, low collimator scatter, energy spectrum, and ability for SRS treatments to be delivered to each patient with one fraction. However, the use of very small MLC field with variable field sizes could add further dosimetric challenges during such advanced treatment. We propose the virtual cone technique with the fixed field geometry, called fixed virtual cone (fVC) technique to generate a very small, spherical dose distribution equivalent to 5 mm cone. It is characterized by 0.5 cm x 0.5 cm HDMLC field of 10 MV FFF beam with a fixed jaws setting at 100 cm SAD in conjunction with multiple non-coplanar arcs. The geometrical details are described in the next section.

1.5 Monte Carlo Simulations in Radiation Therapy

Monte Carlo (MC) is a stochastic method of numerical integration in which statistical sampling is used to approximate the outcome of probabilistic events by using random number generation.\textsuperscript{46} The MC method is used in radiation transport to simulate the possible interactions between incident radiation and matter by tracking the random trajectories of
individual particles over time to obtain the average physical quantity. In photon dosimetry, Monte Carlo simulation is used as a highly accurate method in situations where physical measurements are not possible. Radiation transport mainly includes photon and electron interactions. The photon interactions include Rayleigh scattering, Compton scattering, Photoelectric effect, and Pair production. Electron interactions include elastic and inelastic collisions. The probability of such interactions depends on parameters such as particle type, particle energy and the atomic composition of the materials.

Furthermore, the results of these interactions are defined by probability distributions. Given the multitude of interactions within a linear accelerator, and then within the patient or phantom, it is apparent that a simple solution for defining the dose deposited is not easy task. Monte Carlo simulation provides a solution by using random numbers to sample probability distributions of each radiation interaction of particles on step-by-step basis. Once, an extremely large number of particles (usually > 1 million) have been simulated, an overall representation of the radiation beam is built up. The statistical uncertainty often stated alongside Monte Carlo simulation results, usually refers to the overall sampling uncertainty of distributions in the beam model. The statistical uncertainty is proportional to the square root of the number of particles simulated. The MC simulations of an infinitely large number of particles will reflect the most realistic representation because of the convergence of sample mean to true mean according to central limit theorem. The efficiency ($\varepsilon$) of a simulation is defined by

$$\varepsilon = \frac{1}{s^2 * T}$$
Where \( s^2 \) is an estimate of the true variance (\( \sigma^2 \)) of the quantity of interest and \( T \) is the simulation/computation time that is proportional to number of particles (\( N \)) used in the simulation.

**Figure 11:** A simple representation of particle tracks during the MC simulations in a box.\(^{32}\)

The use of the Monte Carlo method for calculations of dosimetric parameters in radiotherapy has become the most efficient and consistent tool for simulations of basic dosimetric quantities, like stopping-power ratios and perturbation correction factors for reference ionization chamber as well as for fully realistic simulations of clinical accelerators, detectors and patient treatment planning.\(^{46,53}\) Its accurate use requires consistency in the data throughout the entire dosimetry chain, and the recent updates of key dosimetric data provided by the ICRU Report 90 are necessary in reference dosimetry. Although data consistency is probably less critical for treatment planning, their
implementation also in this field is well advised. As there are number of other issues involved in MC simulations, although less significant, it should be noted that MC calculation should not be considered as free of errors. This is particularly important regarding applications in MC treatment planning, where the uncertainties involved remain “uncertain”, a general problem that is also applicable to other methods and algorithms used in different types of treatment planning systems. The most common MC codes available for radiation therapy applications are EGSnrc, Geant4, Penelope, MCNPX. However, the EGSnrc is the most widely accepted/gold standard MC package in radiation therapy. The GESnrc is the general package for MC simulations of coupled electron and photon transport that employs the condensed history (CH) technique. In this technique, large number of subsequent transport and processes are condensed to a single step. The cumulative effect of individual interactions is considered by sampling the change in particle’s energy, direction of motion, and position, at the end of the step from appropriate multiple scattering distributions. The CH technique, motivated by the fact that single collision with the atom mostly causes only minor changes in the particle’s energy and direction, made the MC simulations of charged particles possible but introduced an artificial parameter called a step-length.
2. Materials and Methods

2.1 Experimental Methods

All experimental measurements were performed on the Varian Edge™ radiosurgery system (Varian Medical Systems, Palo Alto, CA) at South Florida Proton Therapy Institute, Delray Beach, Florida as shown in figure 12. The Varian EDGE linac equipped with a high definition multi-leaf collimator (HD120 MLC) which is capable of delivering numerous non-coplanar, smaller, and intensity modulated photon beams of energies 6 MV and 10 MV with sub-millimeter accuracy. The HD120 MLC consists of two banks of tungsten leaves, with 60 leaves on each bank. It has two different types of leaves: 32 central leaves of 2.5 mm width (quarter leaves) and 28 outer leaves of 5 mm width (half leaves) at isocenter. The leaf ends are rounded, and leaves are 6.9 cm thick. The EDGE linac system is utilizing the Eclipse treatment planning system (TPS) with an anisotropic analytical algorithm (AAA) and a deterministic Acuros XB dose calculation algorithm for the treatment planning. The comprehensive dosimetric evaluation of both algorithms were reported in the literature. The Acuros XB dose calculation algorithm (version 15.6/16.0) was used for dose calculations using a grid size of 1 mm x 1 mm x 1 mm.
Dosimetric measurements for each energy beam, i.e., 6 MV and 10 MV flattening filter-free (FFF) beams, were conducted at two clinical settings: (1) at source to surface distance (SSD) of 90 cm, 10 cm depth, and (2) at SSD of 95 cm, 5 cm depth in the water phantom. The measurement geometry consisted of an isocentric set up with a source to axis distance (SAD) of 100 cm as shown in figure 13. Two types of water phantoms: a standard scanning water tank and a PTW MP3-M water phantom were used for the dosimetric measurements.
Major dosimetric (relative) parameters such as percent depth dose (PDD), tissue-maximum ratio (TMR), and off-axis ratio (OAR), output factors (OF) were determined. Various reference detectors (micro-chambers, diodes, EBT films) were used as part of beam commissioning and machine-specific quality assurance (QA) prior to measurements. To account calculate PDDs at different SSDs and depths settings for 10 cm x 10 cm reference field, the commissioned/published data was utilized and the change in PDD due to the change in SSD was calculated using Mayneord’s formula. It implies the PDD will increase with the increase of SSD and vice versa for the beam of same nominal energy, field size and depth of the measurement. However, this method has some limitations such as (1) the error will be significant for a large SSD, field size and depth, and (2) the error will be significantly high for a lower energy radiation due to more scattering. But it is more applicable/accurate for the small fields as well as relatively higher energy photon beams due to minimal photon scattering.
Figure 13: An experimental set up for the output factors measurements using ion chamber and diode at 100 cm SAD in the water phantom: (a) SSD = 90 cm, d = 10 cm (left) and (b) SSD = 95 cm, d = 5 cm (right).

In this study, multiple small-fields detectors: a small volume ion chamber (PTW 31021, semi flex 3D ion chamber), a small field diode (PTW 60018, SRS diode), SRS MapCHECK® (Sun Nuclear Corporation, FL), and EBT3 (Ashland Inc., NJ) film were selected based on the most recent guidelines\cite{10,14} for selecting suitable detectors for small-field outputs measurements. The semi-flex 3D ion chamber is a waterproof, cylindrical ionization chamber with a sensitive volume of 0.07 cm$^3$ which provides high spatial resolution for the fields $\geq$ 3 cm x 3 cm.\cite{63} The SRS diode (PTW 60018) is also a waterproof,
p-type silicon diode with a sensitive volume of 0.3 mm$^3$ which provides higher dose response and better spatial resolution for 10 x 10–1 x 1 cm$^2$ fields.$^{63}$ The SRS MapCHECK consist of 1013 n-type solid-state diodes (SunPoint$^\circledR$ 2 diode with a sensitive area of 0.48 x 0.48 mm$^2$) arranged on a 77 mm x 77 mm array with inter-detector spacing of 2.47 mm.$^{64}$ Various static MLC fields from 10 cm x 10 cm reference field to 0.5 cm x 0.5 cm smallest field (at 100 cm SAD) of both 6 MV FFF and 10 MV FFF photon beams were considered for dose outputs measurements. For each point of measurement, 100 MU was delivered and at least three measurements were taken unless otherwise specified.

During dose/output measurements, ion chambers were mounted in the water tank such that their axis of symmetry was perpendicular to the beam axis. They were aligned to water surface very carefully across the entire scanned length (in-plane and cross-plane directions). While measuring with the diodes, they were mounted in the water tank such that their axis of symmetry was either parallel (PTW diode) or perpendicular (SN Edge diode) to the beam axis. The lateral alignment along the beam central axis was done in three ways:$^{17,62}$ (1) initial setup using room laser, (2) re-adjustment after acquiring beam profiles along in-plane and cross-plane directions, and (3) by moving the diode manually by 0.2 mm steps along in-plane and cross-plane directions until the maximum reading was found. The position of optimal diode reading was considered as the central beam axis and final position of detector. The process of lateral alignment mentioned above was done separately for each energy. When the diode needed to be moved to a new depth, the process was repeated to verify the diode was centered in the field.
2.1.1 Reference and Relative Dosimetry

Reference and relative dosimetric measurements were performed using the detectors PTW 31021, Semi flex 3D Ion Chamber, PTW 60018, SRS diode, and SRS MapCheck® (Sun Nuclear Corporation, Melbourne, FL) with 6 MV FFF and 10 MV FFF photon beams in water phantom. The dosimetric measurements were performed according to the most recent detectors guidelines published on the joint IAEA-AAPM CoPs 15,17 for the reference and relative dosimetry.

Assuming the figure 13 was to the proper scale; \( f_1 \) would be the SSD of 90 cm with a depth of 10 cm, totaling the SDD (SAD) of 100 cm and \( f_2 \) would be SSD equaling 95 cm with a depth of 5 cm, equaling the SDD of 100 cm respectively. The dosimetric measurements at the two different depths were used to compare the TPS calculated doses and outputs for the volumetric change in treatment depth. The cases considered for this study were treated with static field arc therapy with the treatment PTV being at isocenter. The SSD was changing throughout the arc therapy due to the elliptical cranial shape. However, the SAD was held at isocenter requiring the machine quality assurance be completed to verify the output factors were within acceptable limits. While measuring using the diode, it was centered in the field when the optimal measurement was found. The diode was centered with the field light, and then moved to each depth. The optimum measurement was found by moving the diode in the x-y plane until the maximum reading was found. The maximal reading was found in the x-plane first, and then while keeping the x-plane stationary, the diode was moved in the y-plane until the ultimate reading was found as described in the literature.21,25 When the diode needed to be moved to a new depth, the process was repeated to verify the diode was centered in the field. The ion chamber was
centered using the field light.

The output for the two nominal energies was measured indirectly with the two detectors. Instead of calculating the dose outputs for reference conditions using the AAPM TG-51\textsuperscript{65} and its addendum,\textsuperscript{32} the depth doses at 5 cm and 10 cm were calculated using the clinical commissioning data for the Edge™ radiosurgery system\textsuperscript{25,55} and the QA doses calculated using the Eclipse TPS with both AAA and Acuros XB.\textsuperscript{59,61} The commissioned data was used to calculate the depth dose at reference level. The percentage depth dose (PDD) of the commissioned data was used to calculate the reference conditions for the output of the linac at 100 cm SSD for the 10 cm x 10 cm reference field. The reference \( PDD_1 \) was first calculated using the commissioned data. Due to a change in the source-to-surface distance, a new \( PDD_2 \), was calculated using the Mayneord’s formula to calculate the change of SSD from the commissioned SSD of 100 cm, to the QA SSD of 95 cm or 90 cm as\textsuperscript{4}

Equation (28)

\[ PDD_2 = MF \ast PDD_1 \]

Where \( MF \) is the Mayneord F factor, which is defined as

Equation (29)

\[ MF = \left( \frac{f_2 + d_{max}}{f_1 + d_{max}} \right)^2 \ast \left( \frac{f_1 + d}{f_2 + d} \right)^2 \]

Where \( f_2 \) is the commissioned SSD, \( f_i \) is the new SSD (90 cm/95 cm), \( d_{max} \) is the maximum dose distance in the medium, and \( d \) is the distance to the measurement in the medium.

Clearly, \( MF > 1 \) if \( f_2 > f_i \) and \( MF < 1 \), if \( f_2 < f_i \) for all \( d > d_{max} \). It means PDD will
increase with the increase of SSD and vice versa for the beam of same nominal energy, field size and depth of the measurement. However, this method has some limitations such as (1) the error will be significant for a large SSD, FS, and depth, and (2) the error will be significantly high for a lower energy radiation due to more scattering. Therefore, it is more applicable/accurate for the small fields as well as relatively higher energy photon beams because minimum scattering occurs. The shortening of the total distance requires the inverse square law to correct this factor to calculate the dose of the reference field. The ratio of monitor units was used to calculate for the change of monitor units from 100 MU to 200 MU, where 200 MU was the monitor units throughout the machine specific quality assurance. The dose for 200 MU was calculated by the product of the new PDD\textsubscript{2} with the inverse square law and the ratio of commissioned MU’s and new MU’s to be delivered. This new dose is the calculated dose for a reference 10 cm x 10 cm beam at the corresponding SSDs of 90 cm and 95 cm and depths of 10 cm and 5 cm, respectively. In other words, the EDGE linac was defined at 100 cm SSD such that it delivers 1 cGy/MU at depth of maximum dose (dmax) and it was calibrated at 95 SSD, 5 cm depth (100 cm SAD) such that it delivers 0.925 cGy/MU and 0.992 cGy/MU for 6 MV and 10 MV FFF beams respectively. For relative dosimetry, apart from output factor (OF) or field output factor, other parameters such as percent depth dose (PDD), tissue-maximum ratio (TMR), and off-axis ratio (OAR) defined by equations (24), (25) and (26) were also determined.

2.1.2 Output Factors

With the reference dose calculated from the 10 cm x 10 cm field, the dose for each clinical field could be calculated from corrected output factors obtained from measurements of 6 MV and 10 MV FFF photon beams at SSDs of 90 cm, 95 cm and depths of 10 cm, 5 cm
respectively. Because a suitable detector for the entire range of field sizes from \( f_{msr} \) to \( f_{clin} \) may not be available. In this case, it is recommended to use an ion chamber down to an intermediate field, \( f_{int} \) (as small as possible but without small field conditions) and use of a suitable small field detector such as diode for smaller fields, thereby limiting the effect of diode’s energy dependence. This method is called an intermediate field method (IFM), and the final field output factor is defined as the product of those obtained from ion chamber and small field detector defined by equation (23). An Accredited Dosimetry Calibration Laboratory (ADCL) calibrated ion chamber measured readings from the reference field size, specified in CoP, down to the intermediate field size. This method cross calibrates a detector equipped to measure clinical small fields, to a detector equipped to measure a machine specific reference field (\( f_{msr} \)). For the Varian Edge\textsuperscript{TM} linac, the reference field of 10 cm x 10 cm at 100 cm SSD can be achieved and there is no need to define the machine-specific reference field. The clinical field output factor requires two output factors, one from the clinical field detector and one from the intermediate detector that reaches the reference conditions of a 10 cm x 10 cm field. The clinical output ratio was found by dividing the measured reading with the diode for the clinical field by the measured reading with the diode for the intermediate 4 cm x 4 cm field. The intermediate field of 4 cm x 4 cm was used after the calculation of the range of charge particle to create lateral charged particle equilibrium. The lateral range of charged particle equilibrium \( r_{LCPE} \) can be calculated by using equations (15) or (16), where the \( TPR_{20,10}(10) \) was calculated by

Equation (30)

\[
TPR_{20,10}(10) = 1.266 \times PDD_{20,10}(10) - 0.0595
\]
Where $PDD_{20,10}(10)$ is the percentage depth dose at 20g/cm² and 10g/cm² at an SDD of 100 cm where the field size is defined as a 10 cm x 10 cm field. The range of the lateral charged particle equilibrium calculated were 0.907 cm and 1.543 cm for the 6 MV FFF and 10 MV FFF beams, respectively.

Particularly, the micro-ion chamber (PTW 31021 Semiflex 3D chamber) with an active volume of 0.07 cm², was used to measure output factors for the reference field and intermediate fields (10 x 10–3 x 3 cm²), an SRS diode (PTW 60018) was used for intermediate to smaller fields (5 x 5–0.5 x 0.5 cm²), and SRS MapCheck® was used for all patient-specific fields. Each measurement was conducted at two different clinical settings: (1) at source to surface distance (SSD) of 90 cm, 10 cm depth, and (2) at SSD of 95 cm, 5 cm depth for both beams. The 3D axis was controlled remotely with a PTW MP3-M Water Phantom System. The SSD was measured using a mounted SSD ruler to minimize any positioning errors of the surface of the water. The SSD ruler was verified to be at the proper SSD distance when it broke the water’s surface tension and could be visually perturbing the inside of the surface tension and in line with the surface of the water. The dosimetric measurements were measured at source-detector-distance (SDD) of 100 cm with SSDs of 90 cm and 95 cm for the depths of 5 cm and 10 cm respectively using the ion chamber and diode as shown in figure 13.

The uncorrected clinical field size output factor was calculated using the intermediate field method between the reference measurement and the measurements for the clinical field. The corrected output factor for the clinical field was calculated using the reference dose from the commissioned calculated data, output factor without corrections, and k-factor correction using equation (18). The k- factors were linearly extrapolated from
TRS-483, based on the $S_{\text{clin}}$ and the FWHM of the corresponding field as reported in the literature\textsuperscript{17} because the $TPR_{20,10}(10)$ for 10 MV FFF and 6 MV FFF calculated were the same as they reported. The clinical field dose was calculated using the reference field dose at that depth and the corrected output factor using equation (19). The requirement for a detector to be suitable for the small field is given by equation (13) and was used to calculate the corresponding FWHM, minimum clinical field size, for the ion chamber (PTW SemiFlex 31021) using 0.55 cm as the maximum distance of the outer boundary of the detector. This outer boundary of the detector is published by PTW to be 0.55 cm for the width corresponding to the width parallel to the stem. A width perpendicular to the stem must be noted that the outer boundary diameter is smaller and, therefore, may have orientation influences for small fields when calculating the minimum FWHM. The FWHM calculated was 2.36 cm and 3.64 cm for the 6 MV FFF and 10 MV FFF beams, respectively. Therefore, the minimum field size to complete reading measurements was a 4 cm x 4 cm collimator defined field. The ion chamber and diode output ratios are multiplied by their ratios of beam quality factors to find the output factor. The product of each output ratio gave the final clinical output factor.

2.1.3 Fixed Virtual Cone Technique

In this study, we started with a retrospective measurement (SRS MapCHECK) of 30 clinical cases treated with a small field ($S_{\text{clin}} \leq 1.0$ cm) arc therapy for intracranial lesions to investigate possible dosimetric changes caused by the gravity or low maintenance of MLC during a rotation of the gantry. The retrospective measurements of treatment plans consisting of various arc fields, different collimator angles, and couch positions were
conducted using both 6 MV and 10 MV FFF photon beams. The difference (between the TPS plans and measurements) in dosimetric parameters such as field output factors, dose/MU were compared between the 6 MV and 10 MV FFF beams at 100 cm SAD.

**Figure 14:** A flowchart of generating a small spherical dose distribution comparable to 5 mm (diameter) physical cone using the fixed virtual cone (fVC) technique with 10 MV FFF beam. The fVC is characterized by 0.5 cm x 0.5 cm high-definition (HD) MLC field defined by two pairs of central leaves with a fixed (1.5 cm x 1.5 cm) jaws position at 100 cm SAD.

Next, we adopted a “fixed virtual cone (fVC) technique” to generate a very small, spherical dose distribution comparable to that produced by physical cones, by using the smallest (square) MLC field of 10 MV FFF beam in conjunction with multiple non-coplanar arcs as shown in figure 14. The fixed field geometry with the smallest MLC square field was used rather than different non-square (variable) fields with anticipation of safer and more reliable dose delivery during such advanced treatment. The beam geometry of the fVC method in three-dimensional (3D) view is as shown in figure 15. The fixed virtual cone
was characterized by a 0.5 cm x 0.5 cm high-definition MLC field, defined by 2 central leaves along with a fixed opening (1.5 cm x 1.5 cm) of jaws defined at 100 cm SAD. Two central leaves of the MLC were set to create a very small field and the remaining leaves were positioned 3 cm outside of the secondary collimation, which was 1.5 cm x 1.5 cm defined at 100 cm SAD. A very small, spherical dose distribution comparable to 5 mm physical cone, was produced by using 10–14 non-coplanar arcs during the treatment planning. The 10 MV FFF photon beam was used because of their high dose rates allowing for steep dose gradients, low collimator scatter, energy spectrum, and ability for SRS treatments to be delivered to each patient with one fraction. At each couch position, multiple non-coplanar arcs with a clockwise and counter-clockwise gantry rotation (gantry angle, 0–360°, arc width, 30–90°) were used with different collimator angles (0–90 degrees). The dose per degree of the arcs was non-uniform and proportional to the sine of the gantry angle. The weighting of the arcs was chosen in such a way that in the limit of large N and 2n irradiation, it results in uniform fluence per unit area that is ideal for generating a spherical dose distribution.

The treatment plans of 10 clinical cases (left, right trigeminal neuralgia with a prescribed dose 7000–8000 cGy per fraction) treated with the fixed virtual cone technique were selected for verification process or quality assurance and re-planned with 200 MU for 6 MV FFF (with 1400 MU/min dose rate) and 10 MV FFF (with 1600 MU/min dose rate) photon beams. The verification process consisted of the gamma evaluation of two-dimensional dose profiles (planar dose maps) through isocenter using a global 3%/1 mm criteria (2%/1 mm for the whole plan) and the comparison of dose outputs or dose/MU between the TPS and measurements (SRS MapCHECK) at 100 cm SAD using both beams.
and spatial resolutions (0.1 cm, 0.125 cm). The dose parameters were compared on field-
by-field basis plus as a whole. The field-by-field study included the comparison of dose
outputs (at central axis) for 130 fixed virtual cone fields with different arcs (width 45–90°),
collimator angles (0–90°) and couch positions. It’s noted that SSD was changing
throughout the measurements (arc therapy) due to the elliptical cranial shape thereby
keeping the same SAD of 100 cm. The collimator was rotated for differing fields prior to
irradiation of the arcs to deliver uniform dose and reduce MLC transmission. Due to the
leaf ends being rounded shape, the rounded leaf edges were moved outside the jaw settings
to Bank A where the jaw collimators shielded radiation for the MLC. The leaf edges were
moved outside of the jaw settings to minimize leaf transmission through the abutted leaf
ends.
**Figure 15:** A three-dimensional view of a treatment plan to treat a very small tumor in left trigeminal nerve using a fixed virtual cone technique. The fixed virtual cone is characterized by 0.5 cm x 0.5 cm high-definition (HD) MLC field, defined by two pairs of central leaves of the HDMLC.

The fixed virtual cone beam model was validated using both AAA and Acuros XB algorithms as well as corresponding measurements using the SRS diode, SRS MapCHECK and EBT3 films. The dosimetric accuracy (end-to-end test) of the final treatment plan was verified with SRS MapCHECK measurement using StereoPHAN, as well as EBT3 film dosimetry using an anthropomorphic skull phantom (CIRS Inc., Norfolk, VA). A unique CT and MR protocols was applied during image acquisitions to meet the requirement of
high precision dose delivery thereby optimizing the image quality of tumor site. Retrospectively, 15 clinical cases treated for trigeminal neuralgia (prescribed dose, 70–80 Gy) using the fixed virtual cone method (10 MV FFF beam) were collected and important clinical parameters such as number of arcs, mean and maximum doses to the target, maximum dose at critical organ (brainstem), irradiated area/volume enclosed by 90%, 50%, and 25% isodose lines were recorded from treatment plans. Similarly, the irradiated areas encompassed by 90%, 50% and 25% isodose curves in addition to gamma indices (passing rate, criteria) were manually acquired from the SRS MapCHECK QA using the SNC patient software tool. Dimensions of isodose curve (ellipsoidal) was defined by major and minor diameters (i.e., \(d_1\) and \(d_2\)) and corresponding diameter (d) of equivalent circle or sphere in a given volume was calculated as \(d = \sqrt{d_1 \times d_2}\). The equivalent diameters were compared between the TPS and measurements for further evaluation.

Furthermore, gravitational effects to the linac head may cause uncertainties due to the unwanted movement of the collimator leaves in the MLC during the SRS arc therapy. There is a strong possibility of somewhat leaf drift due to the gravity when the gantry is at 90° or 270° and the collimator is at 90° specifically when the MLC was not well maintained. When the MLC leaves are mostly impacted by the gravity, it could result significant dosimetric uncertainties specifically in SRS arc treatments using very small fields.\(^9,14,44,66,67\) Such dosimetric inconsistencies should be properly evaluated and applied proper corrections before such a high precision treatments to optimize dose accuracy.\(^45,68\)

The fixed virtual cone QA plans (at 0° gantry angle and 90° collimator angle) including virtual cone fields defined by an infinitesimal leaf shift of \(\pm (0.5–1.0)\) mm were generated for both beams and clinical settings. Corresponding measurements of output factors using
the SRS diode were performed to quantify the dosimetric uncertainties produced by such leaf shift. For instance, for each energy beam and clinical setting, four different virtual cone plans (with a leaf shift of ± 0.5 mm and ± 1.0 mm) were considered for dosimetric measurements. The field output factors were determined by using intermediate field method, given by equation (23). The main reason of considering such virtual cone fields is to evaluate dosimetric uncertainties associated with possible fluctuations in MLC fields during the small field SRS arc treatment using the fVC method.

**Figure 16:** A beam eye view for the virtual cones: (a) VC_0.5 x 0.5_1I, (b) VC_0.5 x 0.5_1O, (c) VC_0.5 x 0.5_2I, (d) VC_0.5 x 0.5_2O for 10 MV FFF beam defined at 95 cm SSD, 5 cm depth in water phantom using the Eclipse TPS. These virtual cone fields were created to evaluate dosimetric uncertainties caused by a diminutive shift on leaves.
The statistical data analysis for the comparison of dosimetric parameters (D) between the calculations (TPS) and measurements (SRS diode) was performed by: 69,70

Equation (31)

\[
\text{Difference in } D \ (\%) = \frac{D_E - D_T}{D_T} \times 100
\]

Where \( D_E \) and \( D_T \) are the expected (calculated) and true (measured) parameters respectively. A positive value means the expected value is greater than the true value and vice versa.

2.2 Monte Carlo (MC) Simulations

A Monte Carlo model was designed by using EGSnrc MC Package 52 for the Varian EDGE linac equipped with a high-definition multi-leaf collimator (HD120 MLC) which has an ability to deliver both 6 MV and 10 MV FFF photon beams. Under the EGSnrc, the BEAMnrc 71 was used to model linac head including jaws (Y, X) and HD120MLC by utilizing the phase-space files above the Y jaw that provided by the vendor (Varian Medical Systems, Palo Alto, CA). The phase-space file contains the data related to particle energy, position, direction, charge, etc. for every particle crossing a scoring plane. These phase-space files (version-2) were stored at 6.5 x 6.5 cm² two-dimensional plane defined at 26.7 cm downstream from the source. Recently, those phase-space files were validated with measurements for photon 72 and electron 73 beams, respectively. Two pairs of metallic Jaws (Y, X), one pair in the cross-plane and the other pair in the in-plane, were used to shape the beam above the MLC. The HD120 MLC consists of two banks of tungsten leaves, with 60 leaves on each bank. It has two different types of leaves: 32 central leaves of 2.5 mm
width (quarter leaves) and 28 outer leaves of 5 mm width (half leaves) at isocenter. The leaf thickness is 6.9 cm with a rounded shape leaf end. The schematic diagram of the Varian EDGE linac used in the simulations is as shown in figure 17. The HD120 MLC was created based on the details provided by the Vendor (Varian Medical Systems, Palo Alto, CA) by using SYNCHDMLC module under the BEAMnrc as shown in figure 18. A cross-sectional view of the MC simulated linac for the 10 cm x 10 cm field size of the 6 MV photon beam is as shown in figure 19. The PHSP A and PHSP B represent the phase-space files scoring planes just below the jaws and MLC respectively, defined in the beam simulations.
**Figure 17:** A schematic diagram of the Varian EDGE Linac (Varian Medical Systems, Palo Alto, CA) used in MC simulations.
Figure 18: A schematic representation of the HD120 MLC geometry. This MLC has two banks of tungsten-based alloy leaves, each bank with 60 leaves: 32 central (quarter) leaves of width 2.5 mm and 28 outer (half) leaves of width 5 mm at isocenter.\textsuperscript{57}

The leaves openings defined in terms of XY co-ordinates at mid-plane of MLC (50.5 cm down from the target). The transformation of the coordinates for MLC openings between MC and treatment plans was defined as\textsuperscript{74}

Equation (32)

\[ X_{SCD} \text{ (MC plan) } = \left( \frac{SCD}{SAD} \right) * X_{SAD} \text{ (Treatment plan)} \]

Equation (33)

\[ Y_{SCD} \text{ (MC plan) } = \left( \frac{SCD}{SAD} \right) * Y_{SAD} \text{ (Treatment plan)} \times n \]
Where SCD (50.5 cm), source to center (mid-plane of MLC) distance, SAD (100.0 cm), source to axis distance, and n, number of leaves opened to define a field.

The detailed explanation of MC simulations was presented below. After the beam model of a linac was designed, the media data files 700icru and 500icru, which contain the physical properties such as the mass density, electron density and the atomic number of the materials for each component, were fed into the model. If these files do not contain the material you are using for any component, a separate data file can be created. The users interact with the program through the input file (like a source code) for MC simulations, i.e., mainly two files (input files and cross-section data) are required for the entire simulations. The input file consists of five major sets of parameter files those act as instructions for the BEAMnrc to perform the entire simulation. They are MC control inputs, source geometrical configuration, particle transport control, component modules geometrical configuration, and default EGS parameters. The MC control parameters include the number of initial particles in the simulation, the random number seed, and the maximum simulation time. The source geometrical configuration parameters specify the properties of the original particles in the simulation. One can choose whether the original particles are photons, electrons or positrons, the energy they carry, their angle of incidence relative to the first component module and their fluence distribution. It also allows to use a phase-space file as an incident source. The phase-space file contains the data related to particle energy, position, direction, charge, etc. for every particle crossing a scoring plane. The scoring of phase-space files increases the simulation efficiency, but it requires large space for storage. The user is allowed to choose the plane from which the properties of the particle will be recorded, and the information will be stored in a phase space file.
Figure 19: A cross-sectional view of the 10 x 10 cm² field defined by jaws (Y, X) and HDMLC generated by the BEAMnrc. The PHSP A and PHSP B represent the planes for phase-space files scoring just below the jaws and MLC respectively.

The particle transport parameters describe how a particle behaves in a medium. They also determine when a particle will be eliminated in the phantom which includes the parameters: ECUTIN, PCUTIN, ESAVE_GLOBAL and IREJCT_GLOBAL. These parameters were implemented into the program to save the simulation time by eliminating those particles that have no significant contribution to the dose in the region of interest. The ECUTIN and PCUTIN are the energies below which an electron and a photon will be discarded in the simulation. ESAVE_GLOBAL is the maximum energy at which an electron will be
considered for range rejection as explained below. The process of range rejection is controlled by a parameter called IREJCT_GLOBAL. If the particle energy falls below ESAVE_GLOBAL, then IREJCT_GLOBAL will calculate the range of the electron. If the range is not large enough so that the electron can cross the next boundary then it will be discarded, and all its energy will be assigned to the absorbed energy in the surrounding region. The simulation of the particle stopped when its energy reached a threshold specified by the variance reduction parameter or when it exits the final component modules. These parameters also contribute to reduce statistical uncertainty, called variance reduction techniques.

The Direct Bremsstrahlung Splitting (DBS), which is the most efficient variance reduction technique, was used for the simulations (Kawrakow, Rogers et al. 2004). In this case, the bremsstrahlung photon will be split NBRSPL times as defined by the user and algorithm will then determine whether the split photons are aiming inside or outside a circular splitting field at certain distance away from the source. This splitting field is perpendicular to the beam direction and has a radius defined by a parameter called FS. The distance between this splitting field and the source is defined by a parameter called SSD. For those photons that are propagating inside the splitting field, the weighting factor is reduced by a factor of 1/NBRSPL, and the same particle transport algorithms are applied. Photons that are propagating outside the splitting field will be subjected to a selection process called Russian Roulette. The Russian Roulette process compares the survival threshold (1/NBRSPL) of each photon outside the splitting field with a random number. A photon will be discarded if the random number is more than its survival threshold. The
surviving photons called fat photons will have its weighting factor increased to 1 and will be tracked normally.

The idea of Bremsstrahlung Splitting is to create a uniform particle fluence over a particular area. The average dose in the area is the same because the weighting factor of every split photon is reduced. The simulation takes longer time to complete because there are more particles to track. However, the uncertainty of the dose calculation is reduced relatively faster than the increase in the simulation time (Kawrakow, Rogers et al., 2004). The overall efficiency in the simulation is therefore increased. In addition, photons that have no significant contribution to the dose in the region of interest will be eliminated by Russian roulette. This process further improves the efficiency in the simulation as the efficiency of MC simulation \( \varepsilon \), depends on the statistical uncertainty \( s^2 \), and the computation time, \( T \) (proportional to number of particles in the simulation, \( N \)), given by equation (27). The optimal values for the DBS parameters such as the FS, electron splitting plane, Russian roulette plane and NBRSPPL were reported (Kawrakow et al., 2004). For 6 MV photon beam, the optimum value for NBRSPPL is 1000 which was used. The radius of the splitting field should not be much larger than the width of the actual radiation field. For the 10 x 10 cm\(^2\) field, the splitting field radius of 10 cm is recommended. The Russian roulette plane should be above the electron splitting plane and they both should be close to the bottom of the flattening filter. With the optimum setting, DBS increases the central axis dose efficiency by a factor 6.4 compared to the previous Selective Bremsstrahlung Splitting method and even higher compared to non-splitting simulation. The efficiency decreases with the radiation field size and energy. Next lines of codes describe the geometrical information of all components (target, primary and secondary collimators, filter, shields,
etc) that are used to simulate the linac head as shown in figure 19, and default EGS MC control parameters used in the simulations.

The DOSXYZnrc was used to simulate absorbed dose in there-dimensional water phantom. The cartesian coordinate system with X, Y, Z-axes describes the position and size of the phantom orientated in the lateral, longitudinal and vertical directions respectively as in the treatment room. Another set of parameters specifies the type of medium of each voxel and the region for which the calculated dose is displayed in the output-listing file. The source configuration parameters describe a type of radiation source with a charge, energy, or information about the phase-space file if it is used. In our case, the phase-space files (PHSP A and PHSP B) generated by the BEAMnrc were used as the source for dose simulations using the DOSXYZnrc. An additional line of code is required to specify the location of the phase-space file in the input file. In general, the output files contain Listing and Log files which provide a detailed description of the simulation and any uncertainties produced in the simulation. Graphics files are created so that the simulation can be viewed graphically via the EGS_Windows program. The Data file contains the simulated absorbed doses in every voxel in the phantom. The data file is created in a format so that a built-in program, called STATDOSE, can be used for statistical analysis. However, the data files (.3ddose files) were extracted and analyzed using MATLAB (The MathWorks.com) 2020 during the research.

After the beam model was developed, MC dose calculations of the 6 MV FFF and 10 MV FFF photon beams for (a) the reference fields (10 cm x 10 cm and 5 cm x 5 cm) with or without MLC at 100 cm SSD, d = d_{max} (b) the machine-specific MLC fields (10 cm x 10 cm–1 cm x 1 cm) at 95 cm SSD, 5 cm depth (100 cm SAD), and (c) the virtual cones
at 100 cm SAD (i.e., at 95 cm SSD, 5 cm depth and 90 cm SSD, 10 cm depth) were performed respectively in the water phantom. Major input variables during the simulations were incident particle/histories (30–50 millions), voxel size (1–5 mm) of phantom and variance reduction techniques such as range rejection, directional bremsstrahlung splitting and particle splitting to reduce the statistical dose uncertainty < 5%. The phantoms were defined as three-dimensional water phantoms with dimensions (30 x 30 x 30) cm³ for larger fields and (20 x 20 x 20) cm³ for smaller fields. The dose outputs (.3ddose files) from the DOSXYZnrc program were extracted and analyzed using the MATLAB 2020. The entire MC modeling can be dose as shown in figure 20.

Figure 20: A flowchart of MC simulations for the dose calculations using the BEAMnrc and the DOSXYZnrc programs.
2.3 MC Model Validation

The MC simulations were conducted several times by varying main input parameters; (1) MC control inputs (i.e., initial number of particles/histories from 10–50 million), (2) variance reduction techniques (i.e., directional bremsstrahlung splitting, photon forcing, particle splitting, Russian roulette), and (3) transport control inputs (i.e., range rejection, global cut off energy) until the targeted uncertainty (< 5%), more precisely ≤ 3%, was achieved for large fields. The phase-space files (i.e., PHSP A and PHSP B) were stored just below the jaws and HDMLC to optimize the simulation efficiency and the optimal model parameters were applied to minimize the dose (relative) uncertainty. We have also varied voxel size of the water phantom from 1–5 mm during the MC dose simulations.

At first, the MC simulations were performed for the reference fields (10 x 10 cm\(^2\) and 5 x 5 cm\(^2\)) of 6 MV FFF and 10 MV FFF photon beams with or without MLC at 100 cm SSD in water phantom until the dose uncertainty < 5%. The dose outputs (2D dose profiles & depth dose curves) between the simulations with/without MLC were analyzed to validate the MC model and compared with those calculated from Eclipse treatment planning system. The comparison of dose profiles at 100 cm SSD, d = d\(_{\text{max}}\) was based on full-width half maximum (FWHM) calculations, while the depth dose curves were compared using Percentage Depth Dose (PDD) or Tissue Maximum Ratio (TMR). In MC simulations of 10 x 10 cm\(^2\) reference field, depths of maximum dose were 1.3–1.4 cm for 6 MV FFF beam and 2.2–2.4 cm for 10 MV FFF beam respectively. Secondly, the dosimetric parameters such as PDD, TMR, OAR, OF between the MC simulations and TPS were compared for the MLC fields of 10 x 10–1 x 1 cm\(^2\) at 95 cm SSD, 5 cm depth in
the water phantom. Finally, the dosimetric (OF) uncertainties in fixed virtual cones were evaluated between the MC, TPS for further assessment.
3. Results and Discussion

3.1 Field Output Factors (IFM)

Figures 21 and 22 display the field output factors (using IFM) as the function of equivalent square field size for the 6 MV and 10 MV FFF beams at 100 cm SAD respectively. The field output factors are determined from equation (23) in which the dose outputs are obtained from measurements with the micro-chamber (PTW 31021) and SRS diode (PTW 60018), and the published outputs corrections are applied to account detector’s over-response (diode) and under-response (chamber) in small fields.25,79 The output correction applied was up to 5% (i.e., k-factors used were 0.966 and 0.953 for 6 MV and 10 MV FFF beams respectively) for the smallest field (0.5 cm x 0.5 cm). Clearly, it is seen that the field output factors are SSD dependent (i.e., increase in field output factor with a higher SSD and vice versa). The fall-off in outputs is somewhat smooth from 10 cm x 10 cm to 2 cm x 2 cm fields, whereas they are dropping very sharply for smaller fields. It is also noted that the 6 MV FFF beam yielded relatively higher values than the 10 MV FFF beam as expected. However, the deviation of outputs between two clinical settings is slightly bigger for the 6 MV FFF beam (up to 7%) compared to the 10 MV FFF beam (up to 5%). It means the 10 MV FFF beam delivers more stable and consistent dose outputs regardless of geometrical settings used for the measurement.
**Figure 21:** Field output factors as a function of equivalent square field size ($S_{\text{clin}}$) for the 6 MV FFF beam at (a) 95 cm SSD, 5 cm depth and (b) 90 cm SSD, 10 cm depth using the intermediate field method (IFM). The blue and brown colors represent the field output factors at settings (a) and (b) respectively with their respective uncertainties.

Usually, the small field dosimetry requires a use of multiple reference detectors such as micro-ion chambers, diodes to achieve higher dose accuracy.\cite{10,14} The ion chambers are considered as gold standard for dosimetric measurements but not preferred to measure relative dose in the smaller fields due to volume averaging effect and relative size compared to the lateral range of secondary charged particles. Despite of such limitations, small-volume ionization chambers have been successfully used for dosimetric measurements.
with substantial correction factors for small and very small fields. The diodes (unshielded or stereotactic) are recommended for smaller fields due to their smaller size (active volume) and higher resolution. However, they tend to over respond in the smaller fields due to the presence of higher density material (silicon) that tends to perturb a fluence considerably.

A possible solution to overcome/minimize such detector related issues in the smaller fields, is to use the intermediate field method (IFM) or the daisy chain technique to determine the field output factors normalized to 10 x 10 cm² field. This method applies the cross-calibration of diode against the chamber in an intermediate field to connect measurements in large fields to small fields that is represented by Eq. (23). For instance, the dose output measured in any clinical field can be connected to 10 cm x 10 cm reference field via 4 cm x 4 cm intermediate field which is considered as the reference field for diode measurements. The over-response of diode and under-response of micro-chamber in smaller fields are corrected using appropriate or published detector-specific correction factors. In figures 21 and 22, a slight shift of curves for field size ≤ 5 cm represents the effect of applied corrections in dose outputs. However, the dose deviation between the uncorrected versus corrected outputs becomes pronounced in very small fields. Although this technique accounts for energy dependence of large fields, it does not consider the electron fluence perturbation due to the high density of silicon in small fields. It should be noted that even with the use of IFM, each small-field detector reading still needs to be corrected with the appropriate field output correction factors.
Figure 22: Field output factors as a function of equivalent square field size ($S_{\text{clin}}$) for the 10 MV FFF beam at (a) 95 cm SSD, 5 cm depth and (b) 90 cm SSD, 10 cm depth using the intermediate field method (IFM). The blue and brown colors represent the field output factors at settings (a) and (b) respectively with their respective uncertainties.

It is also be noted that the field output factors or output correction factors for small fields are expressed as a function of the equivalent square field size ($S_{\text{clin}}$) given by Eq. (22). It is observed that once $S_{\text{clin}}$ begins to drop from 5 cm to smaller ones, the dose output begins to drop notably, and it drops dramatically for the smallest field size of 0.5 cm (Figures 21 and 22). Clearly, the significant drop of dose outputs in intermediate to smaller fields is mainly due to small field conditions such as loss of LCPE, penumbra overlapping, size of detector and the dramatic drop of output in the smallest field due to extreme small field
conditions. The FWHM of lateral profile is considered the most representative field size parameter for accurate small-field dosimetry because the detector response and perturbation effects are determined by it at the depth of measurement. To facilitate establishing a relationship between FWHM and collimator setting, it is recommended that both FWHM and geometrical field size, together with the depth it belongs to, be recorded when reporting small-fields data. The clinical field sizes, defined by the FWHM of the field at the position of detector are almost identical to the nominal field sizes (defined by the projected collimator setting) for field sizes ≥ 3.0 cm. But they differ meaningfully for the smaller fields (< 3.0 cm), thereby resulting maximal difference (by 1.0 mm) for the smallest field size of 0.5 cm. This effect (apparent widening of field size) is dominant for higher energy beam due to increase of the range of secondary electrons that is responsible for a sharp drop in beam output. Additionally, the collimator that defines a small-field shields photons that are scattered from different components of linac head, including primary collimator, flattening filter. Thus, the number of low-energy photons scattered from the linac head reaching the center of small field is decreased. It results in a hardening of the photon energy spectrum at any point on the beam axis with decrease in field size. This effect has the potential to impact the ratio of mass-absorption coefficients and stopping power ratios between water and the detector material during the OF measurements.

3.2 Fixed Virtual Cone Technique

The preliminary results of retrospective measurements (SRS MapCHECK) of 30 SRS treatment plans (intracranial cases) containing several small arc fields (S_{clin} ≤ 1.0 cm) with different collimator angles have shown a notable (≥ 5%) dose difference between the
treatment plans and measurements. The 10 MV FFF beam has shown relatively better agreement than the 6 MV FFF beam.

3.2.1 Quality Assurance

Table I shows results of the field-by-field study (gamma evaluation) of 10 trigeminal cases (total of 130 fields) treated with the fixed virtual cone technique. Figure 25 represents the gamma evaluation (patient-specific quality assurance) for a trigeminal treatment plan using the SRS MapCHECK, in which a cumulative absolute dose between the TPS plan and measurement are compared with a global 3%/1 mm criteria. The absolute value of the average difference in dose/MU (obtained from the equation 31) between the TPS and SRS MapCHECK measurements for the 10 MV FFF beam was found (2.35 ± 1.33) % compared to (4.05 ± 1.50) % for the 6 MV FFF beam with 0.1 cm spatial resolution. Whereas the corresponding differences were obtained (4.10 ± 1.50) % and (6.95 ± 1.70) % respectively with 0.125 cm spatial resolution. However, the dose (at central axis) difference jumped up to 10% for certain fields (arcs) depending on the gantry and collimator angles, energy beams, spatial resolution. The discrepancy was observed comparatively higher for the arc field with the gantry at 90°/270° and collimator at 90° angles. In addition, the comparison of whole treatment plans (i.e., cumulative dose comparison) resulted a higher passing rate (≥ 97%) for the 10 MV FFF beam compared to the 6 MV FFF beam (≥ 95%) with both 2%/2 mm and 2%/1 mm gamma criteria (figure 25). The beam delivery time for 10 MV FFF plan was ~ 5 minutes (~ 30% decrease) shorter than for the 6 MV FFF plan. Moreover, the measured dose profiles defined by $S_{clin}$ are seen relatively wider than the planned ones and apparent widening of field size can be clearly visible in very small fields. For instance,
the clinical field sizes ($S_{clin}$) given by Eq. (22) are reported in the ranges 0.57–0.60 cm and 0.60–0.65 cm for 0.5 cm fixed virtual cones of 6 MV and 10 MV FFF beams, respectively.

Table I: Comparison of dose (cGy) per monitor unit (MU) between the TPS (Acuros XB) and measurements (SRS MapCHECK) during the field-by-field study of 10 treatment plans (130 fields) of trigeminal cases treated with the fixed virtual cone technique.

<table>
<thead>
<tr>
<th>Beam → Resolution ↓</th>
<th>6 MV FFF</th>
<th>10 MV FFF</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Calculated</td>
<td>Measured</td>
</tr>
<tr>
<td>0.100 cm</td>
<td>0.454 ± 0.023</td>
<td>0.472 ± 0.022</td>
</tr>
<tr>
<td>0.125 cm</td>
<td>0.442 ± 0.023</td>
<td>0.475 ± 0.021</td>
</tr>
</tbody>
</table>

During the pre-treatment quality assurance, the field-by-field (10 treatment plans, 130 fields) dose/output comparison using SRS MapCHECK showed acceptable (≤ 3% difference) agreement on average (Table I). However, the dose (at central axis) difference jumped up to 10% for certain fields (arcs) depending on the gantry and collimator angles, energy beams, spatial resolution. A higher dose discrepancy was associated with the arc field when the gantry at 90°/270° and collimator at 90° angles. It confirmed our initial hypothesis that a meaningful dosimetric variation can be observed with different gantry/collimator angles during the small field SRS arc treatment. Such variation could arise from either the effect of gravity or improper MLC QA that will impact the accuracy of dose delivery.\(^{45,68}\) However, our approach tends to minimize these inconsistencies by utilizing the fixed field geometry plus multiple non-coplanar arcs and to deliver accurate
dose to the target with a higher degree of precision. It allows freedom to manipulate the dose delivery (i.e., with selective arcs, arc span, and arc weight, variable MUs) such that the tumor receives maximum dose and critical organs receive minimal dose. As the dose per degree of the arcs is proportional to the sine of the gantry angle to generate a spherical dose distribution. If it is constant, the resulting dose distribution is oblate or ellipsoid in shape. Due to the constancy of dose per degree of arcs, the isodose lines/curves for trigeminal treatment plans are appeared to be ellipsoidal with significant dose spill in anterior-posterior direction as illustrated in Fig. 25 (a), (b). One reason of selecting the 10 MV FFF beam for the treatment (trigeminal neuralgia) is that the dosimetric results obtained using 10 MV FFF beam are more accurate and consistent compared to the 6 MV FFF beam. Another reason is due to its capacity of delivering higher dose rate (up to 2400 MU/min) that can minimize the treatment time notably. Automated table motion during the beam delivery will further reduce treatment time thereby making our technique even more efficient. As this technique utilizes the fixed field geometry (0.5 cm x 0.5 cm MLC field) in addition with multiple partial arcs, it is expected to limit the uncertainties associated with undesirable movements of leaves during the gantry rotation resulting more accurate and reliable dose delivery.

Table II shows the comparison of dosimetric parameters (field output factors and dose/MU) between the TPS and diode (PTW 60018) measurements for 0.5 cm x 0.5 cm fixed virtual cone field of 6 MV and 10 MV FFF photon beams at 95 cm SSD, 5 cm depth and 90 cm SSD, 10 cm depth settings. For 10 MV FFF beam, the differences of field output factors between the TPS and measurements (obtained from the equation 31) were - 1.26% and - 1.00% at 95/5 cm, and 90/10 cm settings respectively. Whereas such differences for
the 6 MV FFF beam were - 2.42% and - 1.45% at 95/5 cm, and 90/10 cm depth, respectively. Similarly, difference in dose/MU for the 10 MV FFF beam were - 1.27% and - 1.15% at 90/10 cm and 95/5 cm settings respectively. Whereas the corresponding values were - 3.30% and - 2.50% respectively for the 6 MV FFF beam. It should be noted that the measured dose parameters are consistently higher as compared to the calculated ones because of the calibration constraints on the TPS (based on ion-chamber) as well as dose over-response of diode in small fields.

**Table II:** The comparison of field output factors (OF) and Dose/MU between the Eclipse TPS (Acuros XB) and measurements (SRS diode) for the fixed virtual cone (0.5 cm x 0.5 cm MLC field) for 6 MV FFF and 10 MV FFF photon beams at 95 cm SSD, 5 cm depth and 90 cm SSD, 10 cm depth respectively.

<table>
<thead>
<tr>
<th>Beam + Setting</th>
<th>Field output factors</th>
<th>Dose (cGy)/MU</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Calculated</td>
<td>Measured</td>
</tr>
<tr>
<td>6 MV FFF 95/5 cm</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>0.605</td>
<td>0.620</td>
</tr>
<tr>
<td>10 MV FFF 95/5 cm</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>0.550</td>
<td>0.557</td>
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<tr>
<td>6 MV FFF 90/10 cm</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>0.560</td>
<td>0.568</td>
</tr>
<tr>
<td>10 MV FFF 90/10 cm</td>
<td></td>
<td></td>
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<tr>
<td></td>
<td>0.510</td>
<td>0.515</td>
</tr>
</tbody>
</table>

Table III represents the comparison of field output factors, dose/MU between the calculated (TPS) and measured (SRS diode) for all virtual cone fields (including ± 1.0 leaf shift) with respective uncertainties (in percentage) in the brackets for different energy beams and
The average difference of field output factors (given by equation 31) were found approximately - 3.0% and - 6.0% for 10 MV FFF and 6 MV FFF beams respectively as shown in figure 23. The TPS results underestimated the output factors variation (standard deviation (SD) ~ 4.0%) in those non-square (asymmetric) MLC fields as compared to measurements (SD ≤ 2%). But the dosimetric mismatch was seen comparatively insignificant (≤ 3% error) in the virtual cones with a leaf shift ± 0.5 mm. Similarly, Figure 24 displays the comparison of dose/MU in those fields resulting overall discrepancies approximately - 3% for the 10 MV FFF beam and - 5% for the 6 MV FFF beam respectively.

**Table III:** Comparison of field output factors, dose (cGy/MU) between the Eclipse TPS (Acuros XB) and measurements (SRS diode) for virtual cone fields including ± 1.0 mm MLC shift of the 6 MV FFF and 10 MV FFF beams at 95 cm SSD, 5 cm depth and 90 cm SSD, 10 cm depth clinical settings. The corresponding uncertainties (in percentage) associated with the calculated/measured values are given in the brackets.

<table>
<thead>
<tr>
<th>Beam + setting</th>
<th>Field output factors</th>
<th>Dose (cGy)/MU</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Calculated</td>
<td>Measured</td>
</tr>
<tr>
<td><strong>6 MV FFF 95/5 cm</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>0.596 (4.50)</td>
<td>0.636 (1.65)</td>
</tr>
<tr>
<td><strong>10 MV FFF 95/5 cm</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>0.546 (4.10)</td>
<td>0.564 (1.60)</td>
</tr>
<tr>
<td><strong>6 MV FFF 90/10 cm</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>0.556 (4.25)</td>
<td>0.585 (1.55)</td>
</tr>
<tr>
<td><strong>10 MV FFF 90/10 cm</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>0.506 (3.90)</td>
<td>0.524 (1.52)</td>
</tr>
</tbody>
</table>
It is noted that relatively higher difference of dose outputs between measurements and the TPS (Table III) in concert with a larger dosimetric variation in fixed virtual cones (non-square/asymmetric fields) using the 6 MV FFF beam compared to the 10 MV FFF beam (Figs. 23 and 24). Comparatively lower values of dose parameters for the 10 MV FFF beam than the 6 MV FFF beam indicate that more particles escape the field due to a larger lateral range of the secondary charged particles and the higher transmission through the penumbra for higher energy beams. An overall dosimetric inconsistency in those fields are found to be insignificant (difference ≤ 3%) with ± 0.5 mm leaf shift.\textsuperscript{45,68} Whereas it is noteworthy (up to 6% difference) with ± 1.0 mm shift. These results are consistent with the field-by-field study during the quality assurance process specifically for arc fields with the gantry at 90° or 270° and the collimator at 90° angles. Note that all possible dosimetric fluctuations due to a tiny geometrical shift (± 0.5–1.0 mm) of leaves either caused by gravity or improper MLC QA during the gantry or collimator rotation are examined in advance, which gives the flexibility of applying corrections during the treatment planning if needed. It was observed that the clinical setting of 90/10 cm has shown relatively consistent results than the 95/5 cm setting regardless of energy beams. Overall, the 10 MV FFF beam has delivered more accurate and reliable dose outputs than the 6 MV FFF beam for virtual cone fields and the beam at 90/10 cm setting produced better results.
**Figure 23**: Comparison of field output factors (OF) between the measurements (SRS diode) and Eclipse TPS (Acuros XB) for the fixed virtual cones (with ± 1.0 leaf shift) of 6 MV FFF and 10 MV FFF photon beams at 95/5 cm and 90/10 cm clinical settings.
3.2.2 Retrospective Case Studies

Figure 25 represents the gamma evaluation (patient-specific quality assurance) for a trigeminal treatment plan (10 MV FFF beam) using the SRS MapCHECK, in which a cumulative absolute dose between the TPS plan and measurement are compared with 2%/1 mm gamma criteria and 10% threshold. The gamma evaluation of 15 trigeminal cases has shown the average passing rate of (98.5 ± 1.5) % with 2%/1 mm gamma criteria.
Figure 25: A representative case for gamma evaluation of treatment plans (10 MV FFF beam) for trigeminal neuralgia using the fixed virtual cone technique. Sections (a) and (b) represent planner dose maps obtained from the TPS (Acuros XB) and SRS MapCHECK measurement respectively. The sections (c) and (d) show the corresponding dose profiles in x and y directions respectively. Solid line represents the planned dose profile, whereas yellow circle represents the measured one.

Isodose curves are ellipsoidal in shape due to the usage of multiple non-coplanar arcs and variable MUs during the treatment. The isodose lines/curves are ellipsoidal in shape due to the usage of various non-coplanar arcs and variable MUs during the treatment as shown in
The dimensions (i.e., major, and minor diameters) of 90%, 50% and 25% isodose lines are manually computed from the treatment plans and measurements. The average major diameters of the 90%, 50% and 25% isodose curves are (2.1 ± 0.3) mm, (7.0 ± 1.1) mm, and (11.5 ± 1.1) mm respectively. Whereas the respective minor diameters are (1.82 ± 0.25) mm, (5.8 ± 0.7) mm, and (7.6 ± 0.8) mm. The equivalent diameters of circle or sphere in each volume encompassed by 90%, 50% and 25% isodose curves (coronal plane) for the calculated and measured dose distributions are provided in Table IV. The equivalent diameters of the corresponding circles or spheres of given volume are computed to approximate the prescribed, irradiated volumes, dose coverage, proximity of critical organ (brainstem). Comparatively the TPS underestimated the calculation of diameters. The equivalent diameters of 50% isodose lines are comparable to that generated by 5 mm physical cone. The maximum dose delivered to the critical organ such as brainstem was reported 18 Gy (< 25 Gy) on average. The average volume of brainstem receiving maximum dose was 0.03 cm³.

**Table IV:** Calculated and measured equivalent diameter of 90%, 50% and 25% isodose lines in coronal plane for 15 trigeminal cases treated with fixed virtual cone method.

<table>
<thead>
<tr>
<th><em>Equivalent diameter (mm)</em></th>
<th>90% isodose curve</th>
<th>50% isodose curve</th>
<th>25% isodose curve</th>
</tr>
</thead>
<tbody>
<tr>
<td>Calculated</td>
<td>Measured</td>
<td>Calculated</td>
<td>Measured</td>
</tr>
<tr>
<td>1.95 ± 0.30</td>
<td>1.97 ± 0.20</td>
<td>6.10 ± 0.70</td>
<td>6.40 ± 0.50</td>
</tr>
</tbody>
</table>

*The equivalent diameter is the diameter of a circle or a sphere of the given volume*
3.3 MC Dose Modeling and Validation

3.3.1 Reference Fields

The optimal MC results (≤ 3% dose uncertainty and ≤ 3% dose difference to TPS) were obtained after simulating 30–50 millions of incident particles for 10 x 10 cm² and 5 x 5 cm² fields of both 6 MV FFF and 10 MV FFF beams at 100 cm SSD in water phantom. Figures 26–28 represent the comparison of MC simulated dose profiles and depth dose curves for the 10 cm x 10 cm reference field. Whereas the figures 29-31 exhibit the corresponding dose profiles and depth dose curves for 5 cm x 5 cm field. The comparison of two-dimensional dose profiles between the MC simulations and the Eclipse TPS have shown a reasonable agreement (≤ 2.5% difference) for 10 x 10 cm² and 5 x 5 cm² fields with/without MLC of both photon beams, defined at SSD = 100 cm and d = d_{max} in water phantom. The dose (relative) uncertainty during the simulations was achieved ≤ 3% and depths of maximum dose were recorded at 1.3–1.4 cm for 6 MV FFF beam and 2.2–2.4 cm for 10 MV FFF beam respectively. Note that the comparison of MC simulated versus TPS generated dose profiles exhibited that the important geometrical features of MLC (leaf tip, tongue and groove, dosimetric leaf gap) were accurately modeled in the MC simulations. Similarly, the comparison of corresponding depth dose curves has displayed consistent dose fall off as expected for both energy beams as shown in figures 28 and 31.
Figure 26: Comparison of the MC simulated dose profiles (2D) for 10 x 10 cm² field of 6 MV FFF photon beam (a) without MLC and (b) with MLC at SSD = 100 cm, d = d_{max} in water phantom using a voxel size 5 mm.
Figure 27: Comparison of the MC simulated dose profiles for 10 x 10 cm$^2$ field of 10 MV FFF photon beam (a) without MLC and (b) with MLC at SSD = 100 cm, d = $d_{\text{max}}$ in water phantom.

The MC dose calculation method is subject to statistical uncertainty that is proportional to number of particles/histories in simulations.$^{28,53}$ However, it tends to provide the most accurate and realistic outcomes $^{53,82}$ as compared to existing analytical algorithms. Generally, there are two common sources of uncertainties in MC dose calculations—those resulting from the simulation of linac head and those arising from fluctuations in the phantom. In this case, the statistical uncertainties in the simulations resulted from the phase-space file of linac head above jaws, MLC geometry and phantom. The initial targeted dose uncertainty (relative) was set below 5% for simulations of static fields during the model validation. The optimal MC results with $\leq 3\%$ statistical uncertainty were achieved after simulating 30–50 millions of incident particles in concert with the optimal model.
parameters\textsuperscript{76} for 10 x 10 cm\textsuperscript{2} and 5 x 5 cm\textsuperscript{2} reference fields of both 6 MV FFF and 10 MV FFF beams in water phantom.\textsuperscript{56} However, the uncertainty was appeared to be increasing for smaller fields as expected due to the presence of small field conditions.\textsuperscript{10} Besides, the MC dose profiles (2D) exhibited more realistic dose distributions, which indicated the accurate modeling of important geometrical features of MLC such as leaf tip, leaf type/shape, tongue and groove, dosimetric leaf gap.

![Image](image_url)

**Figure 28:** Comparison of the MC simulated depth dose curves with and without MLC for 10 x 10 cm\textsuperscript{2} field of (a) 6 MV FFF beam and (b) 10 MV FFF beam at SSD = 100 cm in water phantom (voxel size, 5 mm).
Figure 29: Comparison of the MC simulated dose profiles for 5 x 5 cm$^2$ field of 6 MV FFF photon beam (a) without MLC and (b) with MLC at SSD = 100 cm, $d = d_{\text{max}}$ in water phantom (voxel size, 3 mm).
Figure 30: Comparison of the MC simulated dose profiles for 5 x 5 cm$^2$ field of 10 MV FFF photon beam (a) without MLC and (b) with MLC at SSD = 100 cm, $d = d_{\text{max}}$ in water phantom (voxel size, 3 mm).
**Figure 31:** Comparison of the MC simulated depth dose curves with and without MLC for 5 x 5 cm² field of (a) 6 MV FFF beam and (b) 10 MV FFF beam at SSD = 100 cm in water phantom (voxel size, 3 mm).

### 3.3.2 Machine Specific Fields

Figures 32 and 33 represent the comparison of output factors between the MC simulations and TPS for static MLC fields (10 x 10–1 x 1) cm² of 6 MV FFF and 10 MV FFF respectively at 95 cm SSD, 5 cm depth (100 cm SAD). The output factors differences were found within 2.5% and 2% for the 6 MV and 10 MV FFF beams respectively. Similarly, the comparison TMR values between MC simulations and TPS has shown within 2.5%
discrepancy for both beams. In particular, the MC simulated TMR values were relatively lower compared to TPS ones, while the MC simulated field output factors were comparatively higher than those obtained from TPS for all fields except for very small fields (≤ 1.5 x 1.5 cm²). It is also noted that the statistical (dose) uncertainty as well as dosimetric discrepancy were increased from large to small fields as expected. Overall, the 10 MV FFF beam has shown a better agreement than 6 MV FFF beam.

The 10 MV FFF beam (Fig. 33) has delivered relatively better matching results than 6 MV FFF beam (Fig. 32). Our model calculated slightly higher output factors for 2 x 2–10 x 10 cm² fields whereas it computed relatively lower values for very small fields (≤ 1.5 x 1.5 cm²) for both beams. In addition, the MC calculated values of $S_{clin}$ given by Eq. (22) are found relatively larger compared to the TPS for smaller fields, which is on a par with measured results.\textsuperscript{17,62} It is evident that the apparent widening effect becomes pronounced in very small fields\textsuperscript{9,56} and even more for higher energy beams.\textsuperscript{10} It is due to the fact that more particles can escape the field due to a larger lateral range of the secondary charged particles and the higher transmission through the penumbra for higher energy beams.
Figure 32: Output factors (OF) are plotted as a function of field size for 10 x 10 – 1 x 1 cm² MLC fields of 6 MV FFF photon beam at 95 cm SSD, 5 cm depth in water phantom. The pink (diamond) and blue (circle) color lines represent the OFs calculated using MC and Acuros XB, respectively.
Figure 33: Output factors (OF) are plotted as a function of field size for 10 x 10 – 1 x 1 cm\(^2\) MLC fields of 10 MV FFF photon beam at 95 cm SSD, 5 cm depth in water phantom. The pink (diamond) and blue (circle) color lines represent the OFs calculated using MC and Acuros XB respectively.

### 3.3.3 Fixed Virtual Cone Fields

An independent MC study of the fixed virtual cone technique was performed, and dose profiles and parameters (TMR, OF) were compared against the TPS using water phantom. It included two smallest MLC fields of 0.5 cm x 0.5 cm and 0.75 cm x 0.75 cm as fixed virtual cones and evaluation of dosimetric variation due to a minute leaf shift (± 0.5–1.0 mm). Figures 34 and 35 show the comparison of dose profiles between the MC and TPS for the static MLC field defining the fVC for the 6 MV FFF and 10 MV FFF beams.
respectively at 90/10 cm clinical setting. The equivalent square field size \( S_{clin} \), given by Eq. (22), of the MC simulated dose profiles were recorded slightly higher than that obtained from the TPS. Particularly, for nominal square field sizes of 0.5 cm and 0.75 cm, \( S_{clin} \) were 0.60 cm and 0.79 cm (MC) as compared to 0.57 cm and 0.77 cm (TPS) respectively using 6 MV FFF beam (Figure 34), whereas 0.64 cm, 0.81 cm (MC) versus 0.59 cm, 0.79 cm (TPS) respectively using 10 MV FFF beam (Figure 35). Similarly, the comparison of TMR and OF values has shown a reasonable (≤ 5% difference) agreement thereby the 10 MV FFF beam producing relatively better results that the 6 MV FFF beam.

**Figure 34:** Comparison of the (a) MC simulated versus (b) TPS generated dose profiles for the fixed virtual cone (0.5 cm x 0.5 cm HDMLC field) of the 6 MV FFF photon beam at 90 cm SSD, 10 cm depth in the water phantom (voxel size, 0.5 mm).
Figure 35: Comparison of the (a) MC simulated versus (b) TPS generated dose profiles for the fixed virtual cone (0.5 cm x 0.5 cm HDMLC field) of the 10 MV FFF photon beam at 90 cm SSD, 10 cm depth in the water phantom (voxel size, 0.5 mm).

Figure 36 represents the overall field output factors comparison for the fVC fields considering ± 1.0 mm leaf shift among the diode measurements, TPS and MC calculations. Early results of MC study for dosimetric variation of two smallest fields including a minute ± (0.5–1.0 mm) leaf shift have shown similar (pattern) dosimetric variation (SD ≤ 2.5%) as the measured ones (SD ≤ 2.0%) but relatively better than the TPS results (~ 4%). However, the model needs further investigation/optimization for the quantitative comparison as shown in figure 36. The potential sources of the larger dosimetric uncertainties as well as discrepancies reported in such very small fields could be due to: 9,14,56,83 (1) the presence of prominent small-field conditions, (2) an inaccurate modeling
of geometry (MLC), (3) the adoption of different dose calculations modalities (i.e., a volume-based dose calculation in MC versus a point-based dose calculation in TPS), and (4) the difference in source type (i.e., energy spectrum or fine tuning versus monoenergetic beam). It should be noted that the TPS has shown a lack of ability of handling such asymmetric fields. It could result a significant dosimetric inconsistencies during small-field SRS arc treatments using dynamic MLC fields unless these are properly accessed in advance. These results justify the importance of MC study specifically for smaller fields besides the TPS and measurements. However, the MC model needs further improvements that can be done with rigorous simulations by inserting the accurate geometrical (i.e., MLC, detector, and phantom) details for the complete assessment.

**Figure 36:** Comparison of field output factors among the measurements (uncorrected, corrected), TPS and MC for the fixed virtual cone fields (~ 0.5 cm x 0.5 cm HDMLC field) of 6 MV FFF and 10 MV FFF beams at two clinical settings.
Overall, the 10 MV FFF beam has shown better performance than the 6 MV FFF beam in both clinical settings and the setting of 90 cm SSD, 10 cm depth has shown more favorable results. These results substantiate that our fixed virtual cone technique using the 10 MV FFF beam is straightforward and reliable, which can be effectively implemented in routine small target radiosurgery such as trigeminal neuralgia.
4. Conclusions

This research presents results of field output factors determined from measurements with a micro-ion chamber and SRS diode and using the IFM for 10 x 10–0.5 x 0.5 cm² MLC fields of the 6 MV and 10 MV FFF beams at 100 cm SAD on the Varian EDGE radiosurgery system.

The fixed virtual cone approach using the 0.5 cm x 0.5 cm HDMLC field for small target radiosurgery (trigeminal neuralgia) is introduced and validated (with ≤ 3% difference) against measurements using the SRS MapCHECK and EBT films. As the pre-treatment QA, the field-by-field comparison of dose outputs between the TPS and SRS MapCHECK measurements has shown relatively better agreement for the 10 MV FFF beam compared to the 6 MV FFF beam. Additionally, the comparison of dosimetric parameters (field output factors, and dose/MU) between the TPS and SRS diode measurements in the fixed virtual cone fields including a minute leaf shift (± 0.5–1.0 mm) has shown notable difference (3–6%) in concert with 2–4% uncertainty, depending on beam energy and clinical settings. Collectively, the 10 MV FFF beam has yielded relatively better results than the 6 MV FFF beam. Furthermore, early results of an independent MC study for the fixed virtual cone method have shown similar (trend) dosimetric variation in virtual cones as measured ones but relatively better than TPS results. However, further
investigation/improvement is needed by simulating the accurate geometrical (i.e., MLC, detector, and phantom) details for the complete assessment.

The viability of the fVC method for small target radiosurgery and its dosimetric robustness considering potential minute leaf shift during the treatment are evaluated from measurements and MC calculations. This method is a straightforward that yields a higher dosimetric certainty and is a reliable or superior alternative to a physical cone that can be routinely applied for the treatment of trigeminal neuralgia. This research can be extended further by a machine learning approach based on MLC log files as well as radiomics analysis from QA images.
5. References


17. Huq MS, Hwang MS, Teo TP, Jang SY, Heron DE, Lalonde RJ. A dosimetric evaluation of the IAEA-AAPM TRS483 code of practice for dosimetry of small static fields used in conventional linac beams and comparison with IAEA TRS-398,


63. PTW Freiburg. Detectores PTW - Diseños básicos y especificaciones. Published online 2013:100.


80. Huq MS, Hwang MS, Teo TP, Jang SY, Heron DE, Lalonde RJ. A dosimetric evaluation of the IAEA-AAPM TRS483 code of practice for dosimetry of small

