Small fields: Nonequilibrium radiation dosimetry

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Advances in radiation treatment with beamlet-based intensity modulation, image-guided radiation therapy, and stereotactic radiosurgery (including specialized equipments like CyberKnife, Gamma Knife, tomotherapy, and high-resolution multileaf collimating systems) have resulted in the use of reduced treatment fields to a subcentimeter scale. Compared to the traditional radiotherapy with fields $\geq 4\times 4\ \mathrm{cm}^2$, this can result in significant uncertainty in the accuracy of clinical dosimetry. The dosimetry of small fields is challenging due to nonequilibrium conditions created as a consequence of the secondary electron track lengths and the source size projected through the collimating system that are comparable to the treatment field size. It is further complicated by the prolonged electron tracks in the presence of low-density inhomogeneities. Also, radiation detectors introduced into such fields usually perturb the level of disequilibrium. Hence, the dosimetric accuracy previously achieved for standard radiotherapy applications is at risk for both absolute and relative dose determination. This article summarizes the present knowledge and gives an insight into the future procedures to handle the nonequilibrium radiation dosimetry problems. It is anticipated that new miniature detectors with controlled perturbations and corrections will be available to meet the demand for accurate measurements. It is also expected that the Monte Carlo techniques will increasingly be used in assessing the accuracy, verification, and calculation of dose, and will aid perturbation calculations of detectors used in small and highly conformal radiation beams. [DOI: 10.1118/1.2815356]

Key words: small field dosimetry, nonequilibrium conditions, Monte Carlo

I. INTRODUCTION

With image guidance and improved treatment delivery techniques at the core of modern radiation therapy, the treatment fields that traditionally spanned from $4\times 4\ \mathrm{cm}^2$ up to $40\times 40\ \mathrm{cm}^2$ are now being reduced down to a subcentimeter range in advanced and specialized radiation treatments such as beamlet-based intensity modulated radiation therapy (IMRT), image-guided radiation therapy (IGRT), tomotherapy, stereotactic radiosurgery (SRS) with high-resolution multileaf collimator, Gamma Knife, and CyberKnife. In particular, the SRS, Gamma Knife, and CyberKnife rely on very small field sizes on the order of a few millimeters to treat tumors and spare normal structures. IMRT requires the addition of small fields with nonequilibrium conditions to treat target volumes using optimization routines. Several dosimetric challenges due to this trend are lack of charged particle equilibrium (CPE),[1,2] partial blocking of the beam source giving rise to pronounced and overlapping penumbra,[3–6] the availability of small detectors for sizes comparable to the field dimensions,[7,8] and variations of the electron spectrum inducing changes in stopping power ratios.[9,10]

Electronic equilibrium is a phenomenon associated with the range of secondary particles and hence dependent on the beam energy, the composition of the medium, and particularly the density of the medium. With a large variety of radiation detectors marketed by various manufacturers covering all sizes (from mini to micro), types (ionization chamber, semiconductor, chemical, film, etc.), and shapes (thimble, spherical, plane parallel), the choice of a suitable detector for small field dosimetry could be a challenging and rather confusing task without proper guidelines. It is not too uncommon in clinical practice to compare measurements with various detectors and choose the detector that yields the highest output for a given field size, or to select a measured value that is common to several detectors, without proper consideration of the possible perturbations and corrections for each of the detectors. Such approaches do not provide the scientific basis needed to achieve the confidence for dosimetric accuracy commonly set for clinical practices. To deal with these difficulties, a growing number of authors[10–17] have reported the comparison of measured data with Monte Carlo simulation. However, at the present time Monte Carlo simulations cannot be assumed to invariably provide a gold standard without appropriate experimental validation. These developments challenge the conventional way of performing dosimetric measurement and treatment planning. The pur-

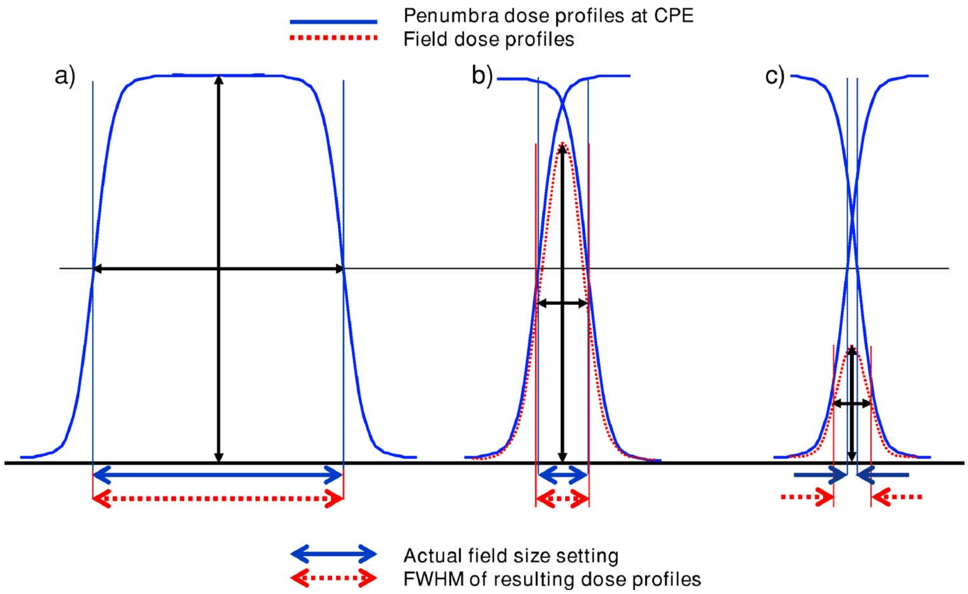


FIG. 1. With field sizes large enough to yield charged particle equilibrium (CPE) and fully viewed sources, the full width at half maximum (FWHM) of dose profiles yields correctly determined field sizes since the field borders will be approximately at the level of 50% of the dose level of CPE, as shown in panel (a). When the field size is of the same order as the charged particle lateral diffusion distance, the penumbra from opposing field edges overlap, causing a small error in field size determination from FWHM data [panel (b)], but completely break down for very small fields since the resulting curve has a lower maximum and hence its half value will be pushed outward from the correct position, resulting in an overestimated field size as shown in panel (c).

pose of this paper is to illustrate the problem, provide possible solutions, and predict future trends in small field radiation dosimetry.

II. PROBLEMS

II.A. What is small?

The definition of a small field in radiation dosimetry is currently very subjective and *ad hoc*. There is no clear consensus definition as to what constitutes a small field. Commonly, a field size of less than 3×3 cm^2 is considered outside the conventional treatment field size that needs special attention both in dose measurements and in dose calculations. A more scientific approach is needed to set the criteria which define a small field condition based on the beam energy and the density of the medium. There are essentially three "equilibrium factors" that determine the scale if a radiation field is to be considered as small or not: (i) the size of the viewable parts of the beam source as projected from the detector location through the beam aperture; (ii) the size of the detector used in measurements; and (iii) the electron range in the irradiated medium. These factors are discussed below.

II.B. Effects of the radiation source size

By collimating a beam from a source of finite width, it is clear that below a certain field size, only a part of the source area can be viewed from a detector's point of view. The output will then be lower than compared to field sizes at which the entire source can be viewed from the detector's field of view.[3,4,6,18–20] The output changes differently depending on position contributing to the phenomena yielding a blurred and widened profile as illustrated in Fig. 1. If the entire source cannot be viewed from the center of the field, then the geometrical penumbra is extended all over the field cross section.[3,4,19] Under such conditions traditional methods for field size determination such as full width at half maximum (FWHM) break down, resulting in overestimated field sizes as shown (Fig. 1). It has been demonstrated by Ding *et al.*[21] that the beam output (planar fluence profile) can be significantly influenced by the auxilary collimator (jaw) settings used to achieve the small field sizes. This is reflected in Fig. 2, indicating the profile patterns for the 6 MV beam for 6×6 and 24×24 mm^2 fields with various jaw settings. For nonfocused multileaf collimators the light fields are not congruent with the radiation field due to the curved nature of the leaves permitting variable amounts of radiation through different thicknesses of the leaves.[22] This positional dependence, varying from one side of the field to the other, further complicates the precise metrics of small field sizes.

II.C. Electron ranges and loss of CPE

The electrons produced from megavoltage photon beams have a considerable range that gets prolonged in a low-density medium. Compared to the field size, the lateral range of the electrons is the critical parameter to the CPE, rather than the forward range of the electrons. Li *et al.*[23] described the lateral range of electrons which are energy dependent. Shown in Fig. 3 are primary dose profiles in water across a collimating edge for different beam energies, specified by their quality index ($TPR_{20/10}$).[24,25] This provides the information of penumbra ranges in unit density media that set the dimensions of when small field conditions apply based on overlapping electron distribution zones from different field edges.[26]

II.D. Measurement

For photon beams, measured data are used in dose calculation to provide the absolute dose normalization at a reference field, and through a beam modeling procedure indirectly drive dose calculations based on relative data such as total scatter factor (S_{cp}), tissue maximum ratio (TMR), percent depth dose (PDD), and off-axis ratio (OAR). The refer-

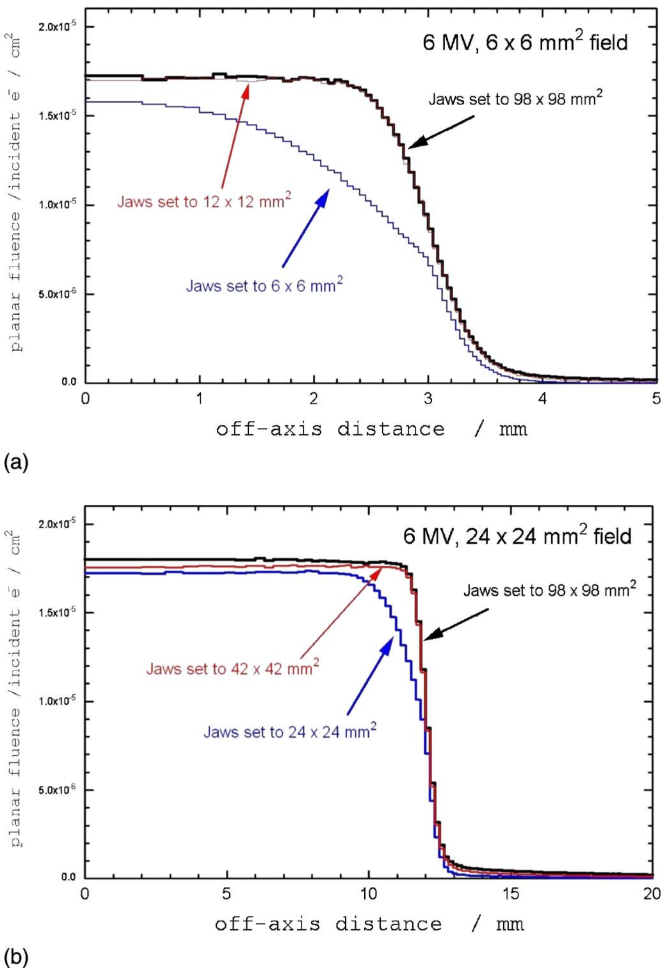


FIG. 2. The effects of source size and beam shaping geometry on the output of a small field, (a) 6×6 mm^2 and (b) 24×24 mm^2. Both field sizes are defined by the micro multi-leave collimator (mMLC) and the variation are caused by different settings of auxilary jaws. (Reproduced with permission from Ding *et al.*, Ref. 21.)

ence dose calibration is performed according to the dosimetry protocols[24,25,27] with a well-defined beam geometry, where beam quality and dosimetric parameters are known to a high degree of accuracy. Dose measurements with ionization chambers rely on the assumptions of cavity theory. When the size of the cavity is smaller than the range of charged particles originated in the medium, the cavity is treated as nonperturbing. In such a situation, the dose to the medium is related to the dose to the air in the cavity by the stopping power ratios of medium to air. However, with decreasing field size, neither the CPE nor the conditions for the cavity theory can be fulfilled due to the lateral range of the electrons. For a small field when the CPE does not exist, the presence of a detector can change the local level of the CPE, adding more perturbations to complicate the problem.

Under electronic equilibrium, cavity theory describes a method to calculate the dose (D) in a medium based on measured charge in the cavity,

$$D_t = \left(\frac{Q}{m}\right)\left(\frac{\bar{W}}{e}\right)\left(\frac{S}{\rho}\right)_a^t, \tag{1}$$

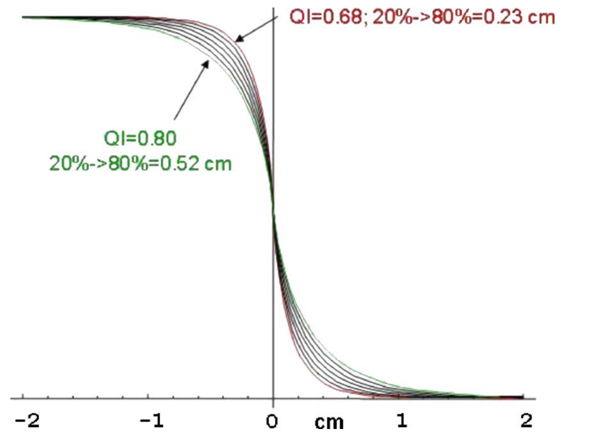


FIG. 3. The graph shows the primary dose profile in water across a collimating edge for different beam energies, specified by their quality index ($\mathrm{TPR}_{20/10}$), ranging from 0.68 to 0.80 at an interval of 0.2. Ideal "spot source" fluence profiles are assumed with only the electron transport that contributes to the penumbra width.

$$\frac{D_t(r)}{D_t(\mathrm{ref})} = \left(\frac{D_t(E,r)}{D_t(\mathrm{ref})}\right)\left(\frac{\bar{W}}{e}\right)_{\mathrm{ref}}^r\left(\left(\frac{S}{\rho}\right)_a^t\right)_{\mathrm{ref}}^r, \tag{2}$$

where Q is the detector reading of charge, m is mass of the air in an ion chamber, W/e is ionization potential of air, r is the field dimension, ref is reference field size, and $(S/\rho)_a^t$ is the mass collision stopping power ratio of tissue (t) to air (a).[28–30] All of the parameters in Eq. (2) are energy dependent; hence, the dose in small fields compared to the reference or calibration field (10×10 cm^2) is uncertain due to spectral variations. The measured ionization readings $Q(E,r)$ are influenced by many factors as shown below:

$$Q(E,r) = Q_m P_{\mathrm{ion}} P_{\mathrm{repl}} P_{\mathrm{wall}} P_{\mathrm{cec}} P_{\mathrm{pcf}}, \tag{3}$$

where Q_m is measured reading, P_{ion} in the ion recombination, P_{repl} is the replacement correction factor, P_{wall} is wall correction factor, P_{cec} is the central electrode correction factor, and P_{pcf} is the perturbation correction factor as described in TG-21 (Ref. 25) and by Nahum.[2] Usually the ratio of these correction factors as shown in Eq. (2) is ignored in routine clinical practices where CPE exists, but they cannot generally be ignored for small fields as noted by Seuntjens and Verhaegen[31] and Sauer and Wilbert.[8] It was observed that for a small volume mini-ion chamber (Exradin A14P) the perturbation factor is larger than unity by about 36%, 30%, and 18% for circular field diameter of 1.5, 3, and 5 mm, respectively. As noted by several other investigators,[8,16,32] these large differences are mainly due to nonequilibrium conditions which are dependent on the type and design of the detector. The magnitude is significantly larger in low-density media. In a heterogeneous medium, these factors (in particular P_{repl}) are not easy to predict because of the uniqueness of measurement conditions in small fields.[33,34] The Monte Carlo simulation has been shown to be an effective tool to study these effects.[33,34] A significant amount of work is still needed to calculate factors for the field size, beam energy, and detector geometry.

The second factor in Eq. (2) is usually ignored since W/e is treated as relatively constant, even though it might depend slightly on beam energy.[35,36] This assumption is based on small changes in the spectrum versus field size. The third factor in Eq. (2) is of significant interest. However, it is often ignored in clinical settings.[37,38] The photon spectrum does change[9,39] with field size and location off axis; however, the stopping power ratio is shown to be relatively insensitive for low-energy photon beams within the accuracy of the measurements ($<1\%$).[9,10,13–15,40–43] However, for high-energy beams it cannot be ignored.[44] With a clear knowledge of the photon and electron spectra in small fields, the terms 2 and 3 in Eq. (2) can therefore be better estimated in the future evaluation. Another measurement-related problem is the experimental determination of field size. When small field conditions prevail, overlapping penumbra cause traditional methods for field size determination based on full width at half maximum (FWHM) metrics to break down, resulting in overestimated field sizes as shown in Fig. 1.

II.E. Corrections and perturbations

The main problem associated with the dosimetry of small fields is the very presence of the detector itself that produces a perturbation hard to quantify in a reliable way.[33,34] This is because the detector is normally different from the medium in both composition and density. The major source of the effect comes from the perturbation of the charged particle fluence, which depends not only on the detector geometry but also on the medium in which the measurement is performed, as well as on the beam energy and field size. Therefore, it is difficult to use standard correction methods in the dosimetric measurement of the small field in heterogeneous media because, in addition to other known corrections,[25] the value of P_{repl} in Eq. (3) is field size- and phantom geometry-dependent.

II.F. Absolute dose

For accurate patient treatment, knowledge of the absolute dose in a reference beam is required. This is performed through TG21 (Ref. 25) or TG-51 (Ref. 27) protocols in North American, and the IAEA protocol[24] or other dosimetry protocols developed in other countries. These protocols provide the methodology to perform dosimetry in a reference field, usually 10×10 cm^2, where the radiological parameters in the reference conditions are available. For some treatment modalities used in SRS treatments, such as Gamma Knife, CyberKnife, and tomotherapy, the reference beam condition (field size of 10×10 cm^2) does not exist. In such a situation there is no simple method to provide absolute or reference dosimetry. It is often indirectly performed by transferring, extrapolating, or intercomparing among various detectors, usually film, TLD, or a small volume ion chamber. The procedure for such a transfer is usually dependent on the choice of an individual physicist and the measurement equipment used. The Radiological Physics Center (RPC) in Houston has undertaken the task of intercomparing doses from participating institutions. Significant deviations in dosimetry from the RPC values have been observed, indicating the seriousness of the dosimetric problem. The accurate dosimetry for small field remains a problem until an appropriate and consistent procedure is available and widely accepted by a majority of users to calibrate nonstandard beams.

II.G. Low-density inhomogeneity

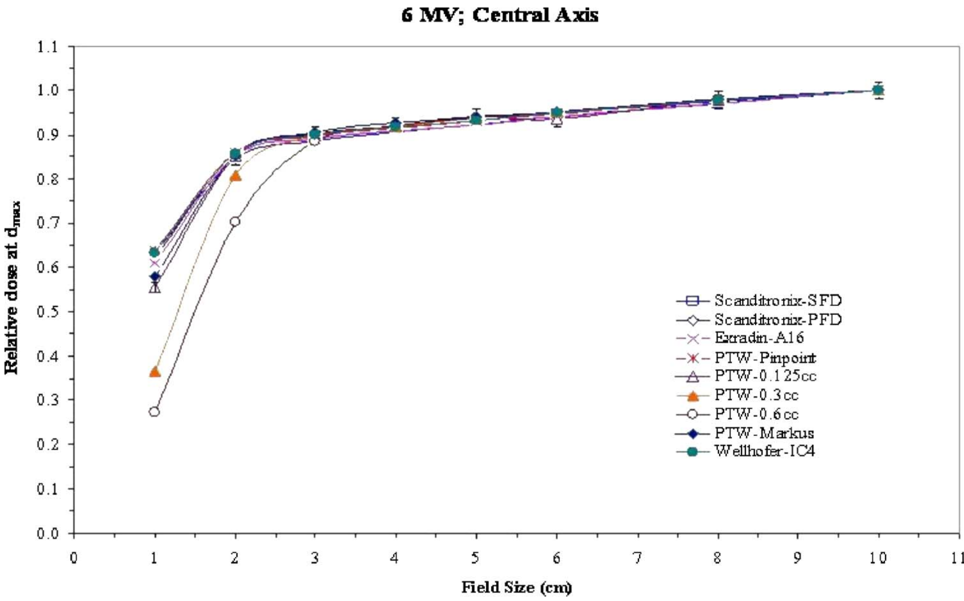
In low-density medium like the lung, small fields are subject to significant perturbations that are energy- and density dependent. Treatment planning algorithms that use simple one-dimensional density scaling fail to provide accurate dose distributions as noted in several publications.[45–51] Advanced treatment planning algorithms in general provide more accurate dose calculations in treatment planning.[34,52–54] However, in clinical trials, a consistent approach in radiation dosimetry is necessary. In order to make the outcome comparison meaningful, new treatment protocols have yet to take advantage of these more accurate treatment planning algorithms, which are now available in many commercial 3D treatment planning systems.

III. PRESENT KNOWLEDGE

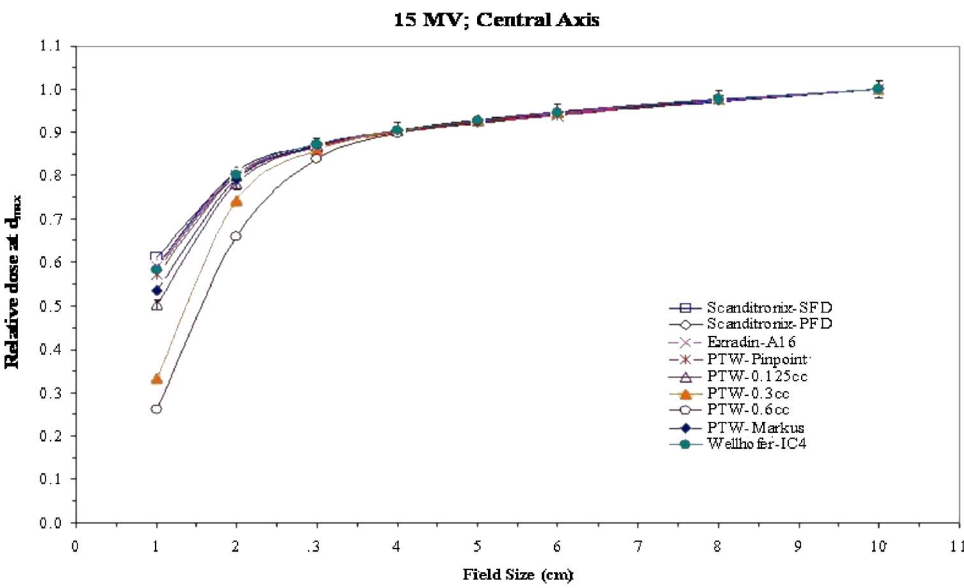
III.A. Experiment

With advances in technology, radiation detectors have evolved and improved in quality. Various manufacturers offer a wide range of radiation detectors including ion chamber, solid-state detector, diode, TLD, scintillator, chemical, Fricke, film, alanine, and others. These detectors can be categorized as standard, mini-, and micro-detectors depending on the sizes $\approx 10^{-1}$, $\approx 10^{-2}$, and $\approx 10^{-3}$ cm^3, respectively.

An assortment of ion chambers and other radiation detectors is available that can be used for a specific task in dosimetry. Ionization chambers have been widely used in radiation dosimetry due to their near independence of energy, dose, and dose rate. They provide a reproducible direct reading and can be calibrated to a national standard to calculate the dose. Ion chambers are relatively inexpensive, readily available, and are manufactured in various shapes (cylindrical, spherical, and parallel plate) and sizes for various applications.[12,17,39,43,55–58] For special procedures such as SRS, Gamma knife, CyberKnife, tomotherapy, and IMRT, the wide availability of radiation detectors questions the selection and proper use of detectors, which has been addressed by various investigators.[7,10,12–15,39,59–67] Figure 4 shows the measured dose at central axis in the form of relative dose, S_{cp}, for 6 and 15 MV beams measured from various detectors. A rapid drop in dose with a certain detector is observed when the field size is decreased. This figure represents only the ratio of reading without any correction, as noted in Eq. (3). The effect is more magnified for high-energy beams, possibly due to a nonequilibrium condition, perturbation corrections as noted in Eq. (3), and volume averaging.[61] With various detectors, the small field produces challenges in dose measurements with a greater probability of significant error.



(a)



(b)

FIG. 4. Total scatter factor (S_{cp}) versus field size measured with various available radiation detectors for (a) 6 MV and (b) 15 MV beams. The data are plotted without the consideration of any corrections or perturbations and simply the ratio of reading in a field size to the reference field.

Semiconductor diode detectors are being widely used for patient dosimetry for both photon and electron beams. Diode detectors have small sensitive volumes and are categorized as mini- and micro-detectors. Characteristics include quick response time (microseconds compared to milliseconds of an ion chamber), excellent spatial resolution, absence of external bias, and high sensitivity. In addition, stopping power ratios for diodes are nearly energy independent, but the presence of low-energy photons causes problems due to the increased photoelectron cross sections in silicon compared to water. The response of the diode detectors depends on temperature, dose rate (SSD or wedge), and energy.[68–70] Depending on the design of the diode, some may have angular dependence as well. In order to achieve the required accuracy recommended by AAPM,[71] these effects should be corrected, or a diode with a minimum dose rate and energy effect should be used. Often diode detector measurements are compared with an ion chamber to provide confidence in small field dosimetry. The stereotactic photon diode (SFD) with nearly micron-size sensitive volume has become an attractive choice for small field dosimetry.[23,72–74]

Diamond detectors are solid-state detectors with large signals and the sensitive volume is relatively small ($1.0-6.0\ mm^3$), which makes them ideal for small field dosimetry and for beam profile measurements.[75] When ionizing radiation is absorbed, it induces a temporary change in the electrical conductivity of the material.[76–80] The response of a diamond detector is directly proportional to the absorbed dose rate. Diamond detectors do not exhibit any directional dependence and they are tissue-equivalent. Diamond detectors do exhibit a small dependence on dose rate, but a correction can be applied for the measurements. The detector is hard to manufacture and hence more expensive than other solid-state detectors.

Thermoluminescent dosimetry (TLD) (Ref. 81) has been used for point dose measurements and *in vivo* dosimetry. The TLD material comes in several different forms, such as rods, chips, and powder. Rods and chips are reusable once they have been properly annealed. TLD exhibits an energy dependence as well as a dose dependence. The accuracy is limited to the irradiation and measuring techniques. It is usually suitable for cross reference of point dose in small fields and IMRT.

Film is used for relative dose measurement. There are two types of films: silver halides[82] and Gafchromic.[83] Silver halide films require processing, whereas Gafchromic films are self-developing. TG-69 provided an overview[82] of silver halide films. Gafchromic film has some superior characteristics; however, its use is limited to relative dosimetry.[84] Even though film exhibits strong energy dependence, it does provide a planar dose maps in small fields[16,18,85] that is superior to other detectors.

Metal-oxide silicon semiconductor field-effect transistor (MOSFET) dosimeters have been investigated for their use in clinical dosimetry[86] and IMRT verification.[87] Due to their small size, MOSFETs are ideal for small field dosimetry in low-density medium,[33] brachytherapy, and *in vivo* dosimetry. The MOSFET detectors are relatively small in size with an active area of 0.2×0.2 mm^2. It is energy independent in MV beams. Also, it is relatively independent of dose rate and temperature. It has been noted that MOSFET dosimeters are similar to conventional dosimeters in reproducibility, linearity, energy, and angular responses.[86] However, the MOSET detector is used mainly for specialized point dose measurements and has a short life. It also requires repeated calibration for accurate dose measurements.

Bang gel detectors[88] are tissue-equivalent and provide a 3D dose map with high spatial resolution. They are energy independent over a wide range of energies, making them ideal for measuring three-dimensional dose distributions. Pappas *et al.*[89,90] provided satisfactory data for small SRS cone with gel detectors. The downside to gel dosimetry is that it takes considerable fabrication time for the proper working conditions. Bang gel readout is based on imaging techniques that are susceptible to imaging artifacts. The new research using MAGIC gel has shown some promising results in its usefulness in dosimetry of SRS and IMRT.[91]

Radiophotoluminescent glass plates have been used successfully to measure dose in SRS and Cyberknife.[13,66,67] Scintillator detectors in various forms have also provided dosimetry in small and elongated fields.[92–94] Alanine pallets have been attempted to measure dose in small fields using the electron spin resonance (ESR) method;[95,96] however, such devices are limited and may not be used clinically except to verify the point dose at a central location or in a national laboratory.

III.B. Theoretical and analytical approach

To derive accurate beam profiles from large volume chamber measurements, deconvolution and extrapolation methods have been used.[59,61,90,97–102] The confidence in dosimetry is greater with larger fields; hence, various mathematical approaches have been proposed for getting either the relative or absolute dose.[103–105] Sauer *et al.*[8] and Cheng *et al.*[72] provided an extrapolation method and mathematical expressions for output in small fields accurately.

III.C. Monte Carlo simulation

Monte Carlo (MC) approaches are rapidly finding a niche where measurements are not possible or rather difficult. Various reports have highlighted the feasibility of MC for small field dosimetry.[9,34,106–110] Monte Carlo simulation provides an opportunity to investigate most of the correction factors as discussed in Eq. (3). It also provides a standard against other techniques for the measurement of relative dose and possibly absolute dose within acceptable accuracy. Monte Carlo simulation also provides the opportunity to understand dosimetry in low-density materials where measurements are difficult with nonequilibrium conditions.[45,46,48–51,54] There are two approaches in using Monte Carlo techniques to improve the accuracy of radiation dosimetry. One is to obtain correction factors for detectors used in the measurements. The other is to calculate the dosimetry quantities directly equivalent to performing a measurement in an ideal condition. However, the use of Monte Carlo needs to be verified with respect to beam modeling parameters (e.g., source size and energy, etc.).

IV. FUTURE DIRECTIONS

IV.A. Absolute dose

There is a growing trend and availability of specialized treatment machines that cannot provide the standard 10×10 cm^2 field for reference calibration as required by standard dosimetry protocols. With advances in treatment devices delivering subcentimeters treatment fields, there is a need for absolute dosimetry protocol. The normal use of some of these machines is to produce dose distributions for targets large enough to establish CPE, but from the superposition of smaller fields. Hence, to calibrate these machines under working conditions representative of their normal operating mode, dosimetry protocols need to be reviewed and formulated to accommodate these new modalities. The International Atomic Energy Agency (IAEA) and American Association of Physicists in Medicine (AAPM) have formed a joint task group to address the absolute dose issue in small fields. As a result, an international protocol will be provided for beam quality specification in small fields, detector-specific correction, and perturbation factors. Any detector-related problem with responses to the superposition of time-separated small field condition subfields should be addressed and investigated so as to provide beam calibration protocols with the same accuracy as present standard beam protocols, i.e., approximately ±2%.

IV.B. Better detectors

Advances in radiation detectors and specialized treatment techniques have fueled the need for better and suitable detectors. Many types of detectors have been used in small fields and cross compared with other detectors;[10,13,14,17,41,43,55,56,62,73,74,84,89,90,93,94,96,111–113] however, the particular perturbation corrections are not known in detail. It is expected that calculation-aided dosimetry will be available where specific correction and perturbation factors are either precalculated for irradiation geometry or calculated online using state-of-the-art radiation transport codes, e.g., Monte Carlo. With improved manufacturing techniques with the emphasis on making reproducible detectors, it is likely that empirical corrections in hardware (e.g., energy-compensated shielding on diodes) will be replaced by calculated correction factors. This type of calculation-aided detector could provide energy, dose, and dose rate independence suitable for small field dosimetry.

IV.C. Treatment planning, beam modeling, and dose calculations

Apart from the accuracy in measured data, the treatment planning systems (TPS) should provide proper modeling to adequately predict the dose distribution in nonequilibrium conditions and with inhomogeneous medium. Properly designed multisource modeling using either accurate measured data[3,4,6] or Monte Carlo derived data, together with accurate dose calculation algorithms that handle non-CPE conditions, will provide acceptable dose distribution. To meet the demand, the TPS vendors should take responsibility for providing suitable implementations of either convolution/superposition algorithms,[26] Monte Carlo, or any other methods known to be capable of handling small field conditions. Collaboration with clinicians and treatment planners will provide planning methodology for lung tumors with respect to the use of margins and immobilization techniques as dense tumors increase CPE and hence dose, compared to the surrounding low-density lung tissue.

IV.D. Monte Carlo techniques and radiation transport calculations

The accurate determination of dose distributions plays an essential role in the success of radiotherapy. Explicit radiation transport calculations based on Monte Carlo or other techniques will increasingly play an important role in nonequilibrium dosimetry. With the aid of calculations in the future, the measurement accuracy should be greatly improved for these cases. It has been shown that, even when the experimental measurements were correct, conclusions drawn from the measured data could be incorrect because the results were misinterpreted due to the lack of knowledge of the perturbation from the presence of a detector.[34,114] In addition, the Monte Carlo simulated beams can be calibrated against measurements under controlled conditions where the measurements can be determined accurately. For example, a Monte Carlo simulated incident megavoltage beam can be calibrated in a reference condition per incident electron at target in the linear accelerator. Since the geometry of the accelerator head can be modeled in detail, the output or the entire realistic beam of any field size can be accurately determined by radiation transport calculations. Therefore, the nonequilibrium dosimetry for the small field in heterogeneous media can be investigated and well understood. The Monte Carlo technique will not only play a crucial role in the small beam output calibration but will also contribute to increased accuracy in patient dose calculations. These types of calculations will play an increasing role in dose verification, beam analysis, and for direct calculation of dose distribution in treatment planning and treatment delivery.

V. CONCLUSIONS

It is expected that the accuracy of small field dosimetry under a nonequilibrium condition can be significantly improved based on the following developments:

- New protocols for absolute dosimetry in small and nonequilibrium condition will be developed to provide procedures for accurate absorbed dose calibrations for new treatment modalities such as Gamma Knife, CyberKnife, tomotherapy, and other devices that do not have standard reference field sizes. This will effectively reduce the uncertainty and variations among different centers in the absolute reference dose calibration.
- Small volume detectors (ion chambers, diodes, and others) will be developed that have minimum perturbations due to its presence and composition. Also, such detectors will have minimum energy, dose, and dose rate dependence.
- The accuracy of small field dosimetry will be greatly improved by Monte Carlo simulations, especially under extreme conditions, for small fields (such as beam size $<3\times 3$ cm^2 with inhomogeneity). Current measurement uncertainty ($>5\%$) will be reduced. Monte Carlo calculations will be able to accurately relate the dose to a small field irradiated in an inhomogeneous media to the dose of reference conditions in which the radiation beam is calibrated. This will ensure that the uncertainty of the dose determination in a small treatment field under extreme conditions remains similar in magnitude to the uncertainty of a large reference field where the beam is calibrated.
- Monte Carlo simulations will provide correction parameters, such as correction factors for specific detectors and for specific measurement conditions, and the stopping power ratio as a function of field size and beam energy. These additional data will be available for routine use and will greatly reduce the measurement uncertainty under the nonequilibrium radiation conditions.
- More accurate implementation of model-based calculation algorithms as well as direct Monte Carlo methods will be available in commercial treatment planning systems for accurate dose calculation under the nonequilibrium radiation conditions.

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