Mass-density compensation can improve the performance of a range of different detectors under non-equilibrium conditions

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Abstract
Dosimeters often consist of several components whose mass densities differ substantially from water. These components cause small-field correction factors to vary significantly as lateral electronic equilibrium breaks down. Even amongst instruments designed for small-field dosimetry, inter-detector variation in the correction factors associated with very small (∼0.5 cm) fields can amount to tens of per cent. For a given dosimeter, small-field correction factors vary not only with field size but also with detector azimuthal angle and position within the field. Furthermore the accurate determination of these factors typically requires time-intensive Monte Carlo simulations. Thus, if achievable, ‘correction factor free’ small-field dosimetry would be highly desirable. This study demonstrates that a new generation of mass-density compensated detectors could take us towards this goal. Using a 6 MV beam model, it shows that ‘mass-density compensation’ can be utilized to improve the performance of a range of different detectors under small-field conditions. Non-sensitive material of appropriate mass-density is incorporated into detector designs in order to make the instruments behave as if consisting only of water. The dosimeter perturbative effects are then reduced to those associated with volume averaging. An even better solution—which modifies detectors to obtain profiles that look like those measured by a point-like water structure—is also considered. Provided that adequate sensitivity can be achieved for a small measurement volume, this study shows that it may be possible to use mass-density compensation (and Monte Carlo-driven design) to produce a
solid-state dosimeter/ionization chamber with a near-perfect non-equilibrium response.

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

1. Introduction

Due to the finite ranges of secondary electrons, lateral electronic disequilibrium occurs at the edges of every megavoltage photon field. Historically radiotherapy treatments utilized photon beams with side lengths of 4 cm or greater, such that a relatively low percentage of the total dose was delivered under non-equilibrium conditions. However, due to the expansion of stereotactic and intensity modulated radiotherapy programmes, non-equilibrium conditions have become increasingly prevalent, and sometimes exist within planning target volumes.

In radiotherapy lateral electronic disequilibrium has long been associated with dosimetric difficulties. For very small fields (\(~0.5\) cm across), point dose measurements obtained using different dosimeters (ion chambers, diodes and diamond detectors) can vary by tens of per cent even for small sensitive volumes of 1–3 mm in diameter (McKerracher and Thwaites 1999, Zhu et al 2000, Sanchez-Doblado et al 2007). Small-field correction factors calculated under the formalism of Alfonso et al (2008) also depend strongly on field size, detector position and detector azimuthal angle (Francescon et al 2011, Underwood et al 2013).

Recent work has indicated that dose area product metrics—measured either using large area detectors or detector arrays—might be used to attain a consistency in small-field QC that is difficult to achieve using point measurements (Djouguela et al 2006, Sanchez-Doblado et al 2007, Underwood et al 2013).

However, in commissioning any small-field system, spatial data (including lateral profiles and percentage depth dose curves) will always be required and should represent the unperturbed dose distribution in water as accurately as possible.

Scott et al (2012) demonstrated that, under non-equilibrium conditions, the water-equivalence of a detector cavity is mainly determined by the mass density of that cavity, rather than its atomic number. The non-equilibrium density effect was further described using cavity theory by Fenwick et al (2013). It is well known that non-sensitive detector components can also influence non-equilibrium dosimetric responses (Bouchard and Seuntjens 2004). Underwood et al (2012) demonstrated that the water-equivalence of various detectors might be improved via minor design modifications based upon the principle of ‘mass-density compensation’. This method was then used in a study of possible diode modifications based on air-gaps (Charles et al 2013), and in this work the ‘mass-density compensation method’ is explored further and applied to diamond detectors and ionization chambers.

2. Methods

2.1. Detector design optimization strategies

The impact of non-equilibrium conditions on a detector’s response is described by the small-field correction factor of Alfonso et al:

\[
k_{\text{f,msr}}=\left[\frac{D_{\text{f,msr}}}{\frac{V_{\text{f,msr}}}{d_{\text{f,msr}}}}/\frac{D_{\text{f,msr}}}{\frac{V_{\text{f,msr}}}{d_{\text{f,msr}}}}\right]
\]  

(1)
in which $D_{w,Q}^{x}$ is the dose to a point of water in field $x$ and $M_{w}^{x}$ is the dosimeter measurement in field $x$ ($x \in \text{clin, msr}$). Here ‘clin’ denotes a clinical field, and ‘msr’ denotes a machine-specific reference field. For an ideal dosimeter we would like to achieve $k_{Q^{\text{clin}},Q^{\text{msr}}}^{\text{f,clin},f,\text{msr}}Q_{\text{clin}},Q_{\text{msr}} = 1$ for all detector positions and orientations, and for all field sizes, shapes and energies.

It is important to bear in mind that one factor influencing $k_{Q^{\text{clin}},Q^{\text{msr}}}^{\text{f,clin},f,\text{msr}}Q_{\text{clin}},Q_{\text{msr}}$ is the finite size of the detector’s sensitive volume: even a hypothetical detector made entirely from water would average the electron fluence gradients over its finite sensitive volume, generating a dose profile that was blurred relative to the ‘point-volume’ profile. In order to separate the effects of the detector media from this volume averaging, we introduce the related term

$$[k_{V^{\text{vol}}}^{D_{\text{vol}}}f,\text{clin},f,\text{msr}]Q_{\text{clin}},Q_{\text{msr}} = \left[\frac{(D_{\text{vol}}^{x})_{w,Q}^{x}}{(D_{\text{vol}}^{x})_{w,Q}^{x}/M_{w}^{x}}\right]_{Q_{\text{clin}},Q_{\text{msr}}}^{\text{f,clin},f,\text{msr}}$$

(2)

where $(D_{\text{vol}}^{x})_{w,Q}^{x}$ is the dose to a finite volume of water identical in geometry to the detector sensitive region.

In this work, Monte Carlo simulations have been used to calculate values of both $k_{Q^{\text{clin}},Q^{\text{msr}}}^{\text{f,clin},f,\text{msr}}Q_{\text{clin}},Q_{\text{msr}}$ and $[k_{V^{\text{vol}}}^{D_{\text{vol}}}f,\text{clin},f,\text{msr}]Q_{\text{clin}},Q_{\text{msr}}$ for several different detectors in a variety of small-field conditions generated using a realistic 6 MV linac beam model. The virtual detector designs have been modified to optimize either $k_{Q^{\text{clin}},Q^{\text{msr}}}^{\text{f,clin},f,\text{msr}}Q_{\text{clin}},Q_{\text{msr}}$ or $[k_{V^{\text{vol}}}^{D_{\text{vol}}}f,\text{clin},f,\text{msr}]Q_{\text{clin}},Q_{\text{msr}}$.

2.2. Applying the optimization strategies to a series of model detectors

The detectors investigated range from simple spherical cavities to complete models of two clinically used dosimeters: the PTW 60003 diamond detector and the PTW 31006 PinPoint air-filled ionization chamber. The detector models and modifications considered are described in the subsections below, whose numbering mirrors that of the relevant Results sections.

The study utilized a square machine-specific reference field of nominal side-length 3 cm—sufficiently large to be free from small-field effects, but sufficiently small to enable relatively fast simulation times. Initially, a square clinical field of nominal side-length 0.5 cm was considered.

2.2.1. Considering spherical cavities of varying mass-density and radius. In order to explore the relationship between cavity mass-density and $k_{Q^{\text{clin}},Q^{\text{msr}}}^{\text{f,clin},f,\text{msr}}$, the mass-density of a water-filled cavity of radius 0.1 cm was increased from 0.001 to 20 g cm$^{-3}$ over a series of nine simulations. The radius of this cavity was chosen to be comparable in size to the sensitive volume of many real detectors. By simulating water of modified mass-density, we were able to isolate the influence of mass-density from effects due to atomic composition. The mass radiological properties of unit-density liquid water were maintained as per the method of Scott et al (2012) i.e. the polarization-effect-correction was not recalculated with changing mass-density.

Next, the impact of cavity radius (volume averaging) upon $k_{Q^{\text{clin}},Q^{\text{msr}}}^{\text{f,clin},f,\text{msr}}$ was investigated. Here $k_{Q^{\text{clin}},Q^{\text{msr}}}^{\text{f,clin},f,\text{msr}}$ was calculated for six spheres filled with unit-density water, their radii ranging from 0.05 to 0.3 cm.

2.2.2. Modifying spherical cavities using non-sensitive shells of contrasting mass-density. Simulations were performed for two spherical cavities of radius 0.1 cm: one made from diamond ($\rho = 3.52$ g cm$^{-3}$) and the other made from air ($\rho = 0.0013$ g cm$^{-3}$). The low density air cavity was modified by the addition of a relatively high density graphite shell
(graphite being a common detector electrode material, $\rho = 1.85 \text{ g cm}^{-3}$), while the high density diamond cavity was modified using a low density air shell.

For each hypothetical detector a range of shell thicknesses - $\delta R$ values - were considered in order to find:

(a) the value of $\delta R$ which gave $k_{0.5, f_{\text{v}}}$
(b) the value of $\delta R$ which gave $[k(D_{\text{vol}})]_{0.5, f_{\text{v}}} = 1$.

The first optimization strategy—(a)—should be considered the ideal: in this case the detector response is made to equate to that of a point-like water structure. However, in the limit where the size of the detector sensitive volume exceeds the size of the field it becomes impractical to compensate for detector volume averaging through design modification. Instead the second strategy—(b)—proves more practical. Here the design of the detector is modified such that its response equates to that of a structure identical in geometry, but constructed entirely from water.

In this study the optimized modifications were tested over: lateral profiles for a nominal 0.5 cm clinical field extending out to an off-axis distance of 1.5 cm, four different clinical field sizes (with side lengths 0.25–1 cm) and three different detector depths (1, 5 and 25 cm).

2.2.3. Varying the half-angle ($\theta$) of the non-sensitive shells. Shells of varying half-angle were also tested, to determine the relative impact of density-compensation-media upstream/downstream from the cavity itself:

2.2.4. Modifying the PTW Diamond 60003 (orientated both vertically and horizontally). The first real dosimeter simulated was the PTW 60003 diamond detector. A virtual model was produced, consisting of a cuboidal block of diamond embedded inside a cylindrical polystyrene case. In the case of our particular detector, according to the individual detector certificate from PTW, the cuboidal diamond block had a height of 0.026 cm and a square front-face with side-length 0.277 cm.

In the diagrams below, the original detector is shown on the left and two different design modifications are shown to the right.

4 Small electrical contacts of course lie below the sensitive volume, but PTW were unable to provide us with any detailed information on this element of the detector design. Nonetheless, an excellent fit to experimental data was obtained for our simple virtual model, as demonstrated in our previous study (Underwood et al. 2013).
The first modification places a density compensating lid (of thickness $\delta L$) on top of the sensitive cavity. The second covers the top and four sides of the sensitive cavity with a density compensating structure (of uniform height and side extension $\delta C$).

2.2.5. Considering the impact of the PTW Diamond 60003 modifications on electron energy fluence. To meet the geometrical constraints of the EGSnrc fluence scoring code (flurznrc), the cuboidal sensitive volume of the real diamond detector was remodelled as a cylinder with the same thickness and cross-sectional area (radius 0.157 cm). The sensitive region was encased within a cylinder of polystyrene, as in the real detector. Above the diamond cavity, a cylindrical lid of height 0.05 cm was built into the model. In the diagram below, region 0 represents the polystyrene case, region 1 represents the ‘lid’ and region 2 represents the diamond cavity.

Three different lid scenarios were considered: (i) a polystyrene lid, such that a near replica of the actual detector was modelled; (ii) an air lid; and (iii) an aluminium lid.

The total electron energy fluence averaged over the volume was scored in both the lid above the cavity and in the cavity itself.

2.2.6. Modifying the PTW PinPoint 31006 ionization chamber. The second real detector simulated was the PinPoint PTW 31006. This ionization chamber consists of a cylinder of steel (the inner electrode), in an air cavity surrounded by a hemispherically-capped-cylinder of graphite (the outer electrode), all coated in PMMA. A detailed virtual model was constructed using specifications provided by PTW (left-hand diagram).

In the modifications considered here (right-hand diagram), the medium of the outer electrode was transformed from graphite ($\rho = 1.85 \text{ g cm}^{-3}, Z = 6$) to aluminium ($\rho = 2.70 \text{ g cm}^{-3}, Z = 13$). Its thickness was also optimized appropriately.

2.3. Monte Carlo methods

A 6 MV Clinac iX accelerator beam model was constructed using the BEAMnrc Monte Carlo system (Rogers et al 2011) and physical machine data provided by Varian Medical Systems. Variance reduction was performed within the flattening filter via directional bremsstrahlung splitting (DBS) using a splitting factor of 1000 and, for all fields of side-length 10 cm or less, a DBS radius of 10 cm. For larger fields, the DBS radius was set to the field side-length. In all cases, DBS electron splitting was also applied and electron range rejection was implemented with varying ECUTRR. BEAMnrc simulations were performed using total electron and photon cut-off energies—ECUT and PCUT—of 0.7 and 0.01 MeV respectively.\(^5\) Phase space files

\(^5\) ECUT includes the rest mass of the electron, 0.511 MeV.
were scored at a distance of 100 cm from the source and were subsequently used as input for the egs_chamber (Wulff et al. 2008) and flurznrc (Rogers et al. 2013) codes. As per the standard output for the EGS codes, all simulated doses were recorded as the dose per incident electron on the linac bremsstrahlung target.

Detectors were modelled within egs_chamber using the EGS++ geometry package. The Monte Carlo beam and PTW detector models were commissioned and validated against measured linac data (Underwood et al. 2013). Within the egs_chamber and flurznrc Monte Carlo codes, global values of 512 and 1 keV were set for ECUT and PCUT respectively (with ECUT = AE, PCUT = PE).

Simulations were performed within a large virtual water phantom located at an SSD of 100 cm. The msr field utilized was a square field of side-length 3 cm. For all msr field simulations the detector/water voxel was maintained on-axis at a depth of 5 cm. Initial simulations were typically run using a square field of side-length 0.5 cm as the small (‘clinical’) field. For all $k_{Qmsr,clin}^{fQmsr,clin}$ values calculated in this study, the ‘point-like’ water structure considered was a $0.25 \times 0.25 \times 0.25$ mm$^3$ water voxel. For water voxels of this size, volume averaging effects were found to be insignificant for fields with side lengths of 0.5 cm or greater i.e. $k_{Qmsr,clin}^{fQmsr,clin}$ remained constant for a $0.5 \times 0.5$ cm$^2$ clinical field when test simulations using smaller point-like structures were performed.

For simulated data, 1 s.d. Type A statistical uncertainties are included as error bars on data plots. For detailed analysis of Type B uncertainties (uncertainties in cross-section, and, for the real detectors, uncertainties in geometrical parameters/material densities) the reader is referred to the work of Francescon et al (2011). For similar dosimeters, Francescon et al determined the impact of Type B uncertainties upon $k_{Qmsr,clin}$ values to be $< 0.7\%$.

3. Results

3.1. Considering spherical cavities of varying mass-density and radius

As shown in figure 1(a), increasing the mass-density of a cavity (of fixed radius and atomic composition) lowers its small-field $k_{Qmsr,clin}^{fQmsr,clin}$ values. In the simulations considered here, $k_{Qmsr,clin}^{fQmsr,clin}$ fell by approximately 15% over the range of mass-densities found in existing detectors, 0.0013 (air)–3.52 g cm$^{-3}$ (diamond).

Increasing the radius of a unit-density water-filled cavity from 0.05 to 0.3 cm resulted in a 30% rise in $k_{Qmsr,clin}^{fQmsr,clin}$.

Trends in $k_{Qmsr,clin}^{fQmsr,clin}$ are almost entirely attributable to the (small-field) numerator of equation (1), the denominator being almost invariant with cavity size and mass density (figure 1).

Together figures 1(a) and (b) demonstrate that whilst increasing the mass-density of a cavity lowers $k_{Qmsr,clin}^{fQmsr,clin}$, increasing the radius of that cavity raises $k_{Qmsr,clin}^{fQmsr,clin}$; two competing effects are at play.

3.2. Modifying spherical cavities using non-sensitive shells of contrasting mass-density

The results of section 3.1 demonstrate that $k_{Qmsr,clin}^{fQmsr,clin}$ can be manipulated by changing the cavity radius or mass-density. In this section we consider the impact of material outside of the cavity: we study the modification of spherical cavities using non-sensitive shells of contrasting mass density.

Figure 2 shows the variation in $k_{Qmsr,clin}^{fQmsr,clin}$ and also $[k(D_{vol})]_{Qmsr,clin}$ with thickness $\delta R$ of additional shell. For both air and diamond cavities, trends in $[k(D_{vol})]_{Qmsr,clin}$ and $k_{Qmsr,clin}^{fQmsr,clin}$ are...
Figure 1. Considering the impact of mass-density and radius on $k_{Q_{0.5},Q_{3}}$ for hypothetical spherical cavities. (a) Dependence of $k_{Q_{0.5},Q_{3}}$ on cavity mass-density (for modified water of various densities and a fixed cavity radius of 0.1 cm). (b) Dependence of $k_{Q_{0.5},Q_{3}}$ on cavity radius (for cavities filled with unit-density water).

Figure 2. Modifying hypothetical spherical detectors of 0.2 cm diameter: exploring the thickness of shell ($\delta R$) required to attain unity in (a) $k(D_{vol})_{Q_{0.5},Q_{3}}$ and (b) $k_{Q_{0.5},Q_{3}}$. Quadratic fits (obtained using linear regression) are also shown.

similar. The thickness of the optimized shells ranged from 70–130% of the radius of the sensitive cavity.

In order to consider their robustness, the optimized modifications of table 1 were applied and tested for four different clinical field sizes and three different detector depths, with results shown in figure 3. The $k_{Q_{0.5},Q_{3}}$ modification to the air-filled cavity (optimized at a depth of 5 cm for a 0.5 cm field) also improves the correction factors associated with both smaller and larger fields plus shallower and deeper depths. However, for the 0.25 cm field, $k_{Q_{0.5},Q_{3}}$ is too high (by 6–8%) compared to the ideal. A thicker graphite shell would have been proposed had the optimization been performed for this field size.
Figure 3. Robustness of modifications to 0.2 cm diameter spheres ($\delta R$ values detailed in table 1) to changing clinical field size and detector depth: (a) $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ and (b) $[k(D_{vol})]_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$.

Table 1. 0.2 cm diameter sphere modifications indicated by a quadratic fitting of 0.5 cm field data.

|                         | $\delta R$ optimized for $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ (cm) | $\delta R$ optimized for $[k(D_{vol})]_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ (cm) |
|-------------------------|---------------------------------------------------------------------------------|---------------------------------------------------------------------------------|
| Air cavity modified using graphite | 0.113                                                                           | 0.069                                                                           |
| Diamond cavity modified using air       | 0.099                                                                           | 0.130                                                                           |

For the diamond cavity, the proposed $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ modification performs well across all field sizes (figure 3). At the standard depth of 5 cm, the deviation of $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ from unity is $<0.7\%$ (around the level of the statistical uncertainties of the simulations), with the performance fairs little worse at shallower/deeper levels. However, the core diamond sphere plus the $[k(D_{vol})]_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ modified shell has a total diameter of 0.46 cm: the modified detector extends significantly beyond the boundaries of the $0.25 \times 0.25$ cm² field. This is problematic in the case of $[k(D_{vol})]_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$: not enough of the compensatory air shell ($\delta R = 0.130$ cm) lies within the boundaries of the field. For the $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ optimization, the required shell is much thinner ($\delta R = 0.099$ cm), hence the $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ modification performs better for the smallest field size.

From figure 3, despite their location within the build-up region of the PDD, the cavities positioned at a depth of 1 cm exhibit very similar correction factor values to those positioned at the reference depth of 5 cm. As the depth of the cavity is increased to 25 cm, the correction factors do vary significantly, however in all cases their values move closer to the ideal. At first glance, this behaviour might be assumed to be due to beam divergence. However, if we consider the data for the basic air sphere, we would expect the nominal (isocentric) field size to increase by a factor of 1.2 between depths of 5 and 25 cm. But if we interpolate to larger

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6 It may be possible to address this problem by changing the modification design: shifting compensatory material from the sides of the spherical cavity to the region upstream.
field sizes on the depth = 5 cm plot (e.g. from an x-axis value of 0.25 to 0.3) we see that the reduction in $k_{\text{f,msr}}^{\text{Q, clin}}$ due to beam divergence alone appears insufficient to account for the low values of the data obtained at a depth of 25 cm. This discrepancy may be attributed to increased photon scatter with depth (blurring the beam profile and widening the penumbra), or variations in energy spectra with depth (Ding and Ding 2012).

Figure 4 shows how both correction factors and calibrated dose vary according to detector off-axis position within a clinical field of side-length 0.5 cm. In the plots of calibrated dose, for each off-axis position the simulated detector meter-reading was multiplied by a simulated detector-specific calibration factor:

$$[\text{Calibration factor}]_{\text{msr}} = \frac{\text{Dose to on-axis point of water}}{\text{Meter-reading from on-axis ‘detector’}}_{\text{msr}}.$$  (3)

The data indicate that modifications to the spherical cavities also prove relatively robust to changing off-axis position.

### 3.3. Varying the half-angle of the non-sensitive shells

For the optimized modifications, figure 5 shows the impact of varying the half-angle of the non-sensitive shells, a complete shell having a half-angle of 180° and a half-shell having a half-angle of 90°.

The largest improvements in $k_{\text{f,msr}}^{\text{Q, clin}}$ optimization are obtained by adding media in the hemisphere directly above the cavity, little consequence being observed when removing
Table 2. Modification strategies for various PTW Diamond detector configurations.

| orientation | Modifications                                                                 | Modifications                                                                 |
|-------------|-------------------------------------------------------------------------------|-------------------------------------------------------------------------------|
| Vertically  | Volume averaging dominates over high density effect: additional dense material | The diamond cavity is modified using air (ρ = 0.0013 g cm\(^{-3}\)) to counteract |
| detector    | is required, detector is modified using aluminium (ρ = 2.70 g cm\(^{-3}\)). | increased density of sensitive volume.                                        |
| Horizontally| Volume averaging is counteracted by increased density: \(k^{0.5, 0.3}_{Q, \text{clin}, \text{msr}}\) = 1, such that no modification is necessary. | Again the cavity is modified using air (ρ = 0.0013 g cm\(^{-3}\)) to counteract the high density of the diamond sensitive volume. |

material from below the cavity. This is as expected due to the strongly forward-peaked nature of secondary electron distributions arising from radiotherapy beams.

3.4. Modifying the PTW 60003 Diamond

For the unmodified PTW 60003 Diamond, initial simulations for fields of nominal side-length 0.5 cm indicated that the requirement for additional high or low density material varied according to whether optimizations were performed for \(k^{0.5, 0.3}_{Q, \text{clin}, \text{msr}}\) or \([k(D_{\text{vol}})]^{0.5, 0.3}_{Q, \text{clin}, \text{msr}}\) and for vertical or horizontal detector orientations. The complex modification strategies for this detector are summarized in table 2.

Considering the vertically orientated diamond detector, a single modification can correct \(k^{0.5, 0.3}_{Q, \text{clin}, \text{msr}}\) to within 5% of unity over a range of detector depths and field sizes down to 0.5 × 0.5 cm\(^2\) (figure 6). However, as the nominal side-length of the field is reduced further—to 0.25 cm—\(k^{0.5, 0.3}_{Q, \text{clin}, \text{msr}}\) rises to ≈1.08 for the modified detector positioned at depths of 1 or 25 cm. Here the 0.277 cm side-length of the sensitive volume exceeds the side-length of the field: 0.25 cm as defined at the isocentre, increasing to 0.2525 cm at a depth of 1 cm and 0.2625 cm at a depth of 5 cm. Considerably more success is achieved in the optimization of \([k(D_{\text{vol}})]^{0.5, 0.3}_{Q, \text{clin}, \text{msr}}\), where the effects of volume averaging are no longer relevant.

The impact of detector depth is more extreme for this diamond detector than for the spherical cavities (compare figure 6 to 3).
Mass-density compensation for improved non-equilibrium dosimetry

In the case of $k_{Q_{\text{clin}}, Q_{3}}^{f_{\text{clin}}, f_{3}}$, the modification which topped the sensitive cavity with a compensatory lid appeared to be more robust to changing field size than did the modification which enclosed the sensitive cavity within a compensatory case. This finding was consistent with the data shown in figure 3(b), where for the hypothetical spherical detector consisting of a diamond sphere surrounded by an air shell the optimization of $[k(D_{\text{vol}})]^{f_{\text{clin}}, f_{3}}_{Q_{\text{clin}}, Q_{3}}$ breaks down as the extent of the compensatory shell exceeds the field size.

For the horizontally orientated diamond detector, the relatively high mass-density of the sensitive region does a near-perfect job of compensating for the volume averaging evident in a $0.5 \times 0.5$ cm$^2$ field (figure 7). However, for a $0.25 \times 0.25$ cm$^2$ field, the compensation is less ideal: $k_{Q_{\text{clin}}, Q_{3}}^{f_{\text{clin}}, f_{3}}$ lies at $\sim 0.96$. Here the cuboidal diamond cavity of width 0.026 cm and length 0.277 cm extends beyond the edges of the $0.25 \times 0.25$ cm$^2$ field such that extreme volume averaging occurs.

The variations in modification performance with changing off-axis position are shown in figure 8.

3.5. Considering the impact of the PTW 60003 Diamond modifications on electron energy fluence

Figures 9(a) and (b) show that under small-field conditions, differences in electron fluence in the lid of the detector model propagate through to differences in the electron fluence in the cavity. Whilst the detector modifications lead to pronounced differences in cavity energy fluence in the case of the $0.5 \times 0.5$ cm$^2$ field, for the $3 \times 3$ cm$^2$ field—where lateral scattering of electrons into the cavity is much more important—the impact of the modifications is relatively minor.

3.6. Modifying the PTW 31006 PinPoint ionization chamber

In the original PTW 31006 PinPoint detector, the graphite outer electrode had a thickness of 0.015 cm. Here, in the modified versions, the total thickness of an alternative aluminium
Figure 7. Horizontally orientated PTW 60003 diamond detector, considering the robustness of the proposed modifications with field size and detector depth. (a) $k_{Q_{0.5},Q_{3}}^{f_{\text{clin}},f_{3}}$. Due to the near unit value of $k_{Q_{0.5},Q_{3}}^{f_{\text{clin}},f_{3}}$ for a square clinical field size of side-length 0.5 cm, in this case the detector was not modified. (b) $k(D_{\text{vol}})^{f_{\text{clin}},f_{3}}_{Q_{0.5},Q_{3}}$.

Figure 8. Robustness of modifications of the horizontally orientated PTW 60003 diamond detector to off-axis position within a square clinical field of side-length 0.5 cm. NB the horizontally orientated detector required no modification for $k_{Q_{0.5},Q_{3}}^{f_{\text{clin}},f_{3}} = 1$. The top-left inserts are expanded versions of the data for detector off-axis positions from $-0.25$ to $0$ cm.
Figure 9. Scoring the total electron energy fluence averaged over the volume of (i) the lid above the diamond cavity and (ii) the cavity itself. In all cases, the centre of the cavity was positioned at a depth of 5 cm. The model detectors were irradiated inside a large model water tank, located at an SSD of 100 cm. (a) 0.5 cm field fluence scoring in the lid (b) 0.5 cm field fluence scoring in the cavity itself (c) 3 cm field fluence scoring in the lid (d) 3 cm field fluence scoring in the cavity itself.

The electrode is quoted as $\delta E$ (in figure 10). In addition to the material transformation, the thickness of the outer electrode was increased by a factor of up to 4.33 in the case of $k_{0.5,0}$. This factor could have been reduced if a conducting material of greater density had been utilized.

The robustness of these proposed PinPoint modifications to changing field size and detector depth is also considered. For the original PinPoint detector, the magnitude of the correction factors is large (e.g. at a depth of 5 cm, $k_{0.5,0} = 1.28$ and $|k(D_{vol})|_{0.5,0} = 1.18$). Although the performance of the modified detectors is not ideal, the resulting correction factors are greatly improved. In a manner similar to the data shown previously (figures 4 and 8) the modifications proved relatively robust to changes in detector off-axis position within a 0.5 cm field.
4. Discussion

Detectors often consist of several components whose mass densities differ substantially from water. It is well known that inserts of non-unit mass-density have the capacity to perturb radiotherapy dose distributions both in dosimeters and in patient treatment plans. However, mass-density compensation has not yet been utilized to improve dosimeter water-equivalence in non-equilibrium conditions.

Small-field/non-equilibrium measurements are strongly influenced by:

(i) the mass density of the sensitive medium;
(ii) the shape and size of the sensitive volume (i.e. its susceptibility to volume averaging);
(iii) the mass density of the surrounding medium.

For a cavity filled with air (or another low-density sensitive medium) both low-density-effects and volume averaging reduce the detector reading obtained on-axis. Thus, if the density of the surrounding medium is modified with the aim of achieving unity in \( k_{Q_{\text{clin}},Q_{\text{msr}}} \), components with \( >1 \) mass-density will always be required.

The situation is different for a detector whose sensitive medium has a mass-density \( >1 \). Here the two effects compete: density acts to increase the on-axis reading, whereas volume averaging acts to decrease it. Thus, depending on the detector geometry, either high- or low-density compensatory material may be required to achieve \( k_{Q_{\text{clin}},Q_{\text{msr}}} = 1 \).

In this work an excellent degree of success was achieved in obtaining unit values of \( k_{Q_{\text{clin}},Q_{\text{msr}}} \) for the modified diamond sphere (figure 3(a)), but the cuboidal geometry of the real PTW diamond 60003 proved more difficult to address (figures 6(a) and 7(a)). For the compensatory shells applied to the air-filled sphere and the PinPoint PTW 31006 the \( k_{Q_{\text{clin}},Q_{\text{msr}}} \) optimization broke down for the smallest field size (side-length 0.25 cm, see figures 3(a) and 10(a)).

In cases where all of the compensatory material was positioned within the field, \( [k(D_{\text{vol}})]_{Q_{\text{clin}},Q_{\text{msr}}} = 1 \) was achieved for both spherical and real detector geometries alike. The process for optimizing detector designs to obtain \( [k(D_{\text{vol}})]_{Q_{\text{clin}},Q_{\text{msr}}} = 1 \) on-axis was straight forward: sensitive media with mass-densities \( <1 \) required additional components with mass-densities \( >1 \); sensitive media with mass-densities \( >1 \) required additional components with...
mass-densities < 1. For a single detector orientation (horizontal or vertical) one modification to the PTW diamond detector 60003 led to water-equivalent performance under a wide variety of non-equilibrium conditions (different field sizes, off-axis positions and detector depths, figures 6(a) and 7(a)). Ionization chambers such as the PinPoint PTW 31006 require considerable quantities of high-density material to compensate for the low density of air. However, provided that the compensatory material is positioned appropriately (within the field boundaries), it is again possible to make the detector behave as though it were constructed from water alone (figure 10(a)).

The simulations considered in this study utilized a maximum (reference) field size of \(3 \times 3 \text{ cm}^2\). If a single modified dosimeter were to be used for both small and large field measurements, its performance under conditions with a greater component of low energy photon scatter (e.g. at the centre of a \(40 \times 40 \text{ cm}^2\) field) would also have to be considered. Under such conditions, the application of mass-density compensation using materials with relatively high atomic numbers may prove detrimental (due to an increased level of photoelectric interactions). Thus a careful balance may have to be achieved between material mass-density and atomic number. Additionally, it is important to note that this study considered a single beam energy: 6 MV.

Our findings explain a number of previous observations in retrospect. For instance, Martens et al. (2001) placed a metallic plate above a liquid-filled ion-chamber array and observed penumbral sharpening (for intensity modulated fields): the performance of their array was improved via mass-density compensation. And the promising results of the IMRT calorimetry probe developed by Renaud et al. (2013) are likely to be attributable to the cylindrical nesting of relatively high-density graphite (1.72 g cm\(^{-3}\)) components with relatively low-density Pyrogel\(^{\oplus}\) (0.17 g cm\(^{-3}\)). The results of our study also add to the evidence that plastic scintillation detectors—with near unit density—may become the non-equilibrium dosimeter of choice (Beddar et al. 2001, Guillot et al. 2011, Morin et al. 2013).

However, even assuming that plastic scintillation detectors fulfill many non-equilibrium dosimetry ideals, it is still likely that demand will exist for multiple detector types. In addition to demonstrating the validity of mass-density compensation as a route to water-equivalence, this study indicates that it may be possible to produce a novel dosimeter (with, for instance, a solid-state or liquid-filled sensitive region) whose small to large field response ratio equates to that obtained for a ‘point-like’ water structure. If such a dosimeter were to be realized, its sensitive volume would have to be carefully designed to meet certain size and geometry constraints (our ‘point-like’ modifications performed better for spherical cavities than for thin slabs).

5. Conclusions

Compensatory material of appropriate mass-density can be used to redress non-equilibrium perturbations arising from the non-unit mass-densities of detector cavities. For existing detectors within 6 MV photon fields, simulations show that near-perfect water-equivalence can be achieved using a single modification, which for a given detector orientation will perform well over a wide range of irradiation conditions (field sizes, depths and off-axis positions). If such an entirely water equivalent detector were to be manufactured, the dose distributions it measured would be perturbed only due to volume averaging which is well understood and could be corrected for on a local basis.

Furthermore, this work shows that for detector cavities of certain shapes and small sizes, mass-density-compensation may be used to manipulate dosimeter response in order to obtain unit small-field correction factors. In this manner, dosimeters can be made to behave like point-like water structures. Thus, provided that adequate sensitivity can be achieved for a
small-sensitive volume, it may be possible to use Monte Carlo-driven design to produce a solid-state dosimeter/ion-chamber with a near-perfect small-field response, eliminating the need for correction factors in small-field dosimetry.

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References

Alfonso R et al 2008 A new formalism for reference dosimetry of small and nonstandard fields Med. Phys. 35 5179–86
Beddar A S, Kinsella K, Ilkilef A and Sibata C 2001 A miniature ‘scintillator-fiberoptic-PMT’ detector system for the dosimetry of small fields in stereotactic radiosurgery IEEE Trans. Nucl. Sci. 48 924–8
Bouchard H and Seuntjens J 2004 Ionization chamber-based reference dosimetry of intensity modulated radiation beams Med. Phys. 31 2454–65
Charles P H, Crowe S B, Kaim T, Knight R T, Hill B, Kenny J, Langton C M and Trapp J V 2013 Monte Carlo-based diode design for correction-less small field dosimetry Phys. Med. Biol. 58 4501
Ding G X and Ding F 2012 Beam characteristics and stopping-power ratios of small radiosurgery photon beams Phys. Med. Biol. 57 5509
Djouguela A, Harder D, Kolhoff R, Rühmann A, Willborn K and Poppe B 2006 The dose-area product, a new parameter for the dosimetry of narrow photon beams Z. Med. Phys. 16 217–27
Fenwick J D, Kumar S, Scott A J D and Nahum A E 2013 Using cavity theory to describe the dependence on detector density of dosimeter response in non-equilibrium small fields Phys. Med. Biol. 58 2901
Francescon P, Cora S and Satariano N 2011 Calculation of $k_{\text{froi,me}}^{\text{Q}_{\text{clin}},\text{Q}_{\text{msr}}}$ for several small detectors and for two linear accelerators using Monte Carlo simulations Med. Phys. 38 6513–27
Guillot M, Beaulieu L, Archambault L, Beddar S and Gingras L 2011 A new water-equivalent 2D plastic scintillation detectors array for the dosimetry of megavoltage energy photon beams in radiation therapy Med. Phys. 38 6763–74
Martens C, Wager C D and Neve W D 2001 The value of the LA48 linear ion chamber array for characterization of intensity-modulated beams Phys. Med. Biol. 46 1131
McKerracher C and Thwaites D I 1999 Assessment of new small-field detectors against standard-field detectors for practical stereotactic beam data acquisition Phys. Med. Biol. 44 2143
Morin J, Beliveau-Nadeau D, Chung E, Seuntjens J, Theriault D, Archambault L, Beddar S and Beaulieu L 2013 A comparative study of small field total scatter factors and dose profiles using plastic scintillation detectors and other stereotactic dosimeters: the case of the CyberKnife Med. Phys. 40 011719
Renaud J, Marchington D, Seuntjens J and Sarfelia A 2013 Development of a graphite probe calorimeter for absolute clinical dosimetry Med. Phys. 40 020701
Rogers D W O, Kawrakow I, Seuntjens J P, Walters B R B and Mainegra-Hing E 2013 NRC user codes for EGSnrc. Technical Report PIRS 702 rev C
Rogers D W O, Walters B and Kawrakow I 2011 BEAMnrc users manual. Technical Report PIRS 509 rev L
Sanchez-Doblado F, Hartmann G, Pena J, Rosell J, Russiello G and Gonzalez-Castao D 2007 A new method for output factor determination in MLC shaped narrow beams Phys. Med. 23 58–66
Scott A J D, Kumar S, Nahum A E and Fenwick J D 2012 Characterizing the influence of detector density on dosimeter response in non-equilibrium small photon fields Phys. Med. Biol. 57 4461–76
Underwood T S A, Winter H C, Fenwick J D and Hill M A 2012 OC-0512 modifying detector designs for small field dosimetry ESTRO 31 Conf. (Barcelona, Spain): Radiother. Oncol. 103(S1) S206
Underwood T S A, Winter H C, Hill M A and Fenwick J D 2013 Detector density and small field dosimetry: integral versus point dose measurement schemes Med. Phys. 40 082102
Wulff J, Zink K and Kawrakow I 2008 Efficiency improvements for ion chamber calculations in high energy photon beams Med. Phys. 35 1328–36
Zhu X R, Allen J J, Shi J and Simon W E 2000 Total scatter factors and tissue maximum ratios for small radiosurgery fields: comparison of diode detectors, a parallel-plate ion chamber, and radiographic film Med. Phys. 27 472–7