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Application of a natural diamond detector for the measurement of relative dose distributions in radiotherapy

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Abstract. The suitability of a natural diamond detector with a special contact system for the measurement of relative dose distributions in selected radiotherapy applications was studied. The performance of the diamond detector was analysed by comparison with measurements using an ionization chamber and a silicon diode detector. The good stability, high response and good spatial resolution of the diamond detector were demonstrated by relative measurements in high-energy photon and electron beams and around the $^{137}$Cs source of an afterloading device. The application of the diamond detectors for relative as well as absolute dose measurements of beta ray ophthalmic applicators was also highlighted. It is concluded that the diamond with the special contact system is a suitable detector for relative dose measurements in a wide variety of applications, especially when high response and high spatial resolution are required.

1. Introduction

Air-filled small-volume ionization chambers and silicon diodes are the detectors which are most often used for the measurement of relative dose distributions in radiotherapy. For relative dose rate measurements in water, the ionization current from an ionization chamber has to be converted to absorbed dose rate using stopping-power ratios (water/air). While this conversion can usually be introduced, the drawback of ionization chambers in some measurements is their finite size. Another drawback in some accurate measurements can be the need to correct for dose rate dependence because of ion recombination. Silicon diodes are smaller in size and, accordingly, have better spatial resolution. However, their main drawback in some cases is their energy dependence for photon radiation below about 100 keV. Silicon diodes may also have a dose rate dependence, unless well prepared and used with suitable electrometers.

Some studies on the use of natural diamond detectors for similar measurements have been published elsewhere (Kozlov et al 1975, Kozlov et al 1977, Planskoy 1980, Konorova and Kozlov 1971, Burgemeister 1983, Keddy et al 1987, Khrunov et al 1990). As diamond is almost equivalent to soft tissue (atomic number $Z = 6$ is close to the effective atomic number of soft tissue, $Z = 7.42$), the detector signal is directly proportional to the absorbed dose rate to tissue and can be used without any correction. The most promising type of diamond detector seems to be the one...
provided with the special contact system proposed by Kozlov (Kozlov et al 1977). It has been shown (see, e.g., Khrunov et al 1990) that this type of diamond detector possesses the most suitable characteristics for relative dose measurements: high response (typically about 0.5 \( \mu \text{C Gy}^{-1} \)), good spatial and time resolution, very small dependence of response on photon or electron energy, good temperature stability (better than about 1% \( ^{\circ}\text{K}^{-1} \)) and good radiation stability (better than 0.05% kGy\(^{-1}\)).

In this study diamond detectors with the special contact system have been tested and compared with other detectors, e.g., an air-filled ionization chamber and a silicon diode detector, for the determination of relative dose distributions in a few selected routine and special radiotherapy applications. The purpose was to prove by practical measurements the suitability and favourable characteristics of diamond, as expected from the published data and other information.

![Figure 1. Construction of the diamond detectors.](image)

2. Materials and methods

The following detectors were used: three diamond detectors (D1, D2 and D3), a thimble ionization chamber (0.1 cm\(^3\) RK chamber), a p-type silicon semiconductor diode (Therados), a parallel plate chamber (NACP chamber for depth-dose measurements in electron beams) and a PTW extrapolation chamber (for depth-dose measurements in the build-up region). The diamond detectors were constructed as described by Kozlov et al (1977). Natural diamond plate of 0.2–0.4 mm thickness and with the special contact system was positioned inside a cylindrical plastic capsule, at a depth of 0.2 mm for detector D1, 0.3 mm for detector D2 and 1.0 mm for detector D3, see figure 1. The material of the capsule was PMMA (D1, D2) and polystyrene (D3). The front window of detector D1 was transparent while the windows of detectors D2 and D3 were opaque to light. The detector bias was 250 V (D1), 150 V (D2) and 100 V (D3). After switching on the high voltage, a pre-irradiation of the detectors up to a dose of about 2 Gy (D1) and 3 Gy (D2, D3) was performed to settle the response of the diamond to a stable level (change of response less than 0.5% h\(^{-1}\) for a constant dose rate). To confirm that the stability of the diamond detectors
after the specified pre-irradiation was adequate for the measurements, irradiation of the diamond detectors in a PMMA phantom in a $^{60}$Co gamma beam was repeated throughout one day, i.e., over a time period exceeding the typical time (a few hours) for a set of measurements. To compare the response of the detectors, all detectors were irradiated under the same conditions in a PMMA phantom in a $^{60}$Co beam. For the measurements of relative dose distributions in a water phantom a CADSCAN radiation beam scanner (Dosetek Oy, Finland) was used. A few measurements were performed with the diamond detectors by connecting them to a Keithley 3561A electrometer.

The following routine or special radiotherapy applications were selected: measurement of relative depth-dose curves and dose profiles in high-energy photon and electron beams, measurement of the depth-dose distribution at the build-up region of high-energy photon beams, measurement of the relative dose distributions around $^{137}$Cs gamma sources of afterloading equipment in intracavitary radiotherapy, and measurement of relative dose distributions of $^{105}$Ru and $^{90}$Sr ophthalmic applicators. The beams or sources for the above were as follows: CGR Saturne 18 MV x-rays and 4 and 20 MeV electrons (scanning beams), Scanditronics Microtron MM 14 14 MV x-rays, CGR Aleyon II $^{60}$Co-unit, Rocus M $^{60}$Co-unit, Selectron LDR $^{137}$Cs afterloading equipment, and Isocommertz type CCC concave $^{106}$Ru applicator and Amersham type SIA concave and planar $^{90}$Sr ophthalmic applicators.

The measurements were carried out in a water phantom except for the build-up region (the second application mentioned above), where a PMMA phantom was used. The measurement of profiles for high-energy photon and electron beams was carried out with the plane of the silicon diode and the diamond plate both perpendicular and parallel to the beam axis. The measurements at the build-up region of the photon beams were compared with measurements carried out with a PTW extrapolation chamber (Pitkänen et al 1985). For the measurements around the $^{137}$Cs source of the afterloading equipment, the source applicator was placed in water, and the beam scanner was used to move the detector around the applicator. For the measurements of the ophthalmic applicators in water, a special holder for the applicator was constructed from water equivalent plastic (RMI solid water).

3. Results and discussion

3.1. Stability and response of diamond detectors

After the pre-irradiation to the dose specified above, the maximum variation of the response of the diamond detectors during the day was found to be less than 1% for detectors D1 and D3 and less than 0.3% for detector D2. For the same dose rate, the signal from the diamond detector was about 40 times (D1, D2) or 4 times (D3) higher than that from the diode. The difference of response between the diamond detectors is due to the differences of the sensitive volumes of the diamond plates. For the same volume of detectors this would mean that the signal from the diamond would be about 160 times higher than that from the diode.

3.2. Measurements in high-energy photon and electron beams

Measurements of profiles in narrow photon beams (field size 1 cm x 1 cm or 2 cm x 2 cm) were performed to compare the spatial resolution of the detectors. The
results by diamond and diode are shown in figure 2 and by diamond and ionization chamber in figure 3. The same results were obtained with both orientations of the diamond and diode relative to the beam axis.

Figure 2. Profiles of a narrow 18 MV x-ray beam as measured by diamond (---) and diode (---). Saturn; field size 2 cm x 2 cm, depth 30 mm, penumbra 4.5 mm, 4.5 mm (diamond); 4.5 mm, 4.5 mm (diode).

Figure 3. Profiles of a narrow 14 MV x-ray beam as measured by diamond (---) and 0.1 cm$^3$ ionization chamber (---). Microtron; field size 1 cm x 1 cm, depth 50 mm, penumbra 4.8 mm, 3.4 mm (diamond); 6.2 mm, 5.5 mm (ionization chamber).
The values of the penumbra (i.e. the distance between relative dose levels of 80% and 20%) obtained with the diamond and diode detectors are in good agreement. This is expected since the size and spatial resolution of the detectors is about the same. The values of the penumbra obtained with the diamond detectors and the ionization chamber show a clear difference: the smaller size and accordingly better spatial resolution of the diamond leads to a smaller value of the penumbra.

Depth-dose curves of a $^{60}$Co gamma beam measured by the different detectors are presented numerically in table 1. At large depths, the relative depth-doses measured by the diode detector exceed by up to 4% those measured by the ionization chamber. The results from the diamond detectors and the ionization chamber are in better agreement, although there seems to be a small relative shift (about 4 mm or 1.5%) between the curves. The former can be explained by the known increase of sensitivity of the diode as photon energy decreases (Rikner and Grussel 1985). The latter could be explained by a small normalization error or inaccuracy in positioning the detectors. Depth-dose curves in the build-up region in a PMMA phantom at $^{60}$Co gamma beam (Rocus M) are shown in figure 4. The curve measured with the diamond detectors agrees fairly well with the ionization curve measured by the extrapolation chamber.

![Figure 4.](image)

Figure 4. Build-up depth-dose curves of $^{60}$Co gamma beam as measured by diamond (---), diode (- - -) and extrapolation ionization chamber (O).

The suitability of the diamond detectors for electron beams was studied by measuring depth-dose curves and profiles in 4-20 MeV scanning electron beams (Saturne). Some results are shown in figures 5 and 6. The depth ionization curve measured by the ionization chamber was converted to the depth-dose curve using stopping-power ratios according to the Nordic recommendations (NACP 1980). No correction was applied to the results using the diamond and diode detectors. The variation of the stopping-power ratio (water/material) as a function of electron energy for the three detector materials is shown in figure 7.
Table 1. Depth–dose curves of $^{60}$Co gamma beam (Aleyon II) as measured by diamond, diode and ionization chamber (field size, 20 cm x 20 cm).

<table>
<thead>
<tr>
<th>Depth (mm)</th>
<th>Diamond</th>
<th>Diode</th>
<th>Ionization chamber</th>
</tr>
</thead>
<tbody>
<tr>
<td>5</td>
<td>100.0</td>
<td>100.0</td>
<td>100.0</td>
</tr>
<tr>
<td>20</td>
<td>95.6</td>
<td>96.3</td>
<td>95.0</td>
</tr>
<tr>
<td>50</td>
<td>83.9</td>
<td>85.9</td>
<td>82.8</td>
</tr>
<tr>
<td>100</td>
<td>64.8</td>
<td>67.0</td>
<td>63.2</td>
</tr>
<tr>
<td>150</td>
<td>48.7</td>
<td>50.9</td>
<td>47.2</td>
</tr>
<tr>
<td>200</td>
<td>36.1</td>
<td>37.9</td>
<td>34.7</td>
</tr>
</tbody>
</table>

In general, the agreement between the profiles and depth–dose curves measured by diamond, diode and ionization chamber is reasonable. As would be expected on the basis of the above, the forms of the depth–dose curves resulting from measurements with the ionization chamber and the diamond detectors are quite consistent. The small relative shift of the curves can again be explained by a small error in the normalization of the curves, or in setting the detectors at the surface at the beginning of the measurements. The form of the curve resulting from measurements with the diode detector is slightly different and resembles that of the depth ionization curve. The correction for the variation of stopping-power ratio (water/silicon, figure 7) would improve the consistency with the other two curves. The results also prove that the fast response of the diamond detectors ensures their proper functioning in scanning beam conditions with relatively large dose per pulse.

The above results indicate that the diamond detector is comparable with the diode detector and can be used for accurate measurements of dose distributions with a high dose gradient in high-energy photon and electron beams.
3.3. Measurements around a $^{137}\text{Cs}$ afterloading applicator

Measurements of dose distributions around afterloading applicators are needed to confirm the accuracy of calculation algorithms which are used for treatment planning. For example, the effects on dose distribution caused by the sources, spacers and the stopping crew of the applicator in a high-dose-rate selectron unit incorporating cobalt sources have been studied by measurements with a diode detector (Pla et al 1987). Similar measurements for low-dose-rate selectron units incorporating caesium sources would be much harder to perform because of a three orders of magnitude lower source activity.

In this study, the suitability of the diamond detectors for measurements in low-dose-rate conditions was demonstrated. The attenuation effect of the stopping crew for one active pellet (about 1.2 MBq $^{137}\text{Cs}$) could be determined with confidence due to the high response of the diamond detectors (figure 8). The study of the dose distribution and the detailed analysis of the attenuation effects will be reported elsewhere.

3.4. Measurements of ophthalmic applicators

Relative dose measurements and absorbed dose calibration of concave ophthalmic beta ray applicators are particularly difficult because of the geometry and the large size of the detectors compared with the source dimensions. Extrapolation ionization chambers and small scintillation or semiconductor detectors have been used (Lax 1991, Pruitt 1987, Schmidt 1977, Davelaar et al 1989). The diamond detector would offer special advantages because of its good equivalence to soft tissue and because it can be made very small while still having a reasonable response. The stopping-power
Figure 7. Ratios of mass collision stopping powers for water and the three detector materials used in this study (ICRU 1989).
where the subscript 'm' refers to the measurements with the applicator, and 'cal' refers to calibration, $M$ is the measured charge from the diamond detector, $p$ is the perturbation correction and $s$ is the stopping-power ratio. The calibration could be performed against absolute measurements with an extrapolation chamber using a ratio of diamond to water changes only about 2% within the interesting range of beta ray energies, i.e., 0.02–5 MeV.

Figure 9 shows the depth–dose as a function of distance from the surface, on the central axis of $^{90}$Sr concave and flat applicators and of a $^{106}$Ru concave applicator, as measured by the diamond detector (D3) or by a diode in water. The results are compared with the depth–dose curves given by the manufacturers of the sources and with the results from studies by some other researchers. For the $^{90}$Sr applicators the measured curves coincide fairly well with those from other publications. For $^{106}$Ru, the measured curve coincides well with a curve recently measured by a diode (Lax 1991), while both of these curves differ clearly from the values given by the manufacturer. The curve by the manufacturer is based on only a few measurements taken using a relatively large scintillation detector (thickness 2 mm), which may explain the difference.

For absorbed dose calibration of concave beta ray applicators, a small calibrated diamond detector could be of special use. In terms of the cavity theory, the absorbed dose to water, $D_{w,m}$, for an ophthalmic applicator would be

$$D_{w,m} = D_{w,cal} \frac{M_m}{M_{cal}} \frac{p_m}{p_{cal}} \frac{(s_{w,c})_m}{(s_{w,c})_{cal}}$$
Figure 9. Relative depth–dose in water as a function of distance from the surface of an ophthalmic applicator. (a) $^{90}$Sr SIA30 flat applicator: 1, Reit et al 1990; 2, diamond; 3, diode. (b) $^{90}$Sr SIA6 concave applicator: 1, Amersham; 2, diamond. (c) $^{106}$Ru concave applicator: 1, type COB (Lax 1991); 2, type CCA (Davelaar et al 1989) (calculated curve); 3, type CCC, Isocommerz; 4, diamond.
corresponding planar applicator. In that case the ratios of perturbation factors and stopping powers could be set to unity to a good approximation. This method is here proposed as a possible further application of the diamond detector and will be studied separately.

4. Conclusions

The advantages of the diamond detectors as reported earlier are (i) very small dependence of response on photon energy over a wide energy interval (better than diodes), (ii) good radiation resistance (better than diodes), (iii) good spatial resolution (comparable with diodes), and (iv) high response (better than diodes). Comparison of diamond detectors with other more conventional detectors, by practical measurements in a few selected routine as well as special applications, has demonstrated the suitability of the diamond for all these applications. After suitable pre-irradiation, the stability of the diamond was very good, and the response was high enough to study dose distributions under low-dose-rate conditions (afterloading equipment). The tissue equivalence and the small size also make the diamond an interesting detector for relative as well as absolute dose measurements of beta ray ophthalmic applicators.

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