The Use of ^{60}Co-sources for Afterloading alternate to ^{192}Ir-sources

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Abstract— **Up to now the production of small sources for HDR afterloading was for technological reasons possible only for ^{192}Ir sources. This causes a dominant prominence for ^{192}Ir sources in HDR afterloading. Since a fairly long time ^{60}Co sources became available with geometrical dimensions identical to those of ^{192}Ir sources. The ^{60}Co sources have the advantage of a longer half time. Therefore, a source change is required in a notably longer interval only. Therewith the problems concerning transport, change and disposal of the sources and quality assurance are reduced. Consequently the introduction of small ^{60}Co sources appears of high interest. For the introduction of ^{60}Co sources the equivalence to ^{192}Ir sources has to be demonstrated. For the same air kerma rate for ^{60}Co an activity is required, which is lower by a factor 2.8 in comparison to ^{192}Ir. For the same dose in water the differences in tissues are so small that ^{60}Co sources can be used alternate to ^{192}Ir sources. Monte-Carlo calculations demonstrate that the radial dose function for the ^{192}Ir source provides higher values than for the ^{60}Co source. Contrary to the ^{60}Co source the anisotropy factor for the ^{192}Ir source deviates entirely at the top of the source and in the region of source mounting from the value 1.00. The comparison is completed by three representative clinical irradiations: The afterloading of carcinoma of bronchus, oesophagus and cervix are examples for linear sources. The combination of ring shaped and linear applicator is used for the cervix carcinoma. The temporary implantation of needles for afterloading of prostate carcinoma represents a more complicated situation. Identical dose distributions result from the investigated ^{60}Co- und ^{192}Ir-sources for these typical situations in brachytherapy. The verification of the alternate usage of ^{60}Co sources to ^{192}Ir sources for HDR afterloading is successfully.**

Keywords: **HDR afterloading, ^{60}Co source, ^{192}Ir source, geometry function, anisotropy factor**

I. Introduction

For the construction of applicators used in interstitial and intracavitary brachytherapy sources with small outer diameter are required. This is mandatory in the usage of flexible applicators with particularly small radius of curvature. Up to now the production of small sources for HDR afterloading (dose rate greater 12 Gy/h) was for technological reasons possible only for ^{192}Ir sources. This causes a dominant prominence for ^{192}Ir sources in HDR afterloading. Since a fairly long time ^{60}Co source became available with geometrical dimensions identical to those of ^{192}Irsources. The ^{60}Co sources have the advantage of a longer half time. Therefore, a source change is required in a notably longer interval only. Therewith the problems concerning transport, change and disposal of the sources and quality assurance are reduced. Consequently the introduction of small ^{60}Co sources appears of high interest. For the introduction of ^{60}Co sources the equivalence to ^{192}Ir sources has to be demonstrated in relation to physical data, source construction, and dose distribution of a single source and clinically applied more complex dose distributions.

II. Materials and methods

Using data from the literature tissue absorption, geometry function and integral dose of the two sources are discussed.

We carried out Monte-Carlo calculations by means of the PC program EGS-Ray (Kleinschmidt) [1] for a water phantom of infinite size. The calculation raster was 1 mm in all three space directions. The gamma radiation was considered only for the calculation because the contribution of electrons can be neglected as result of the shielding by the steel jacket of the source. 10^8 photon histories were evaluated. The cut-off-energy was 10 keV for electrons.

We present three representative clinical irradiations. The afterloading of carcinoma of bronchus, oesophagus and cervix are examples for linear sources. The combination of ring shaped and linear applicator is used for the cervix carcinoma. The temporary implantation of needles for afterloading of prostate carcinoma represents a more complicated situation. Comparative treatment planning was carried out for typical clinical dose distributions using the TPS Plato$^{\text{Nucletron}}$ after implementation of the radial dose function and the anisotropy factor of the ^{60}Co source.

III. RESULTS

From the physical data results that the air kerma rate of ^{60}Co is higher by a factor 2.8 than the air kerma rate of ^{192}Ir, related to the same activity of the sources. For the same air kerma rate for ^{60}Co an activity is required, which is lower by a factor 2.8 in comparison to ^{192}Ir.

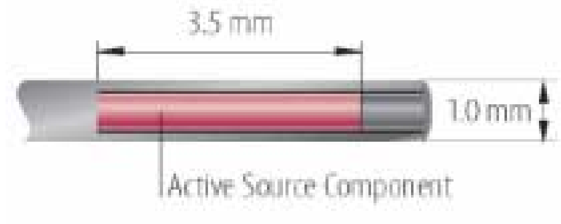

Fig. 1 Structure of the ^{60}Co radiation source from BEBIG

The construction of the ^{60}Co source and the ^{192}Ir source from BEBIG [2] are identical for the afterloading unit BEBIG Multisource. The ^{60}Co source presented in figure 1 consists of a metallic ^{60}Co cylinder of the length 3.5 mm and a diameter of 0.6 mm. The source is surrounded by a cylindrical steel jacket with an outer diameter of 1 mm.

For the use of ^{60}Co sources alternate to ^{192}Ir sources it is essential to know, how the dose values in tissues differ from the dose values for ^{192}Ir sources if for both nuclides the dose values in water do not differ. These differences are presented in figure 2 (data per mille).

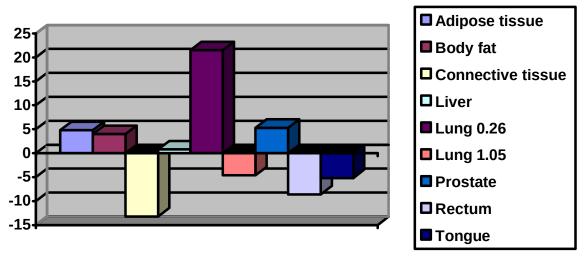

Fig. 2 Differences in absorption (per mille) of several tissues for a ^{60}Co radiation source in comparison to a ^{192}Ir source. Statements for lung are related to the density (gcm^{-3}). Data from Dale et al. [3]

Concerning the integral dose the advisement of Dale et al. [3] is applied which is higher for the ^{192}Ir source.

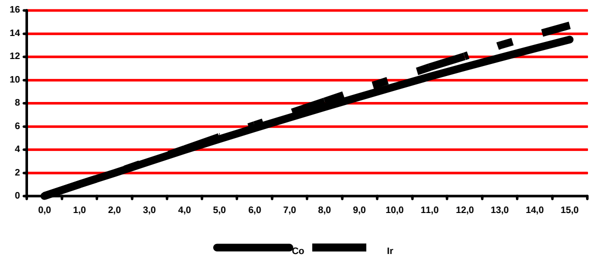

Fig. 3 Integral dose (arbitrary units) as a function of the distance (cm) from the ^{60}Co point source (full) and ^{192}Ir (doted) (calculated using the data of Dale et al. [3]

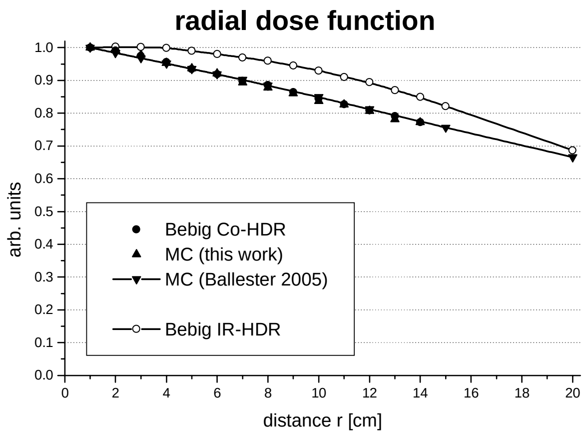

Fig. 4 Radial dose function of linear ^{60}Co sources in comparison to a ^{192}Ir source

The congruence of the results of our Monte-Carlo calculations with Monte-Carlo calculations of Ballester et al. [4] and with the data BEBIG providing for its treatment planning program is evident (figure 4). The radial dose function for the ^{192}Ir source provides higher values than for the ^{60}Co source. At a distance greater than 23 cm the radial dose function of the source provides higher values in comparison to the ^{192}Ir source as a result of the higher energy of the gamma photons of ^{60}Co. This explains the requirement of a higher spatial radiation protection.

The relevant effect of the phantom size is shown in figure 5 which demonstrates the influence of scatter to the dose near the surface of the patient.

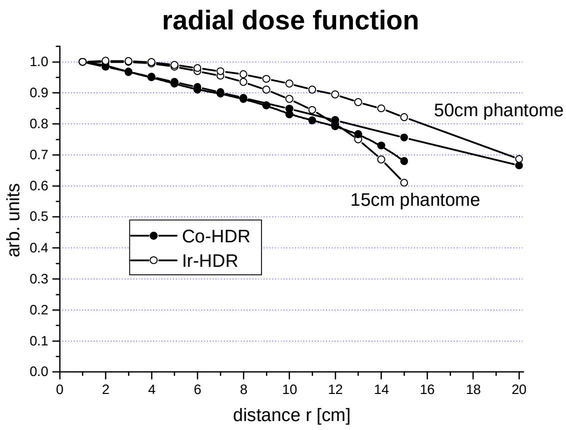

Fig. 5 Influence of the phantom size on the radial dose function of linear ^{60}Co sources in comparison to a ^{192}Ir source

For an ideal point source the anisotropy factor $F(r,\theta)$ is 1.00 independent of the distance r from the source. Contrary for a line source $F(r,\theta)$ depends essential from the angle θ. In practical use a parameterisation of the r dependence is introduced for simplification. Therewith the anisotropy factor $F(r,\theta)$ is reduced to $\Phi(\theta)$. Figure 6 shows $\Phi(\theta)$ for a useable ^{192}Ir source and the ^{60}Co line source from BEBIG. It indicates especially for the ^{60}Co source, that entirely at the top of the source ($\theta = 0°$) and in the region of the source mounting ($\theta = 180°$) deviations exist from the value 1.00.

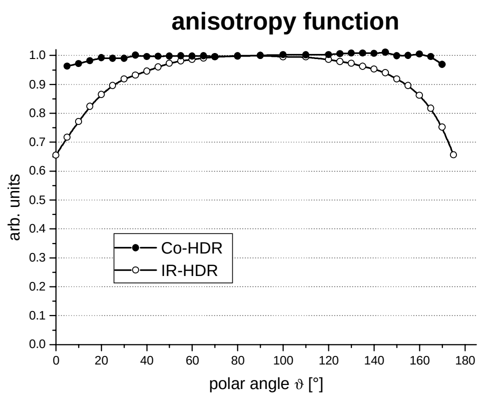

Fig. 6 Anisotropy factor $\Phi(\theta)$ for the ^{192}Ir- and for the ^{60}Co line source from BEBIG (r=4cm)

The presentations in figures 7 to 9 are mirror symmetrical and demonstrate minimal differences between the two nuclides concerning the clinical application. The planning systems for brachytherapy used up to now calculate dose distributions in water equivalent medium. However, figure 1 shows that tissue dependent differences in absorption can be neglected as result of the different medial energies of 375 keV and 1.25 MeV.

In figure 7 the isodoses for a linear applicator near to the source (up to 2 cm) are identical from a practical point of view (upper half Iridium, lower half Cobalt). This is influenced by the normalisation. At the top of the applicator the gradient of ^{192}Ir is steeper because the greater self absorption in the source (smaller value of the anisotropy factor).

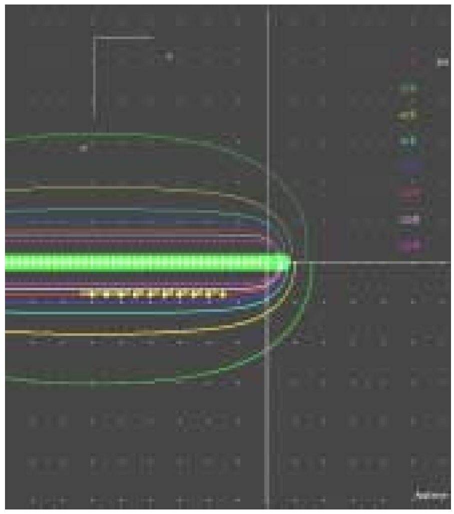

Fig. 7 Comparison of dose distributions of ^{60}Co (inferior half) and ^{192}Ir for linear radiation sources as they are used for the irradiation of carcinoma of oesophagus and bronchus and vaginal irradiations

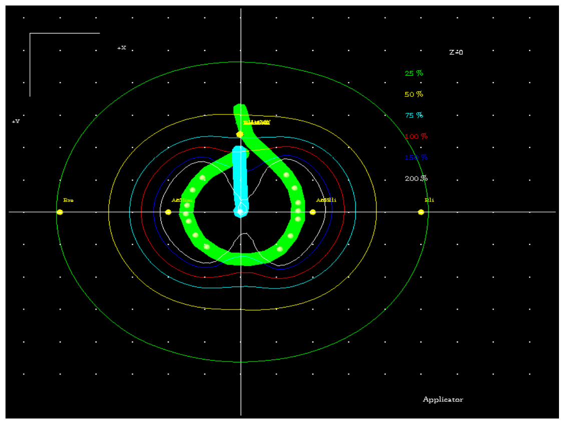

Fig. 8 Comparison of dose distributions for ^{60}Co and ^{192}Ir of a combination of ring shaped and linear applicator for the irradiation of cervix carcinoma (upper half ^{192}Ir, lower half ^{60}Co)

The Combination of ring shaped and linear applicator for gynaecological carcinoma is shown in figure 8 (upper half ^{192}Ir, lower half ^{60}Co. The dose distributions in the transversal section are almost identical for both sources because of the normalisation at the distance 2 cm and the anisotropy factors = 1.0.

Figure 9 shows a comparison of dose distributions of ^{60}Co and ^{192}Ir for afterloading of a prostate carcinoma. The dose distribution of ^{60}Co is mirrored at the median level to get an identical situation for the comparison. The dose distributions differ scarcely.

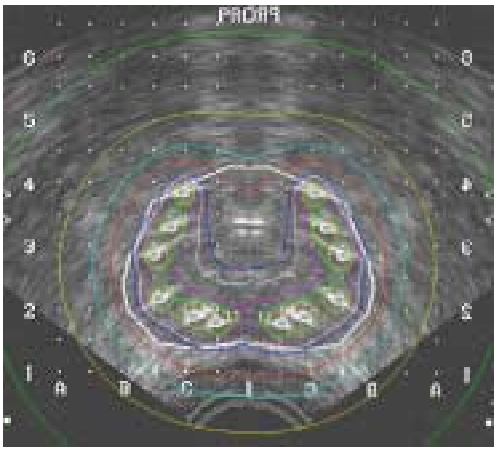

Fig. 9 Comparison of dose distributions of ^{60}Co und ^{192}Ir for afterloading of a prostate carcinoma. The dose distributions in the planning target volume differ scarcely. The dose distributions are normalised to the surrounding 9 Gy isodose (red).

IV. CONCLUSIONS

From physical reasons no disadvantage exists for ^{60}Co sources in comparison to ^{192}Ir sources. There are logistical advantages of the ^{60}Co sources because for the same dose rate only 33 % of the activity of ^{192}Ir sources is needed. Additionally, 25 source exchanges are required for the ^{192}Ir sources before one exchange of the ^{60}Co source.

The construction of ^{60}Co sources can be miniaturized in the same way as it was possible up to now for ^{192}Ir sources only. Herewith the possibilities of application, the forms and diameters of the applicators are identical.

The differences in absorption of gamma radiation in tissues of ^{60}Co sources in comparison to ^{192}Ir sources amount maximally only 2.1 % for lung tissue of low density. This maximal difference is without meaning in clinical situations. Concerning the absorption properties in solid tissues both sources are in practice identical.

At the time of introduction of HDR brachytherapy extensive radiobiological investigations and clinical studies have been necessary to demonstrate the equivalence and advantages as compared with the LDR brachytherapy. For the LDR brachytherapy experience existed over decades because of the radium therapy. The clinical studies regarding this equivalence based on the usage of ^{60}Co sources in a great portion. The present introduction of ^{60}Co sources in miniaturised form is a return to that radiation source which has been used for the validation of HDR brachytherapy.

In total several hundreds of HDR afterloading units equipped with ^{60}Co sources have been used or in use worldwide.

If ^{60}Co sources are used the integral dose in the body of the patient is lower up to about 20 cm distance from the sources for the same dose in the target. However, estimates concerning the relevance for radiation safety aspects include a high grade of uncertainty.

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