Comparison of $^{60}$Cobalt and $^{192}$Iridium Sources in High Dose Rate Afterloading Brachytherapy

Jürgen Richter, Kurt Baier, Michael Flentje

**Purpose:** $^{60}$Co sources with dimensions identical to those of $^{192}$Ir have recently been made available in clinical brachytherapy. A longer half time reduces demands on logistics and quality assurance and perhaps costs.

**Material and Methods:** Comparison of the physical properties of $^{60}$Co and $^{192}$Ir with regard to brachytherapy.

**Results:** Required activities for the same air kerma rate are lower by a factor of 2.8 for $^{60}$Co. Differential absorption in tissues of different densities can be neglected. Monte Carlo calculations demonstrate that integral dose due to radial dose fall off is higher for $^{192}$Ir in comparison to $^{60}$Co within the first 22 cm from the source (normalization at 1 cm). At larger distances this relationship is reversed.

**Conclusion:** Clinical examples for intracavitary and interstitial applications however, show practically identical dose distributions in the treatment volume.

**Key Words:** HDR afterloading · $^{60}$Co source · $^{192}$Ir source · Radial dose function · Anisotropy factor

Betrachtungen zum Einsatz von $^{60}$Cobalt-Quellen alternativ zu $^{192}$Iridium-Quellen

**Ziel:** Seit Neuem gibt es für die Brachytherapie $^{60}$Co-Quellen, die identische geometrische Abmessungen wie $^{192}$Ir-Quellen besitzen. Vorteile sind eine längere Halbwertszeit, durch die logistische Anforderungen, Qualitätssicherung und eventuell Kosten reduziert werden.

**Material und Methodik:** Vergleich der physikalischen Eigenschaften von $^{60}$Co und $^{192}$Ir unter klinischen Gesichtspunkten.

**Ergebnisse:** Die erforderlichen Aktivitäten für dieselbe Luftkerma rate sind für $^{60}$Co um den Faktor 2,8 niedriger, Absorptionsunterschiede in Geweben verschiedener Dichte können vernachlässigt werden. Monte-Carlo-Rechnungen zeigen, dass die Integraldosis infolge der radialen Dosisabnahme für $^{192}$Ir im Vergleich zu $^{60}$Co bis zu 22 cm Abstand von der Quelle größer ist (Normierung bei 1 cm). Bei größeren Abständen kehrt sich diese Beziehung um.

**Schlussfolgerungen:** Klinische Beispiele für intrakavitäre und interstitielle Anwendungen zeigen identische Dosisverteilungen im Behandlungsvolumen.

**Schlüsselwörtern:** HDR-Afterloading · $^{60}$Co-Quelle · $^{192}$Ir-Quelle · Radiale Dosisfunktion · Anisotropiefaktor

**Introduction**

In the past several gamma radiation emitting isotopes have been introduced into clinical brachytherapy, especially $^{137}$Cs, $^{60}$Co and $^{192}$Ir [9, 24, 26].

Henschke and co-workers [6] described the first $^{60}$Co afterloading unit in 1964. Since 1965 the industrially manufactured Cathetron [14] (TEM Instruments) has been distributed. It worked with 9 linear $^{60}$Co sources and was used for the treatment of gynaecological tumours as well as for carcinomas of the bronchus and oesophagus and mould irradiations. The Ralstron [25] was introduced in 1965 by the Hokkaido University Hospital in co-operation with the Shimadzu Company mainly in Japan and Korea. In 1986 146 units of the RAL 303 from Toshiba [28] were in use, 141 of those equipped with $^{60}$Co sources. In 1969 the AGAT V unit [10] with pneumatic movement of $^{60}$Co sources was developed in the Soviet Union and Buchler Company introduced Afterloading Buchler [23] with two fixed and one oscillating Co sources. The GammaMed III [23] of Isotopen Technik Dr. Sauerwein was equipped with 3 channels and offered $^{137}$Cs as well as $^{192}$Ir sources and $^{60}$Co sources. The Selectron [23] of Nucletron used $^{192}$Ir and $^{137}$Cs sources, which moved pneu-

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Received: November 22, 2006; accepted: February 7, 2008
matically as chains of active sources and inactive spacers.

Meanwhile several hundred high-dose-rate (HDR) afterloading units equipped with $^{60}$Co sources or $^{192}$Ir have been put into use. However the dimensions of these sources restricted their application mainly to intracavitory treatment.

Manufacturing of miniaturized high activity $^{192}$Ir sources (dose rate > 12 Gy/h) has been a major breakthrough in modern brachytherapy. The smaller size of the sources and resulting smaller diameter of the applicators allowed also interstitial treatment (flexible applicators and flexible geometry of implants) and individualization of dose distributions by stepping source technology. These obvious advantages shifted market preference to miniaturized $^{192}$Ir sources [16, 27]. Presently also miniaturized HDR $^{60}$Co sources have been made available with geometrical dimensions identical to those of $^{192}$Ir sources.

$^{60}$Co sources have the advantage of a longer half time (5.3 years vs. 74 days). If mechanical stability of the afterloading device is secured over considerably increased treatment cycles, notably less source changes are required. In the following physical basics, source data, dose distributions and clinical applications will be reviewed for $^{60}$Co sources and $^{192}$Ir sources in HDR afterloading technology.

**Material and Methods**

Table 1 contains the maximal energy $E_{\text{max},\beta}$ and the mean energy $E_{\beta}$ of the beta radiation, the range of energy $E_{\gamma}$, the mean energy for the gamma radiation, the half life time $T_{\text{m}}$ the dose rate constant $\Lambda$ and the air kerma rate $K_{\gamma}$ per mCi for $^{192}$Ir and $^{60}$Co.

<table>
<thead>
<tr>
<th>Nuclide</th>
<th>$E_{\text{max},\beta}$ (MeV)</th>
<th>$E_{\beta}$ (MeV)</th>
<th>$E_{\gamma}$ (MeV)</th>
<th>$E_{\gamma}$ (MeV)</th>
<th>$T_{\text{m}}$ (a)</th>
<th>$\Lambda$ (Gy h$^{-1}$ uCi$^{-1}$)</th>
<th>$K_{\gamma}$ (mGy cm$^{-2}$ h$^{-1}$ uCi$^{-1}$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$^{192}$Ir</td>
<td>0.675</td>
<td>0.181</td>
<td>0.136 – 1.062</td>
<td>0.375</td>
<td>73.8 a</td>
<td>1.12</td>
<td>4.13</td>
</tr>
<tr>
<td>$^{60}$Co</td>
<td>0.318</td>
<td>0.096</td>
<td>1.17 – 1.33</td>
<td>1.25</td>
<td>5.27 a</td>
<td>1.09</td>
<td>11.36</td>
</tr>
</tbody>
</table>

$g_{\gamma}(r)$: radial dose function accounting for the effects of absorption and scattering in the medium along the transverse axis of the source,

$F(r, \theta)$: anisotropy factor accounting for the anisotropy of the dose distribution around the source, including the effects of self-absorption,

$S_{\gamma}$: air kerma strength resulting from $S_{\gamma} = K_{\gamma}(d) d^2$, where $K_{\gamma}(d)$ is the air kerma rate in the distance $d$ measured in air.

For an ideal point source the general equation above is reduced to

$$D(r, \theta) = S_{\gamma} \cdot \Lambda \cdot \frac{1}{r^2} \cdot g_{\gamma}(r)$$

because an isotropic dose distribution is produced and the geometry function $g_{\gamma}(r)$ is reduced to $1/r^2$ corresponding to the inverse square law. For an ideal point source a difference between $^{60}$Co and $^{192}$Ir exists only for the radial dose function $g_{\gamma}(r)$. Because of the lower mean energy of $^{192}$Ir scatter near to the source is higher than for $^{60}$Co. Therefore the decline of absorbed dose with distance is less, if both curves are normalized to 1 cm distance from the source. However, when geometry is taken into account the inverse square law dominates over the dose decrease due to absorption and scatter. Using the data of Dale [3] at a distance of 10 cm the relative dose for $^{60}$Co would be 0.86 as a result of radiation interaction. In reality the dose value is only 0.009 due to including inverse square geometry. Therefore, point sources of $^{60}$Co and $^{192}$Ir lead to practically identical radial dose distributions in clinical relevant treatment volumes.

For line sources the clinical dose distribution is additionally to the radial does function $g_{\gamma}(r)$ influenced also by the geometry factor $G_{\gamma}(r, \theta)$ and the anisotropy factor $F(r, \theta)$. $G_{\gamma}(r, \theta)$ is determined by the radioactive material distribution in the capsule, i.e. the length of the source. For an identical construction of both line sources the geometry factors are identically. The specific qualities of the line source determine the radial dose function and the anisotropy factor. These two functions are essential for a comparison of clinical $^{192}$Ir sources and $^{60}$Co sources. For an ideal point source the anisotropy fac-
tor F(r, θ) is 1.00 and independent of the distance of r. In contrast, for a line source F(r, θ) depends upon on the distance, too. In practice parameterisation of the r dependence is introduced for simplification and the anisotropy factor F(r, θ) is reduced to Φ(θ). Φ(θ) in the distance r = 4 cm for the 192Ir source and the 60Co-line source from BEBIG [2] are used for our calculations.

Dimensions and construction of the 60Co source and the 192Ir source from BEBIG are identical and both sources can be used in the same afterloading device (BEBIG Multisource™). For that reason this system was chosen as basis for our calculations. This Co source consists of a metallic 60Co cylinder of the length 3.5 mm and a diameter of 0.6 mm. A cylindrical steel jacket with an outer diameter of 1 mm surrounds the source.

After implementing the parameters described above into the source model we performed Monte Carlo calculations using the PC program EGS-Ray developed by Kleinschmidt [8]. EGS-Ray is a Windows program for a fast modelling of complex geometries. It contains EGS4 with the electron algorithm PRESTA but it is not necessary to program the parts of the desired geometry. Contrary to EGS4 the geometry of the source is introduced by scripts in the form of key words and number combinations. Only gamma radiation emitted from the source was considered because the contribution of electrons can be neglected due to shielding of the steel jacket. 10^8 photon histories were evaluated. The cut-off energy was 10 keV for electrons and photons. Spatial resolution of the calculation for the radial dose functions was 1 mm in all directions.

The influence of phantom size on the dose at larger distances from the source was investigated because the radial dose function g_r(r) is influenced by the phantom size. The missing scatter near air surfaces influences the doses. The radius of the spherical phantom selected for the Monte Carlo calculation was 15 cm and 50 cm respectively. Additionally Monte Carlo calculations were carried out for an infinite water phantom and a calculation grid of 1 cm in all three spatial directions.

Afterloading treatments in carcinoma of bronchus, esophagus and gynaecological tumors [21] are examples for line sources. The combination of a ring shaped and linear applicator is used for cervix carcinoma. The temporary implantation

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**Figure 1.** Differences in absorbed dose in per percent of several tissues for a 60Co radiation source in comparison to a 192Ir source. Statements for lung are related to the density (g/cm³). Data from Dale 1983 [3].

**Abbildung 1.** Unterschiede der Energiedosis in Prozent verschiedener Gewebe für eine 60Co-Strahlenquelle im Vergleich zu einer 192Ir-Quelle. Die Daten für Lunge beziehen sich auf die Dichte in (g/cm³). Daten von Dale 1983 [3].

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**Figure 2.** Anisotropy factor Φ(θ) for the 192Ir and for the 60Co line source from BEBIG (r = 4 cm).

**Abbildung 2.** Anisotropiefaktor Φ(θ) für die 192Ir- und für die 60Co-Linienquelle von BEBIG (r = 4 cm).

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**Figure 3.** Radial dose function of linear 60Co sources in comparison to a 192Ir source.

**Abbildung 3.** Radiale Dosisfunktion von linearen 60Co-Quellen im Vergleich zu einer 192Ir-Quelle.
of needles for afterloading of prostate carcinoma represents a more complicated situation. Comparative treatment planning was carried out for these typical clinical situations using the TPS Plato\textsuperscript{Nuklear} after implementation of the radial dose function and the anisotropy factor of the $^{60}$Co source.

For larger distances (>10 cm) the accuracy of the calculated doses is reduced because the radial dose function implemented in Plato was determined in a cylindrical phantom with a radius of 15 cm.

For dose calculation in clinical brachytherapy water density (dose rate constant $A$) without density correction is used. Due to differences in photon energy absorbed dose in tissues may vary between the two nuclides if the dose values in water are kept identical. A paper of Dale [3] was used for the calculation. He listed in Table I of his paper specific dose constants (unit Gyns$^{-1}$Bq$^{-1}$) for several nuclides and different tissues calculated by the Monte Carlo method. To obtain the differences in absorbed dose in different tissues related to the same dose in water for both nuclides in a first step the dose con-

![Figure 4. Radial dose function for greater distances from the source. Abbildung 4. Radiale Dosisfunktion für größere Abstände von der Quelle.](image1)

![Figure 5. Influence of the phantom size on the radial dose function. Abbildung 5. Einfluss der Phantomgröße auf die radiale Dosisfunktion.](image2)

stants for the different tissues are related to the dose constant for water. In a second step these values for $^{60}$Co are related to values for $^{192}$Ir. The results are plotted in Figure 1.

**Results**

The results of our calculation of the differences between the two sources are presented in Figure 1 (data per percent). For example the dose for $^{60}$Co sources in fat tissue is 0.4 percent higher and for the rectum 0.8 percent lower than for $^{192}$Ir sources. The largest difference is seen for lung tissue of a density of 0.26 gcm$^{-3}$ with 2.1%.

Figure 2 shows the anisotropy factor $\Phi(\theta)$ in the distance $r = 4$ cm for the $^{192}$Ir line source and the $^{60}$Co line source from BEBIG. Especially for the $^{192}$Ir source deviations from unity exist at the top of the source ($\theta = 0^\circ$) and in the region of the source mounting ($\theta = 190^\circ$) demonstrating an advantage of the $^{60}$Co source.

There is good agreement between Monte Carlo calculations of the radial dose function $g_r(\rho)$ in our model and the data provided by other authors [1, 5, 17] and the data implemented in the commercial treatment planning system as seen in Figure 3. This figure contains additionally the radial dose function $g_r(\rho)$ for both nuclides [3]. Especially for $^{60}$Co the minute difference of $g(\rho)$ between sources of different design and an ideal point source is obvious.

As explained above the values of the radial dose function depend on the phantom size. Figure 4 demonstrates the results for both line sources in an infinite phantom. Analogous to the situation in an ideal point source the initial fall off of the radial dose function of the Ir source

![Figure 6. 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). Abbildung 6. Vergleich der Dosisverteilungen für $^{60}$Co und $^{192}$Ir eines Ring-Stift-Applikators für die Bestrahlung eines Zervixkarzinoms (obere Hälfte $^{192}$Ir, untere Hälfte $^{60}$Co).](image3)
is less steep compared to Co. The relationship reverses at distances greater than 22 cm as result of the higher photon energy of $^{60}\text{Co}$.

This phenomenon explains the apparent paradoxes that although the integral dose in the patient is slightly lower, higher room shielding is required for use of $^{60}\text{Co}$ sources in brachytherapy.

The remarkable effect of the phantom size is shown in Figure 5. Especially near the phantom surface of the spherical phantom with the radius 15 cm a reduction of the radial dose function is obviously.

Figures 6 and 7 demonstrate minimal differences between dose distributions the two nuclides in clinical application. The planning system Plato calculates dose distributions in water equivalent medium. However tissue dependent differences in absorption as result of the different mean energies of 375 keV and 1.25 MeV can be neglected as shown above.

In Figure 6 the dose distributions for a combination of a ring shaped and linear applicator in the transversal section are almost identical for both sources because of normalisation at 2 cm and the anisotropy factor $= 1.0$. The dose distribution in the upper half concerns to $^{192}\text{Ir}$ and the lower half to $^{60}\text{Co}$. In the oblique section the influence of the self absorption (smaller value of the anisotropy factor) at the end of the applicator is demonstrated.

Figure 7 shows a comparison of $^{60}\text{Co}$ and $^{192}\text{Ir}$ dose distributions in HDR afterloading of a prostate carcinoma (technique as in [4]). Anatomy and source positions are mirrored in the median level to show identical conditions for dose calculation. No differences are seen in the isodose distribution between 15 Gy (blue line) and 2 Gy (green line).

The practically identical dose distribution for the investigated $^{60}\text{Co}$- and $^{192}\text{Ir}$-sources is also demonstrated in Figure 8 by the cumulative dose volume histogram for the prostate case in Figure 7.

**Discussion**

Miniaturization of $^{60}\text{Co}$ sources similar to $^{192}\text{Ir}$ sources motivated us to review the relationship between the physical characteristics of these isotopes and their properties in clinical brachytherapy. Following the introduction of HDR brachytherapy extensive radiobiological investigations and clinical studies had been performed to determine its effectiveness and toxicity compared to low dose rate (LDR) brachytherapy based on decades of experience from radium treatment [15, 20, 22]. The need for fractionation and reduction of total dose (if single doses of 6–10 Gy were used) was clearly established.

The clinical studies proving biological equivalence were mainly done in uterine cancer using relatively large $^{60}\text{Co}$ sources and intracavitary applicators. Several hundreds of HDR afterloading units equipped with $^{60}\text{Co}$ sources have been used or are in use worldwide.

Miniaturized $^{192}\text{Ir}$ sources opened the possibility of smaller and flexible applicators extending HDR afterloading to interstitial treatment and optimization by variation of dwell times. Dosimetric and theoretical studies supported clinical equivalence [3, 7, 11, 13, 17].

$^{60}\text{Co}$ and $^{192}\text{Ir}$ sources of identical shape and construction show practically identical dose distributions despite definite differences in physical characteristics. This results from the overwhelming effect of geometry in brachytherapy and the rather small differences in radial dose distribution and the
negligible differences in density dependent tissue absorption between these sources. The maximal differences in absorption of gamma radiation from $^{192}$Ir or $^{60}$Co sources are 2.1% for lung tissue of low density. In solid tissues both sources lead to practically identical dose deposition.

However, the typically used treatment planning systems are based on phantoms of homogenous water density and a radius of 15 cm. Our investigation describe additionally to Ballester et al. [1] the influence of phantom size upon measured and calculated dose distributions. Finite phantoms underestimate dose deposition at larger distances from the source. This effect is more pronounced for iridium than for cobalt as shown by our simulations. Looking at typical treatment situations in the pelvis and head and neck there is no treatment relevant impact and no influence on the dose distribution in axial direction. However, in Z-direction dose deposition is higher than predicted from conventional planning systems. It is unknown whether this low dose contribution is of relevance in terms of secondary late effects. Estimates concerning the relevance for radiation safety aspects include a high grade of uncertainty.

Our Monte Carlo simulations show that for the same dose in the target the integral dose is lower for $^{60}$Co sources within a radius of about 20 cm compared to $^{192}$Ir. Beyond this distance the ratio reverses gradually due to the higher photon energy of $^{60}$Co. This higher energy requires more room shielding for the application of the $^{60}$Co source.

No advantage or disadvantage exists for $^{60}$Co sources compared to $^{192}$Ir sources with regard to clinical aspects. There are potential logistical advantages of $^{60}$Co sources because only 33% of the activity of $^{192}$Ir sources is needed to yield an equivalent dose rate. Using typical intervals for replacement due to decay, 25 source exchanges are required for $^{192}$Ir for one exchange of the $^{60}$Co source resulting in reduced operating costs. However, these comparisons are only valid if mechanical stability of the afterloader and of the source capsule allows extended use of such a magnitude.

Up to now such small $^{60}$Co line sources are available from the BEBIG Company only and require the afterloader from this company. But as a result of the logistical advantages other companies will follow.

References


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